2019

Examining the influence of muscle fatigue on knee joint mechanics during an athletic cutting task

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EXAMINING THE INFLUENCE OF MUSCLE FATIGUE ON KNEE JOINT MECHANICS DURING AN ATHLETIC CUTTING TASK

By

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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

Windsor, Ontario, Canada

2019

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ABSTRACT

The purpose of the current study was to examine the influence of lower limb muscle fatigue on the mechanics of the knee joint during an athletic cutting task. A biomechanical methodology was utilized to examine 12 recreationally active females, who cycled through a fatigue-inducing protocol, using a slideboard, followed by the performance of five maximal cuts, until fatigue resulted in trial termination. 3D motion capture was utilized to capture full body movements and changing joint angles of the hip, knee and ankle during the weight acceptance of the cutting maneuver. A force plate was used to record the ground reaction forces of the participants during weight acceptance of the athletic cut. Lastly, surface electromyography monitored the muscle activity of nine muscles on the dominant leg of the participants. Repeated measures ANOVA ($p<0.05$), with Tukey's significant post hoc test, was used to determine significance of the main effect of time on the measured variables. Analysis of the kinematic data revealed that, as fatigue progressed, hip and knee flexion angle significantly decreased during weight acceptance. Kinetic data revealed that peak anterio-posterior shear force significantly increased, and medial-lateral impulse of force significantly decreased, as participants progressed through the fatiguing protocol. Finally, surface electromyography data showed an overall significant decrease in muscle activation from the beginning to the end of trial, however, further investigation of pairwise comparisons indicated that, from 60-100% of the trial, muscle activation significantly increased. This work contributes to the body of work concerning exercise induced muscle fatigue and provides further insight into the underlying mechanism of acute injury during heightened fatigued states. The knowledge gained from this study can be used to advise and improve training prescription and monitoring strategies.

AKNOWLEDGEMENTS

First, I would like to express my sincere gratitude to my outstanding mentors: Jim Potvin, Joel Cort and Chad Sutherland. Each of you have provided me with continuous support, immense knowledge, and guidance during my time in Windsor. I have nothing but great memories from our times together, and I could not have imagined a better team to work with.

I would like to thank Don Clarke, for always being available to help me with any problem we had in the lab, and for his amazing ability to fix things, sometimes by just walking into the room. As well, I would like to thank Joe Kagumba and Aidan Potvin for their help during data collections, and their constant entertainment in the lab.

I thank my thesis committee for their insightful comments and time, but also for challenging me to approach my research from various perspectives and adding more layers to this thesis.

To all of my roommates at 727, I am so grateful for the motivation, emotional support and the most fun times. Especially to Lauren, Tré, Kim, and Katie, who have all listened to my woes on countless occasions and encouraged me to get to the finish line.

To John, you've been there through every emotion I have felt during this journey. This experience would not have been the same without you, and I am so lucky and thankful for your unfailing support.

Finally, I must express my very profound gratitude to my amazing parents, sister, grandparents and all my friends and family who stuck by my side through my years in Windsor. Even though we were far in distance, the continuous love and support I received always made me feel close to home. This accomplishment would not have been possible without them. Thank you.

Sara Santos

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LIST OF ABBREVIATIONS

ACL: Anterior Cruciate Ligament

A/D: Analog to Digital

ANOVA: Analysis of Variance

BF: Biceps Femoris

CMRR: Common Mode Rejection Ratio

ED: Effective Duration

GL: Gastrocnemius Lateral

GM: Gastrocnemius Medial

GR: Gracilis

H:Q: Hamstring to Quadriceps

Hz: Hertz

Kg: Kilogram

LCL: Lateral Collateral Ligament

MCL: Medial Collateral Ligament

MUAP: Motor Unit Action Potential

MVE: Maximal Voluntary Exertion

N: Newton

PCL: Posterior Cruciate Ligament

RF: Rectus Femoris

ROF: Rating of Fatigue

sEMG: Surface Electromyography

SIMM: Software for Interactive Musculoskeletal Modeling

ST: Semitendinosus

TA: Tibialis Anterior

vGRF: Vertical Ground Reaction Force

VL: Vastus Lateralis

VM: Vastus Medialis

CHAPTER 1

INTRODUCTION

1.1 THE PROBLEM

The knee joint is the most common site of injury by athletes, accounting for about 60% of all sports related injuries (Gage, McIlvain, Collins, Fields, & Comstock, 2012; Ingram, Fields, Yard, & Comstock, 2008). Of these, 45% involve injury to the anterior cruciate ligament (ACL) (Majewski, Susanne, & Klaus, 2006). There is consensus in the literature that approximately 70% of ACL injuries in sport happen in non-contact situations, which occur without a direct blow to the knee (Boden, Scott, Feagin, & Garrett, 2000; Sharir, et al., 2016; Shimokochi & Shultz, 2008; Olsen, Myklebust, Engebretsen, & Bahr, 2004). These may involve rapid dynamic activities and multidirectional knee loadings. One of such activities, and arguably the most risk inducing, is an athletic cutting maneuver, which involves a rapid deceleration and change in direction by the athlete. In addition, there is evidence that injury in sport occurs most frequently at the end of matches, and that injury rate increases as each division of the game (half, quarter, period) progresses. Though the reasoning may be multifactorial, this suggests the influential role that fatigue may play with regard to injury (Price, Hawkins, Hulse, & Hodson, 2004).

Mechanical failure, whether it be to the ACL or any other structure, living or not, occurs when the strength capacity of the structure is exceeded by the demand or stress placed upon it (Besier, Lloyd, & Ackland, 2003). The plant and cut movement present risk to the ACL about all three axes of rotation, making it the most common injury inducing action in sport (Hughes & Watkins, 2006). These multi-axial risk factors include increased knee extension, internal rotation, and knee valgus, described as femoral adduction and knee abduction (Laughlin, Weinhandl, Kernozek, Cobb, & Keenan, 2011). The dynamic valgus position is the primary predictor of ACL injury risk, as it pulls the ACL at its distal attachment point on the tibia, in an anterior-medial direction, and twists it medially away from its proximal end (Hewett, et al., 2005).

Fatigue, defined as “a temporary decline in the force and power capacity of skeletal muscle resulting from muscle activity” (Potvin & Fuglevand, 2017), may interfere with lower extremity kinetics and kinematics (McLean, et al., 2007), neuromuscular control and dynamic stabilization (Liederbach, Dilgen, & Rose, 2008), force capacity and contractile rate (Potvin & Fuglevand, 2017), and proprioceptive function (Miura, et al., 2004). By definition, as the muscle fatigues, its ability to generate contractile force decreases and, therefore, energy absorption may also be reduced (Mair, Seaber, Glisson, & Garrett, 1996). This leaves the joints passive stabilizers with a larger demand to dissipate energy, exposing them to a higher risk for failure.

Though the literature is lacking in its demonstration of the effects of fatigue when performing an athletic cut, there is evidence showing its detriments during other exercises and athletic tasks. During vertical jumping, as fatigue progresses, knee flexion angles decrease during landing (Chappell, et al., 2005). Landing forces, with a more extended knee, are significantly increased and have less range to dissipate the impulse over, resulting in higher peak impact forces at the knee (Laughlin, Weinhandl, Kernozek, Cobb, & Keenan, 2011; Kernozek & Torry, 2005). Anterior tibial translation and shear force significantly increases by over 30% with quadriceps fatigue, which is known to stress the ACL and increase the risk of ligament injury (Ireland, 2002; Chappell et al., 2005). Increased valgus moments, and decreased knee flexion angles, have also been demonstrated during landings of stop-jump tasks when fatigued (Chappell, et al., 2005). In single leg drop landings, a task with similar components to cutting, fatigue induced significant increases in hip and knee extension, hip internal rotation, knee abduction and ankle plantar flexion angles, as revealed by Borotikar et al. in 2008. These positions not only strain the structures of the knee joint but, compromise the ability of the hamstrings and quadriceps to lengthen optimally and oppose external knee abduction loads (Borotikar, Newcomer, Koppes, & McLean, 2008). Each of these findings are representative of the many different ways in which fatigued athletes may be at an increased risk of non-contact ACL injury.

Males and females demonstrate different susceptibilities with respect to risk of knee injury. Research has shown that females are 2-8 times more likely to sustain an ACL injury than their male counterparts (Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2009). The reasoning for the increased incidence in female athletes may be multifactorial, including risk factors related to anatomical, hormonal, neuromuscular, environmental and biomechanical differences (Hewett, Lindenfeld, Riccobene, & Noyes, 1999; Gage, McIlvain, Collins, Fields, & Comstock, 2012). Some of the hypothesized anatomical reasons for the increase in female risk include: a more narrow intercondylar space; a wider pelvis, creating a larger Q angle, defined as the acute angle between the line connecting the anterior superior iliac spine to the middle of the patella, and the line connecting the tibial tuberosity to the center of the patella; increased flexibility of ligaments, muscles and tendons due to hormone differences; and a generally lower strength capacity of women (Hewett, Myer, & Ford, 2006; Hughes & Watkins, 2006). Altered neuromuscular control strategies and movement patterns are also likely to contribute to the increased incidence of injury in females during exercise. More specific to the interests of this study, decreased knee flexion angles and increased knee valgus are more pronounced in female athletes when performing an athletic cut (Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001; Pollard, Davis, & Hamill, 2004; Sigward & Powers, 2006). Quadriceps dominance in females is another phenomenon contributing to the imbalanced injury ratio between sexes. This is the general finding, that females activate the quadriceps to a greater extent than males during cutting, and other athletic tasks (Lephart, Ferris, Riemann, Myers, & Fu, 2002; Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2009; Sigward & Powers, 2006). The quadriceps pull the knee into extension and, as a result, translates the tibia anteriorly, placing stress on the ACL. Each of these risk factors, in combination with the suggested anatomical risk factors, create a high-risk profile for ACL injury in female athletes.

Despite all that is known about the separate risks associated with both cutting maneuvers and muscle fatigue, the literature is limited regarding their combined effects on athletes and the physically active population. The mechanical response of knee joint to fatiguing muscles during the plant and cut maneuver remains unclear including: which of the muscles are most fatigable and, how the fatigue states of muscles compromise the safety of the joint during complex tasks.

1.2 THESIS OBJECTIVE

This study examined the influence of lower limb muscle fatigue on the mechanics of the knee joint during an athletic cutting task. Specifically, participants performed a fatigue-inducing protocol, using a slideboard, followed by the performance of five maximal cuts, and then returned to the slideboard. This cycle continued until fatigue threshold was exceeded, which in this study was defined as a 20% decrease in the maximal cut distance during the step out phase of the cut. This study aimed to provide more insight into the relatively shallow research pool related to the effects of fatigue on knee kinetics and kinematics when executing a cutting task. The knowledge gained may be further used to develop better training programs and sport-specific techniques to help prevent risk of ACL injury, as well as strengthen the appropriate muscles which best accommodate the negative neuromuscular and movement patterns which accompany fatiguing exercise.

1.3 HYPOTHESES

1) Participants would demonstrate a decrease in knee flexion angles and increase in knee valgus angles during the weight acceptance phase of the cutting task as fatigue progressed.

2) Participants would experience greater vertical ground reaction forces and effective duration decreases as fatigue progressed.

3) Fatigue progression would have a significant effect on normalized sEMG for RF, VL, VM, BF, ST, GL, GM, GR and TA.

CHAPTER 2

LITERATURE REVIEW

2.1 EPIDEMIOLOGY

For athletes, and the physically active population, the knee joint is a common site for injury, accounting for about 60% of all sports related surgeries (Gage, McIlvain, Collins, Fields, & Comstock, 2012; Ingram, Fields, Yard, & Comstock, 2008). Injury to the ACL is involved in about 45% of all knee traumas (Majewski, Susanne, & Klaus, 2006). Knee injury in sport may result from a contact mechanism, which is defined as resulting from a direct blow to the knee, or a non-contact mechanism, which occurs in absence of direct contact to the knee (Olsen, Myklebust, Engebretsen, & Bahr, 2004). Approximately 70% of ACL injuries in sport, reportedly, occur in non-contact situations, which may involve rapid dynamic activities and multiplane knee loadings such as sudden deceleration, directional changes, jump landings, pivoting and cutting maneuvers (Boden, Scott, Feagin, & Garrett, 2000; Sharir, et al., 2016; Shimokochi & Shultz, 2008). As a result of these injuries, athletes often are unable to compete for the remainder of the season and are likely to require invasive surgical procedures and rehabilitation, which may further limit their participation (Ruiz, Kelly, & Nutton, 2002; Lohmander, Östenberg, Englund, & Roos, 2004). About 90% of individuals who sustain an ACL injury in the USA eventually undergo ACL reconstructive surgery and, as a result, are at a heightened risk of subsequent injury as well as knee osteoarthritis and loss of functional ability (Paterno, Rauh, Schmitt, Ford, & Hewett, 2012). Economically, athletes are burdened with not only the cost of reconstructive surgery, but the treatment of complications, subsequent knee surgery, physical therapy and outpatient visits if they wish to attempt to return to play or try and maintain the same quality of life as before the injury. According to 2010 USA Census data for the age group of 15-24 years, the cost of ACL reconstruction and physical therapy totaled over $32000 (adjusted to 2015 USD) per patient (Stewart, Momaya, Silverstein, & Lintner, 2016). This cost does not include that associated with lost time at work or school, or alternative transportation needs. After all this, the rate of surgical success, with respect to return to play rate at the pre-injury level, is only 60% (Gobbi, Mahajan, Zanazzo, & Tuy, 2003). In addition to the negative effects such an injury may have on the individual's future in sport, potential scholarship opportunities, mental health and day-to-day livelihood are also placed at risk, further emphasizing the potential burden to recreational and professional athletes (Freedman, Glasgow, Glasgow, & Bernstein, 1998; Noyes, Matthews, Mooar, & Grood, 1983).

2.2 KNEE ANATOMY

By drawing attention to the anatomical structures of the knee, the internal and external factors which place stress on it and the mechanisms in which it can be injured may be more easily understood.

The knee joint is located between two of the longest weight bearing bones in the human body, the femur and tibia, and is comprised of two joints. The first is the patellofemoral joint which articulates between the anterior surface of the femoral plateau and the posterior surface of the patella. Second is the tibiofemoral joint, which articulates between the distal end of the femur and the proximal end of the tibia. This joint is held together by both passive and dynamic stabilizers (Hughes & Watkins, 2006). Passive stability comes from non-contractile structures, such as ligaments, which guide and limit joint motion (Daniel, 1991). These include the lateral and medial menisci, lateral collateral ligament (LCL), medial collateral ligament (MCL), posterior cruciate ligament (PCL) and the anterior cruciate ligament (ACL) (Figure 1). Dynamic stability is provided by the muscles crossing the joint, primarily from the quadriceps and hamstrings muscle groups. As a result of their large anatomical moment arms, these muscles possess the ability to support large external loads, as well as reduce the loading on the ligaments (Lloyd & Buchanan, 2001).

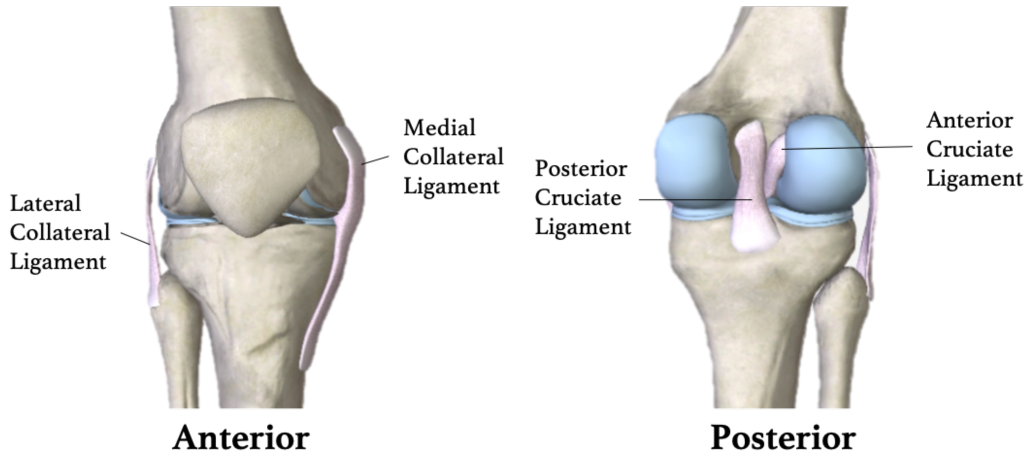

Figure 1: Anterior (left) and posterior (right) views of passive structures of the right knee (BioDigital, Inc., 2018).

The primary role of the ACL in joint stability is to resist anterior translation and internal rotation of the tibia with respect to the femur (Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007). Along with the PCL, the ACL is located in the intercondylar notch where it attaches to the posterior medial aspect of the lateral femoral plateau, and the posterior aspect of anterior intercondylar area of the tibial table (Hughes & Watkins, 2006).

The muscles surrounding the knee joint work together to provide dynamic stability (Lloyd & Buchanan, 2001). For the purpose of this study, the focus will be on vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), biceps femoris (BF), semitendinosus (ST), gracilis (GR), gastrocnemius medial (GM), gastrocnemius lateral (GL), and tibialis anterior (TA) (Figure 2).

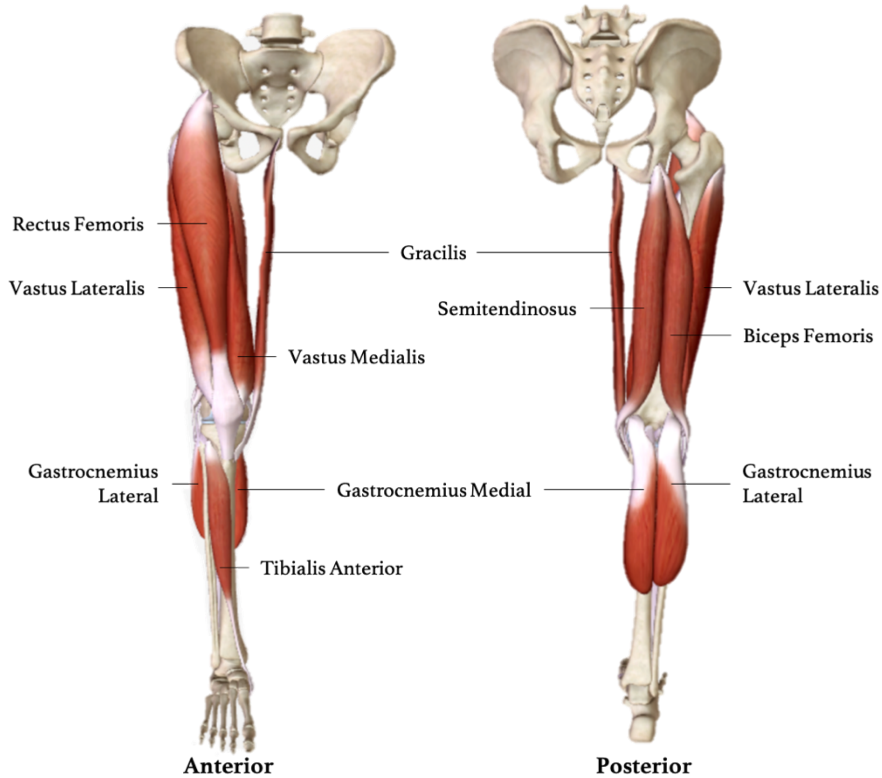

Figure 2: Anterior and posterior view of the muscles of the right lower extremity (BioDigital, Inc., 2018).

The vastus lateralis, vastus medialis and rectus femoris all belong to the quadriceps muscle group. Primarily, these are responsible for the extension of the knee. In addition to knee extension, the RF assists in hip joint flexion. The VL muscle originates on the lateral surface of the greater trochanter of the femur, VM on the distal part of the intertrochanteric line of the femur, and the RF on the acetabular roof of the hip joint. All have a common insertion point on the tibial tuberosity of the patellar ligament. The VM and VL additionally insert on the medial and lateral condyles of the tibia, respectively (Schuenke, Schulte, & Schumacher, 2010).

The hamstring muscles are antagonistic to the quadriceps muscles. Biceps femoris, and semitendinosus belong to this group. Each of these muscles originates on the ischial tuberosity of the pelvis and the sacrotuberous ligament, as well as the linea aspera on the femur for BF short head. The BF inserts onto the head of the fibula, and the ST to the upper part of the medial surface of the tibia. These muscles are responsible for knee joint flexion, as well as hip joint extension and stabilization of the pelvis in the sagittal plane. Knee external rotation and internal rotation moments are provided by the BF and ST, respectively (Schuenke, Schulte, & Schumacher, 2010).

The gracilis muscle is located medial to the femur. It originates on the inferior pubic ramus below the symphysis and inserts onto the medial border of the tibial tuberosity. The GR muscle assists in hip adduction, knee internal rotation and flexion of both the hip and knee joint (Schuenke, Schulte, & Schumacher, 2010).

The gastrocnemius muscles belong to the triceps surae muscle group. The GM and GL originate from the medial and lateral epicondyle of the femur, respectively. Both heads insert onto the Achilles tendon. These muscles are responsible for plantar flexion and supination of the ankle, and also assists with knee joint flexion (Schuenke, Schulte, & Schumacher, 2010).

Lastly, tibialis anterior is located in the anterior compartment of the tibia. This muscle originates at the upper lateral surface of the tibia, the crural interosseous membrane, and the superficial crural fascia. It inserts at the medial and plantar surface of the foot. Tibialis anterior functions to dorsiflex and supinate at the ankle (Schuenke, Schulte, & Schumacher, 2010).

2.3 KNEE INJURY MECHANICS

Mechanical failure, whether it be to the ACL or any other structure, living or not, occurs when the strength capacity of the structure is exceeded by the demand, or stress, placed upon it (Besier, Lloyd, & Ackland, 2003). Injury to the ACL is likely multifactorial, with no single factor being the sole cause for the increased risk of injury (Ford, Myer, & Hewett, 2003). The strength, or capacity, of the ACL may be dependent on variables such as hormone levels, loading history, and prior injury (Besier, Lloyd, & Ackland, 2003). The non-contact mechanism of knee injury commonly involves a deceleration before a change in direction or landing with the knee between 20° of flexion and full extension (Ford, Myer, & Hewett, 2003). During such maneuvers, the position of the lower extremities increases the stress and, thus, the demand, on the structures of the knee, particularly the ACL. Cadaveric studies show that combined applied moments of flexion, valgus and internal rotation place the greatest amount of stress on the ACL (Besier, Lloyd, & Ackland, 2003). These risk factors all describe the events which occur during an athletic cutting maneuver.

The plant and cut movement is the most common action being performed at the time of injury (Hughes & Watkins, 2006). Changes in the alignment of the lower limbs during this movement occur about all three axes, with increasing knee extension, internal rotation, knee abduction and knee valgus (Laughlin, Weinhandl, Kernozek, Cobb, & Keenan, 2011). During weight acceptance, increased peak valgus moments may place an athlete at higher risk of injury due to increased loads on the ACL (Dempsey, et al., 2007). The dynamic valgus position pulls the ACL at its distal attachment point on the tibia in an anterior-medial direction, as well as twists it medially away from its proximal end. Knee internal rotation and abduction moments, through logistic regression analysis, have been identified as significant predictors of ACL injury. These postures contribute directly to dynamic valgus, the primary predictor of ACL injury risk (Hewett, et al., 2005).

The deceleration and change in direction, involved when cutting, requires a significant amount of quadriceps activation. When the quadriceps contract at flexion angles less than 45°, the tibia translates anteriorly and increases the strain on the ACL. This, in addition to the external loading, may increase the risk of injury to the ACL (Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2009). Hamstring activity, working in an agonistic manner, aids the ACL in decreasing anterior shift of the tibia, and thus decreases the stress on the ACL (Simonsen, et al., 2000). Co-contraction of the hamstrings and quadriceps is proposed to protect the knee joint not only from anterior drawer, but also from dynamic valgus (Hewett T. E., et al., 2005).

2.4 EFFECTS OF FATIGUE

Fatigue is defined as “a temporary decline in the force and power capacity of skeletal muscle resulting from muscle activity” (Potvin & Fuglevand, 2017). As fatigue sets in, it may interfere with many functions of the body during exercise such as lower extremity kinetics and kinematics (McLean, et al., 2007), neuromuscular control and dynamic stabilization (Liederbach, Dilgen, & Rose, 2008), force capacity and contractile rate (Potvin & Fuglevand, 2017), and proprioceptive function (Miura, et al., 2004). When the muscle is activated, it can absorb a significantly larger amount of energy. However, as the muscle fatigues, its ability to generate contractile force decreases, and, therefore, energy absorption may also be reduced. One study demonstrated that the reduction in energy absorption decreased by 25% in the fatigued limb when compared to baseline measures (Mair, Seaber, Glisson, & Garrett, 1996). This leaves the joints passive stabilizers with a larger demand to dissipate energy, exposing them to a higher risk of failure, suggesting that fatigue may be an important factor in muscle and knee injury.

There is little research on how muscle fatigue effects body kinetics and kinematics during the cutting maneuver specifically, however, there is evidence showing its detriments during other exercises and athletic tasks, such as: vertical jumping, rapid run and stop maneuvers and single leg drop landings. In vertical jumping tasks, landing softly significantly decreases landing forces (Laughlin, Weinhandl, Kernozek, Cobb, & Keenan, 2011). Greater knee flexion may protect the ACL by allowing for the dissipation of energy over a larger range of motion (Kernozek & Torry, 2005). As the quadriceps muscles fatigue, however, the range of motion during landing decreases due to lowered capacity of the muscle to eccentrically contract and stop the body from downward movement. During rapid run and stop maneuvers, anterior tibial translation showed increases of over 30% during a fatigued state (Ireland, 2002). This action is known to stress the ACL and increase the risk of ligament injury. These findings are in agreeance with a later study by Chappell et al. (2005), where participants had significantly increased peak proximal tibial anterior shear forces after completing a fatigue protocol consisting of consecutive vertical jumps and sprints. In addition, participants also demonstrated increased valgus moments and decreased knee flexion angles during landings of stop-jump tasks when fatigued (Chappell, et al., 2005). In a study by Gehring et al. (2009), using leg press as a fatiguing protocol and drop landing as a post-measure, hamstring and gastrocnemius activation was significantly reduced, and indicated a reduction in the active muscle control of the knee joint after fatigue (Gehring, Melnyk, & Gollhofer, 2009). These muscles act in support of the ACL and, thus, after fatigue the potential for additional ACL loading was increased.

Another study, by Borotikar et al. (2008), investigated the adverse effects of muscle fatigue on single legged drop landing kinematics. Similar to cutting, this exercise involves deceleration and the reduction of momentum using only one limb. That study observed significant increases in hip and knee extension, hip internal rotation, knee abduction and ankle plantar flexion angles when fatigue was induced. These postures are not only more stressful to the knee joint but place the hamstrings and quadriceps further from their optimal lengths for force production and compromise their ability to oppose external knee abduction loads (Borotikar, Newcomer, Koppes, & McLean, 2008).

2.5 SEX DIFFERENCES

Males and females differ with respect to risk of knee injury. Landry et al. (2009) found that females are 2-8 times more likely to sustain an ACL injury than their male counterparts. In a 2007 survey of 100 US high schools, females with ligamentous knee injuries required surgery twice as often as males and were 50% more likely to sustain season ending injuries (Fernandez, Yard, & Comstock, 2007).

The reasoning for the increased incidence in female athletes is multifactorial and includes risk factors related to anatomical, hormonal, neuromuscular, environmental and biomechanical differences (Hewett, Lindenfeld, Riccobene, & Noyes, 1999; Gage, McIlvain, Collins, Fields, & Comstock, 2012). One of the hypothesized anatomical contributors to the increase in female risk includes a narrower intercondylar space, possibly causing increased lengthening of the ACL under tension (Hewett, Myer, & Ford, 2006). Another is that females have, in general, a wider pelvis. This creates a larger Q angle, defined as the acute angle between the line connecting the anterior superior iliac spine to the middle of the patella, and the line connecting the tibial tuberosity to the center of the patella. Larger Q angles allow for greater knee valgus angles and moments. This is known to stress the ACL as supported by findings from Bendjaballah et al. (1997), indicating that the load on the ACL may increase up to six times with a 5° increase in knee valgus. Females also exhibit increased flexibility of ligaments, muscles and tendons due to hormone differences. This may decrease the tensile strength of the ACL, as well as decrease the amount of passive stability to the joint. Finally, a generally lower strength capacity of women decreases the contractile strength that the muscles can apply to provide dynamic stability to the joint. The muscle stiffness of females can range between 55.8-73.9% for the quadriceps, when compared to males. This may increase the dependence on passive structures to maintain stability at the knee, increasing the risk for injury (Hughes & Watkins, 2006).

Altered neuromuscular control strategies and movement patterns are also likely to contribute to the increased incidence of injury in females. As demonstrated in a side cutting task, Malinzak et al. (2001) found knee flexion angles were generally lower than males by about 15° as well as increased knee valgus angles, by 11°, which was consistent throughout the entire movement, when compared to males. These findings are consistent with others which suggest that females demonstrate greater knee valgus and smaller knee flexion angles than males during cutting tasks (Pollard, Davis, & Hamill, 2004; Sigward & Powers, 2006). Females are identified as being quadriceps dominant, such that quadriceps activation occurs as the initial response to injury mechanism perturbations and selected athletic maneuvers (Lephart, Ferris, Riemann, Myers, & Fu, 2002). The general finding that females activate the quadriceps to a greater extent than males during cutting is widely agreed upon in the literature (Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2009; Sigward & Powers, 2006). In support of this, for example, one study by Malinzak et al. (2001), found the normalized quadriceps EMG of females to be 17-40% greater during running and side cut maneuvers than that of males. As previously stated, the quadriceps pull the knee into extension via connection to the tibia through their insertion on the patellar tendon. This as a result translates the tibia anteriorly, causing anterior drawer, directly placing stress on the ACL. Each of these risk factors, in combination with the suggested anatomical risk factors, create a high-risk profile for ACL injury in female athletes.

2.6 PROTOCOLS

2.6.1 SLIDEBOARD EXERCISE

The slideboard exercise is a closed kinetic chain (CKC) exercise, meaning that the foot meets external resistance with the surface during muscle contraction. By using this form of exercise, proprioception about all axes, through Golgi tendon organ and muscle spindle stimulation, as well as dynamic stabilization, through the co-contraction of muscles, can be improved without placing unnecessary stress on the ACL (Heller & Pincivero, 2003; Bunton, Pitney, Kane, & Cappaert, 1993). The co-contraction of these muscles during this exercise is also suggested to benefit the eccentric strength of the quadriceps and hamstrings muscle groups (Blanpied, et al., 2000). Heller and Pincivero conducted a study in 2003 which evaluated the EMG activity of the quadriceps and hamstring muscles during the slideboard exercise. Participants performed six sliding cycles, on a slideboard set at approximately double the length of the lower extremity, measured from the anterior superior iliac spine to the medial malleolus. The cadence was chosen by the participant as a comfortable pace. The results of this study revealed that the VM, VL, TA and medial hamstring muscles showed the greatest amount of EMG activity, followed by the GM and lateral hamstrings. These activation patterns are similar to that of an athletic cut, whereby the VM, VL, GL GM, BF and ST demonstrate the highest EMG amplitudes (Branch, Hunter, & Donath, 1989; Bencke, Næsborg, Simonsen, & Klausen, 2000).

2.6.2 QUANTIFYING FATIGUE

An early study by Viitasalo et al. (1993), aimed at understanding the effects of fatigue during continuous jumping drills, used several biomechanical parameters to quantify the changing state. Male volleyball players were assessed while completing continuous hurdle jumping using EMG on the muscles of the dominant leg, three-dimensional ground reaction forces from force plates, an electrical goniometer, and video footage for unilateral digitization of body segment position. There were significant differences between the first and last series of their fatiguing protocol in average force ($p<0.05$), contact time ($p<0.05$) and time to peak angular velocity ($p<0.01$) during the concentric phase of hurdle jumping. Though not calculated by the researchers, it is important to note that these variables may be used to calculate the average power during the concentric phase ($Power = \vec{F} \cdot \vec{v}$). Based on this particular study, it may be assumed that because there are significant decreases in the average force (N) and increases in the time to takeoff (ms), that there would be a significant decrease in the power during the concentric phase of the hurdle jump as the muscles fatigue.

Micklewright et al. (2017) introduced a new method of measuring perceived fatigue, called the Rating-of-Fatigue (ROF) scale. This ROF scale incorporates numerical, descriptive as well as diagrammatic components. Using empirical data, the alignment of these components was determined. Physiological, performance and psychophysical measurements were also provided in order to calculate a correlation for each measurement against the ROF components. Significant correlations were found between all tested physiologic, performance and psychophysical constructs, some of which included rating of perceived exertion, heart rate, power output and time to exhaustion. The highest correlation with the numeric ROF was the performance measurement of power output ($r=0.992$, $p<0.0001$). There was also some correlation between the ROF scale with recovery, as monitored 30 minutes past the point of volitional exhaustion. The ROF of participants, at the time of volitional exhaustion, ranged from an 8 to a 10. This research team was successful in building a ROF with high face validity that is comprehensive and easy to use (Micklewright, St Clair Gibson, Gladwell, & Al Salman, 2017).

CHAPTER 3

METHODS

3.1 PARTICIPANTS

Twelve healthy females (mean (SD): age 23.4 (1.6) years, height 1.64 (0.05) m, body mass 62.7 (10.9) kg, H:Q 53.2 (4.7) %), were recruited to participate in the study (Appendix A). Participants were recreational athletes with minimal to no prior experience using a slide board as a form of exercise. Participants were excluded from the study if they had sustained either an acute or chronic injury to the lower extremity in the last two years, or if their Hamstring to Quadriceps (H:Q) strength ratio was below 45% as measured on the Biodex (Leuty, 2016). Participants were asked to sign a letter of consent (Appendix B) to participate in the study after a detailed description of methodology was given.

3.2 INSTRUMENTATION AND DATA ACQUISITION

Nine channels of sEMG were used to record electric activity of Vastus Medialis (VM), Vastus Lateralis (VL), Rectus Femoris (RF), Biceps Femoris (BF), Semitendinosus (ST), Gracilis (GR), Gastrocnemius Medial (GM), Gastrocnemius Lateral (GL) and Tibialis Anterior (TA) on the dominant leg (Figure 3). For each muscle, a pair of disposable surface electrodes (Medi-trace, Graphic Controls, Gananoque, ON) was placed along its line of action between the myotendinal junctions and innervation zones, with an inter-electrode distance of 3 cm.

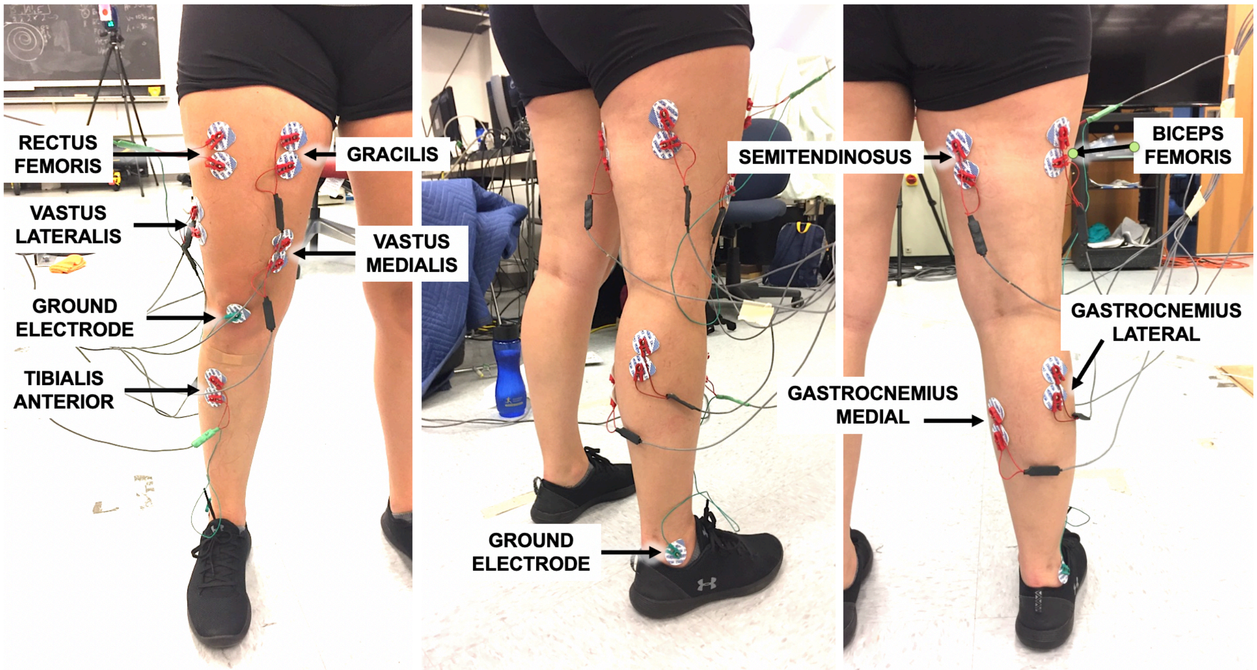

Figure 3: Anterior (left) lateral (middle) and posterior (right) view of sEMG electrode placement on the dominant leg (Appendix C).

Electrode placement (Appendix C) for each muscle was as follows, in correspondence with the protocol used by Cashaback & Potvin (2012), with the addition of GR and TA: VM (one fifth of the distance from the medial tibial plateau to the anterior superior iliac spine, or 2 cm medial to the superior rim of the patella); VL (3 to 5 cm above the patella, on an oblique angle just lateral to the midline); RF (half of the distance between the anterior superior iliac spine and the superior pole of the patella); BF (two thirds of the distance from the greater trochanter to the back of the knee); ST (half the distance between the gluteal fold and the posterior aspect of the knee, approximately 3 cm from the medial border of the thigh); GM (approximately 2 cm medial to the midline of the leg, over the bulge of the muscle belly); GL (one third the length of the distance from the head of the fibula to the tuberosity of the calcaneus on the heel); GR (1/3 of the way from the pubic tubercle to the medial edge of the knee joint); TA (approximately 1/3 of the distance between the head of the fibula to the medial malleolus). Two ground electrodes were placed on the anterior surface of the patella as well as the lateral malleolus.

The sEMG signals were amplified using two 8-channel Bortec AMT-8 systems (gain = 1000-5000 Hz, input impedance = 10 GWs, 10-1000 Hz, CMRR 115 db at 60 Hz, Bortec Biomedical, Calgary, AB), analog to digitally converted (A/D) using a 16-bit A/D card (National Instruments, Austin, TX) at a sampling rate of 2048 Hz.

Ground reaction forces during the weight acceptance and step-out phase of the cutting task for the participant's dominant leg were collected using an AMTI-OR6 (Advanced Medical Technologies Inc., Watertown, MA, USA) force plate with dimensions of 46.35 × 50.8 cm. Along with sEMG, Ground reaction forces were sampled at a rate of 2048 Hz, amplified and converted using the same 16-bit A/D card as with the sEMG (National Instruments, Austin, Texas).

Whole body kinematics were collected using a 14-camera motion capture passive marker system (Raptor 4, Motion Analysis, Santa Rosa, CA), sampled at a rate of 120 Hz, captured using Cortex Software (version 5.5, Motion Analysis, Santa Rosa, CA). The marker set consisted of 45 reflective markers (Appendix D). Markers were placed bilaterally on each foot, lower leg, thigh, shoulder, arm and forearm, as well as on the pelvis, lower trunk, upper trunk and head (Figure 4). Any missing kinematic data identified during post-processing were fitted using a cubic spline interpolation, and trials that resulted in gaps that exceeded 200 ms were removed from further analysis (Howarth & Callaghan, 2010).

Marker	Location
1	Apex of Skull
2	Front Forehead
3	Occipital Protuberance
4	Spinous Process of 7th Cervical Vertebrae
5	Manubrium
6	Right Lower Body of Scapula
7	Right Acromion Process
8	Left Acromion Process
9	Right Triceps Body
10	Right Lateral Epicondyle of Humerus
11	Right Medial Epicondyle of Humerus
12	Right Lateral Surface of Mid-Forearm
13	Right Radial Styloid Process
14	Right Ulnar Styloid Process
15	Left Triceps Body
16	Left Lateral Epicondyle of Humerus
17	Left Medial Epicondyle of Humerus
18	Left Lateral Surface of Mid-Forearm
19	Left Radial Styloid Process
20	Left Ulnar Styloid Process
21	Lower Back L4/L5 Vertebrae
22	Right ASIS
23	Left ASIS
24	Right PSIS
25	Left PSIS
26	Right Greater Trochanter
27	Left Greater Trochanter
28	Right Anterior Mid-Femur
29	Right Lateral Epicondyle of Femur
30	Right Medial Epicondyle of Femur
31	Right Lateral Mid-Shank
32	Right Anterior Distal Shank
33	Right Lateral Malleolus
34	Right Medial Malleolus
35	Right Calcaneus
36	Right Superior Big Toe
37	Left Anterior Mid-Femur
38	Left Lateral Epicondyle of Femur
39	Left Medial Epicondyle of Femur
40	Left Lateral Mid-Shank
41	Left Anterior Distal Shank
42	Left Lateral Malleolus
43	Left Medial Malleolus
44	Left Calcaneus
45	Left Superior Big Toe

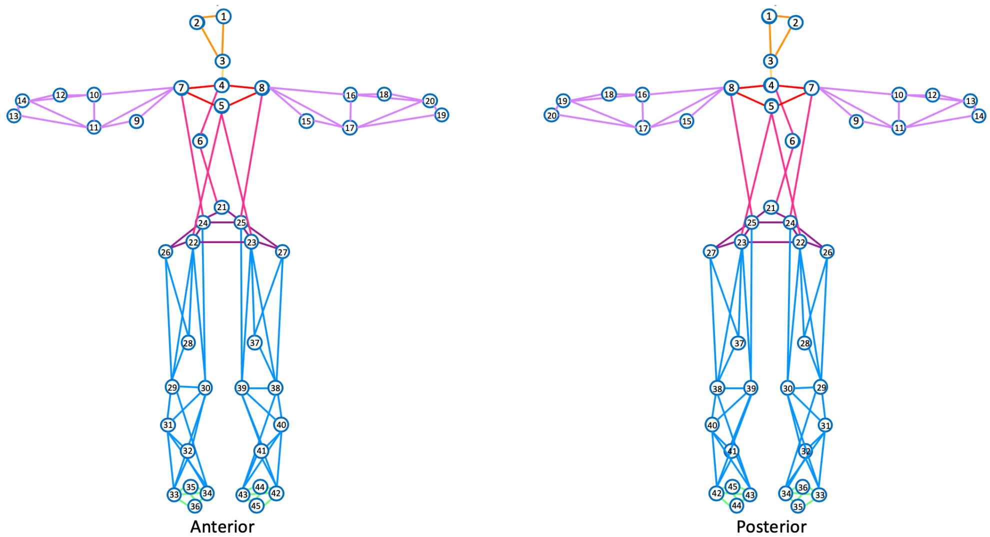

Figure 4: List of the marker set used during motion capture and diagram of marker placements.

All sEMG, force and motion data were collected in synchronization on the same computer using Cortex Motion Analysis Software (Cortex version 5.5, Motion Analysis, Santa Rosa, C A, USA) (Figure 5).

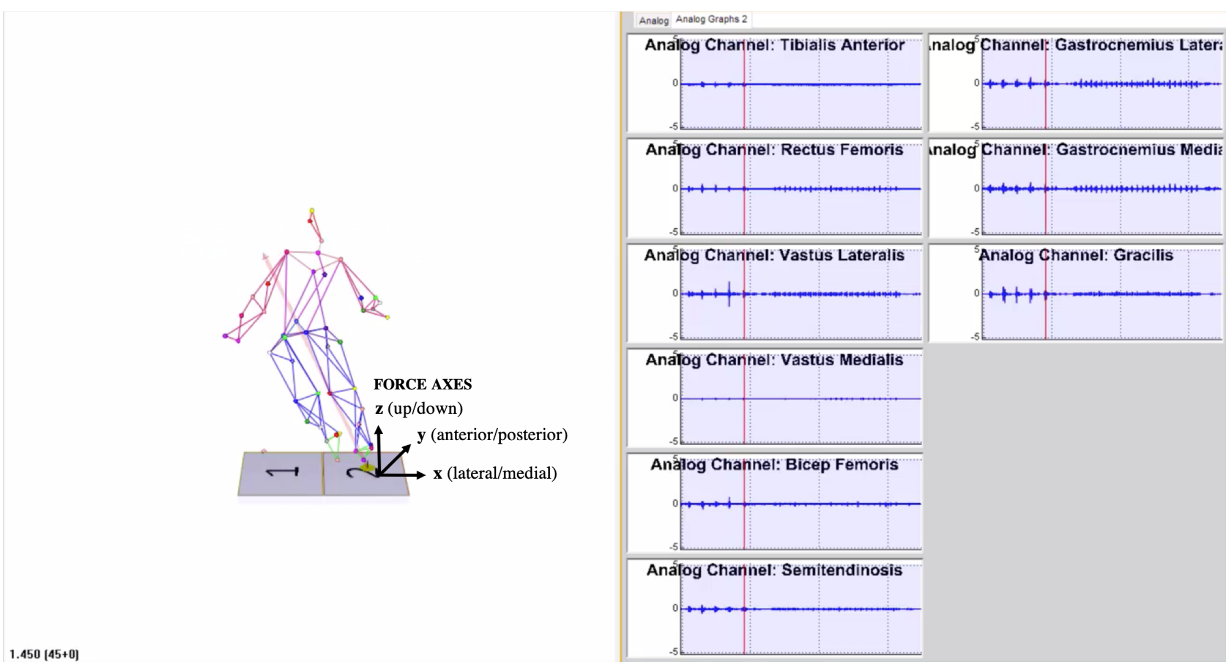

Figure 5: Cortex Motion Analysis display showing synchronized collection between instruments. Posterior view of motion capture MarkerSet and force plate setup (left), and sEMG analog graphs (right).

3.3 EXPERIMENTAL PROCEDURES AND PROTOCOL

Participants were required to attend two different sessions: (1) an orientation session and (2) a testing session, separated by at least 72 hours.

ORIENTATION SESSION

Participants began by signing a letter of consent (Appendix B) to participate in this study, then filling out the Lower Limb Questionnaire (Appendix E) and the Get Active Questionnaire (Appendix F). Participant height and lower limb length were measured using a measuring tape, and weight (N) was measured using a force plate. Participants were then taken through a general warm-up on a stationary bicycle, followed by dynamic stretching. Next, participants were seated on the Biodex (System 4 Pro, Biodex Medical System, Shirley, NY), adjusted so that the hip, knee and ankle angles were at 90˚. The maximum strength of the hamstrings and quadriceps muscle groups were then tested to determine the H:Q ratio. The H:Q protocol consists of the participant performing 3 maximum voluntary exertions (MVE) of the quadriceps, resting for 15 seconds between bouts, then performing 3 MVEs of the hamstrings, resting for 15 seconds between bouts, then 2 bouts of subsequent quadricep and hamstring MVEs, separated by 15 seconds. MVE's were performed at a velocity of 60˚/s, between the angle range of 40˚-100˚ of knee flexion. The peak torque of the hamstrings and quadriceps during these maximal concentric contractions were used to evaluate the H:Q of each individual. Participants whose H:Q > 0.45 were able to continue with the study. The remainder of the orientation session was used to practice the fatiguing slide board task and the cutting techniques, as well as familiarize the participant with the Rating of Fatigue scale to be used in the testing session (Appendix G-H).

TESTING SESSION

Participants arrived at the lab and filled out the Returning Participant Questionnaire (Appendix I) and completed the same warmup as during the orientation session. Once complete, five maximal cuts were performed, led by the dominant leg. Participants started back at a comfortable distance from the force plate, marked on the floor, and cut at a 45˚ angle from the centre of the force plate. The maximal cut distance was observed, and 80% of that distance was marked. Next, nine pairs of sEMG electrodes (Appendix C) were affixed to the skin, and participants laid still and quiet for a 30 second noise trial collection.

Maximum voluntary exertions (MVEs) for each muscle group being recorded were then collected. This involved the participant holding a 3-second contraction against resistance, three separate times with 30 seconds of rest between efforts. From a seated position on the Biodex, with the trunk, knee and hip angles at 90°, participants were instructed to maximally extend, for the quadriceps MVEs, and flex, for the hamstrings MVEs, the knee against resistance (Figure 6).

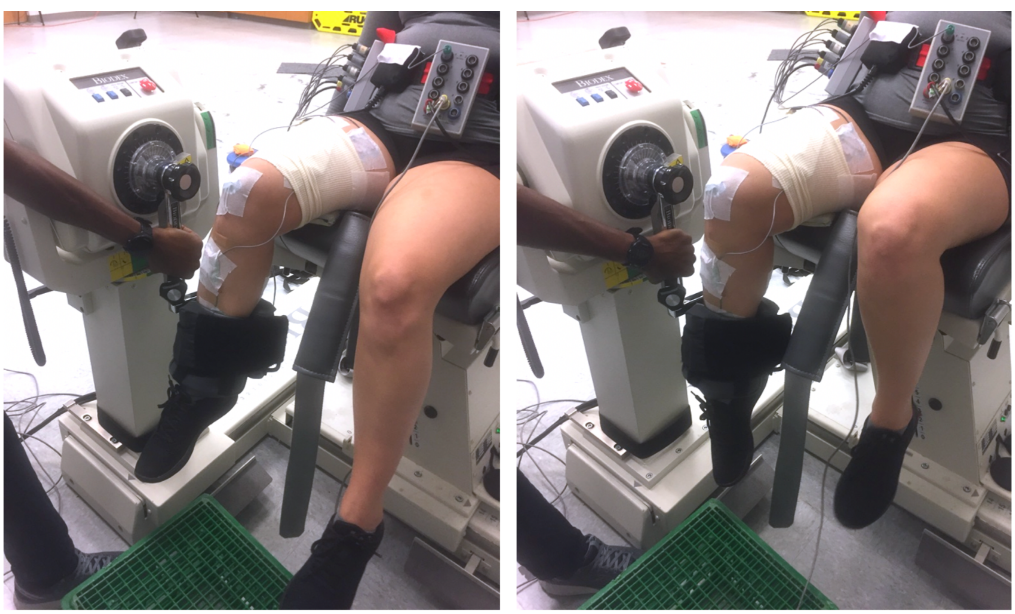

Figure 6: Participant performing MVEs of the quadriceps (left) and hamstring (right) muscle groups while seated on the Biodex.

Participants were then instructed to step into a hand-crafted calf press, and maximally plantar flex their ankle against resistance, for the gastrocnemius medial and lateral MVEs. Next, they were instructed to maximally dorsiflex against resistance for the MVE of TA. Lastly, from a standing position, participants performed an isometric adduction effort with their leg against resistance for the MVE of GR (Figure 7).

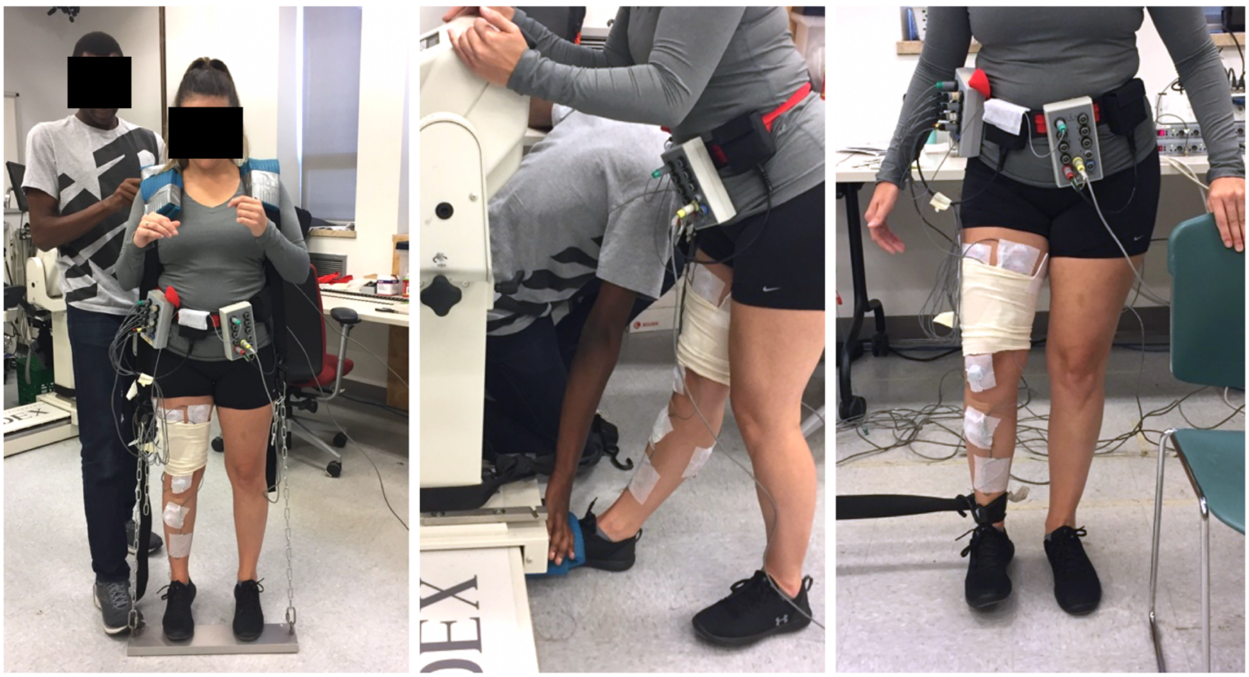

Figure 7: Participant performing MVEs of the gastrocnemius (left) tibialis anterior (middle) and gracilis (right) muscles while assisted by researcher.

Following MVEs, the 45 retroreflective markers (Figure 8) to be used for motion capture purposes were placed on the participants (Appendix D), and a heart rate monitor was strapped around their chest. Range of motion trials and calibration of the Cortex system to the MarkerSet in use were then carried out. During this time participants were refamiliarized with the Rating of Fatigue scale to be used during the trial (Appendix G-H). Once set up of the instrumentation was complete, the trial commenced.

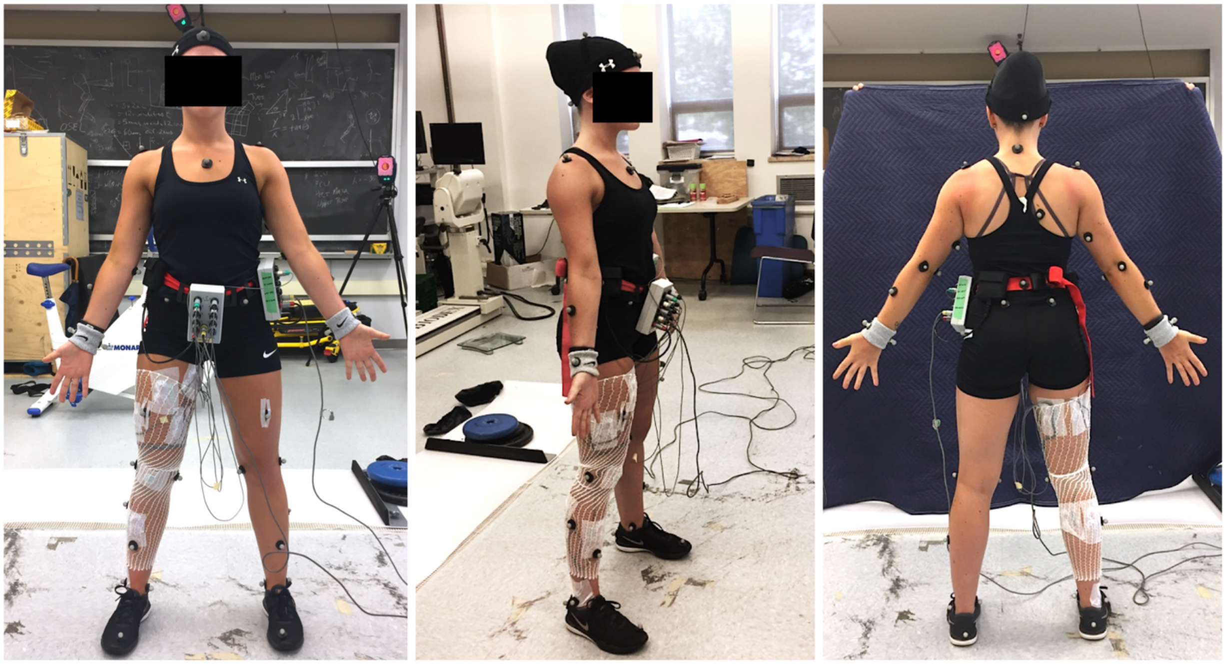

Figure 8: Participant setup with 45 retro-reflective markers for motion capture, and 9 pairs of sEMG electrodes.

Participants began the trial by performing five maximal cuts on the force plate as a baseline measure. At this time, participant's heart rate and Rating of Fatigue were recorded. After this, participants began the fatiguing protocol, which consisted of sliding back and forth, on alternating legs, on an adjustable slide board (Blue Sports Import-Export Inc., Nicolet, QC, Canada). 'Speed booties' were worn in order to reduce friction during sliding (Figure 9).

Figure 9: "Speed Booties" to be worn during sliding in order to reduce friction.

Participants began, on cue, by pushing off of the end plate of the slide board with the dominant leg (Figure 10). A metronome then cued the beginning of each subsequent stride. Sliding lasted for one minute, during which participants completed 45 slides. Once complete, the participant was assisted in removing the speed booties quickly. The participant's Rating of Fatigue and heart rate were recorded at this point. They then immediately stepped to the force plate area and performed five maximal cuts on the force plate.

Figure 10: Push-off (left), mid-stance (middle) and end (right) phases of one stride using a slideboard.

A successful cut was one in which the participant planted their dominant foot within the area of the force plate, and their new direction was in line with a line drawn on the floor at a 45° to the point of contact. Participants then rapidly returned to the slide-board, and were assisted with putting on the speed booties, to continue the fatiguing protocol. This process continued until the trial was terminated. Trial termination could result from any of the following criteria: (1) the inability to reach 80% of the participant's maximal cut distance, as recorded at the beginning of the session; (2) a Rating-of-Fatigue of 9 or higher is reached, or a reading of 8 three consecutive times; (3) noticeable change in body mechanics putting the participant at risk; or (4) volitional exhaustion by the participant.

3.4 DATA ANALYSIS

Collected digital sEMG data were high pass filtered with a cut-off of 140 Hz, using a second order Butterworth filter, then full-wave rectified, and low-pass filtered using a second order Butterworth filter at 2.5 Hz (Potvin & Brown, 2004). Force data were low-pass filtered using a second order Butterworth filter with a cut-off of 50 Hz. All sEMG and force data were down-sampled from 2048 Hz to 120 Hz to match the sample rate of the collected motion data, thereby syncing the collected data in the same time domain. All sEMG data recorded during experimental trials was normalized to the highest value recorded during the MVE trials for each individual muscle.

For the purposes of this study, the cutting maneuver was separated into three phases: (1) step-in, (2) weight acceptance, and (3) step-out (Figure 11). The step-in phase was defined as the interval of flight prior to the cut to initial contact of the dominant foot. The weight acceptance phase is defined as the time interval between initial contact and peak knee flexion of the dominant leg. The step-out phase is defined as the time interval from the point of peak knee flexion of the dominant leg to the moment the foot leaves the ground. Average sEMG activation for each muscle was calculated during the weight acceptance phase of the side-cut and examined throughout the duration of the trial to monitor fatigue related changes.

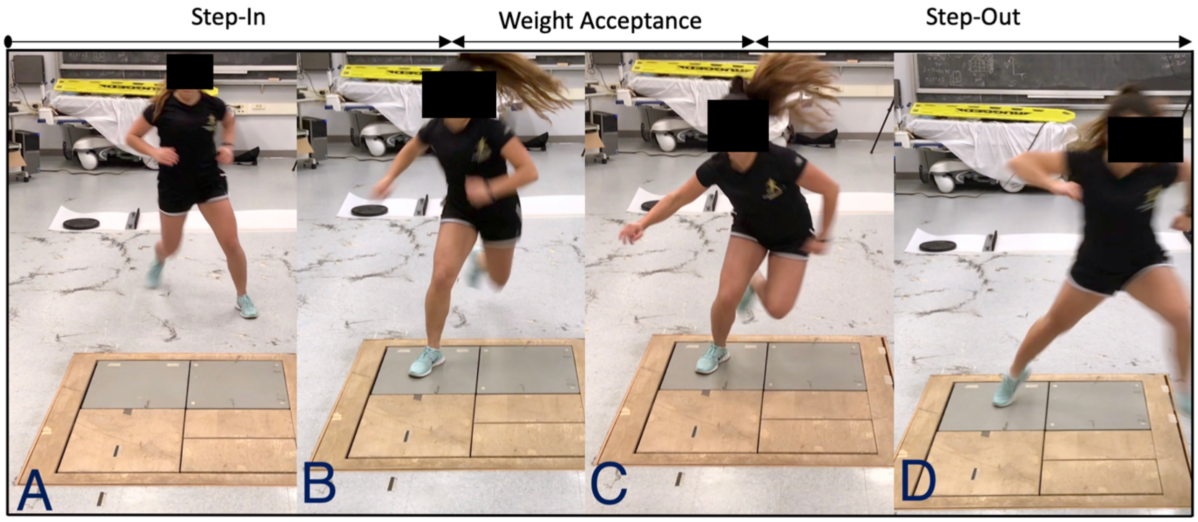

Figure 11: Phases of cutting maneuver: Step-In (a to b), Weight Acceptance (b to c) and Step-Out (c to d).

Motions of the retro-reflective markers were used to calculate three-dimensional joint angles. These data were imported into the Software for Interactive Musculoskeletal Modeling (SIMM, MusculoGraphics, Inc., Santa Rosa, CA, USA) (Figure 12), and examined during the weight acceptance phase, as this is when non-contact ACL injuries are suggested to occur (Jamison, McNally, Schmitt, & Chaudhari, 2013). The calculated joint angles included: hip flexion (negative values are extension), adduction (negative values are abduction) and internal rotation (negative values are external rotation); knee flexion (negative values are extension); ankle dorsiflexion (negative values are plantar flexion).

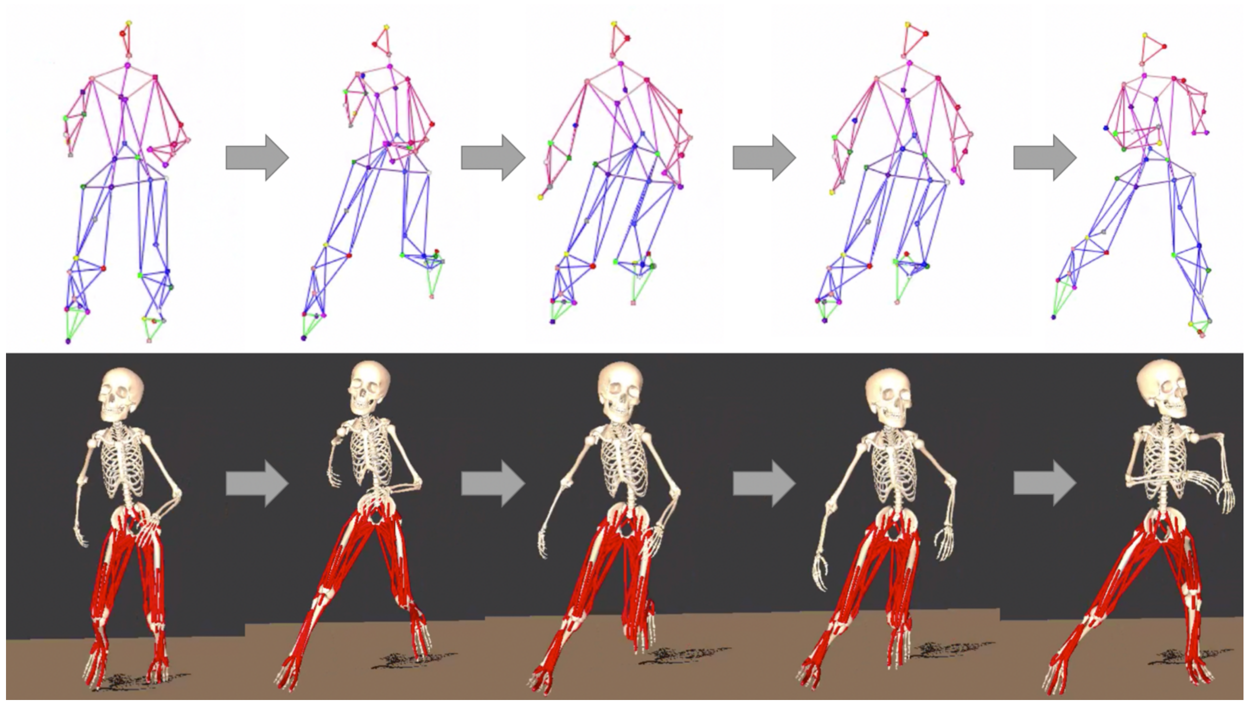

Figure 12: Model skeleton of a participant completing a cut as captured through Cortex Motion Analysis (top) and imported to SIMM (bottom) in order to calculate joint angles.

The weight acceptance phase of the cutting maneuver was the period of interest, from which all values of dependent variables were recorded (Figure 13). For each effort, this time period began with the point of initial contact on the force plate and ended at the point of maximal knee flexion, as indicated by motion capture data. Because all data collections were synchronized, this same chunk could be distinguished from each dataset. Calculations for the mean, effective duration, force at peak knee flexion and impulse of each variable were done for each dependent variable for each effort during this time.

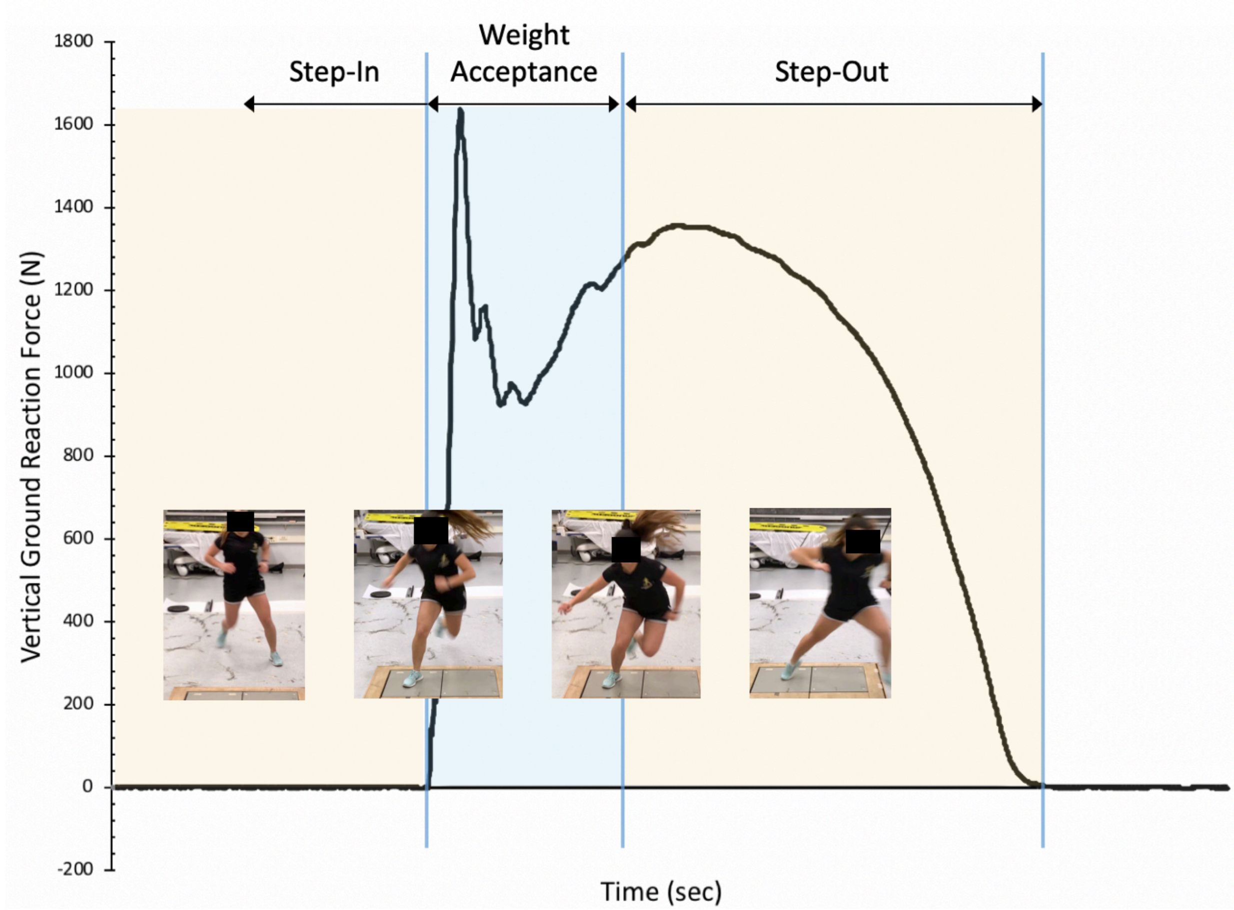

Figure 13: Example of vertical ground reaction force (vGRF) curve from an athletic cut performed on the force plate, outlining the timepoints at which the phases of cutting occur. The Weight Acceptance phase, indicated in blue, samples the time period in which the dependent variables were evaluated throughout the phase of the cut.

A method referred to as 'rubber-banding', adapted from Winter (2005), was used in order to time-normalize the data based on the number of cutting efforts completed by each participant (Figure 14). Because of individual differences in fatigue progression and trial length, this allowed for averaging and comparison between each participant to be made. A particular point in the trial may be referred to in relation to its total relative duration, and thus fatigue progression, (%Trial) as opposed to the absolute number of cutting efforts. Using this method, each participant's total sample period was fit to a polynomial curve ranging from the beginning of trial (%Trial=0), defined as the participant's first cutting effort on the force plate, to the termination of the trial (%Trial=100).

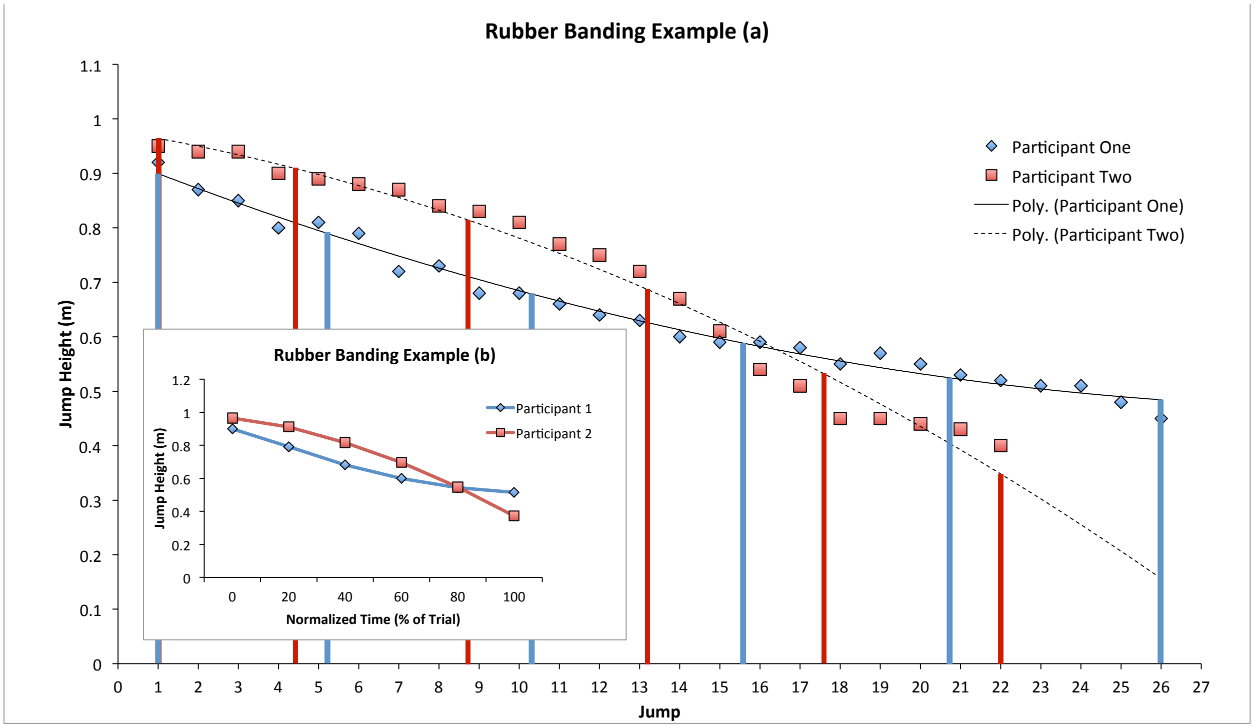

Figure 14: (a) A schematic of the rubber-banding technique used to normalize trial time across participants for peak knee flexion. The blue diamond's represent participant 1, whereas the red squares represent participant 2, as each participant progressed through the maximal fatigue trial. The lines (solid for participant 1 and dashed for participant 2) 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 (0%), completed slides = 12 (20%), 24 (40%), 36 (60%), 48 (80%) and 60 (100%), and compared with participant 2 (red bars), intervals from completed slide 1 (0%), 10 (20%), 20 (40%), 30 (60%), 40 (80%) and 50 (100%). (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 (a). This allows for averages to be calculated over time, across participants performing a different number of jumps

Values for the angle of the resultant force were calculated using the average force at peak knee flexion from each linear force component (X, Y, Z). The arctan of the resultant of the X and Y force components and Z force component were used to get the average angle of the resultant force for each tenth of trial completion.

This study used the variable ‘effective duration’ in order to evaluate the strategies utilized by participants to dissipate downward momentum during the weight acceptance phase. The effective duration is calculated as the impulse normalized to the peak force during landing and, essentially indicates how long the peak would need to occur for the same impulse as that measured (Figure 15).

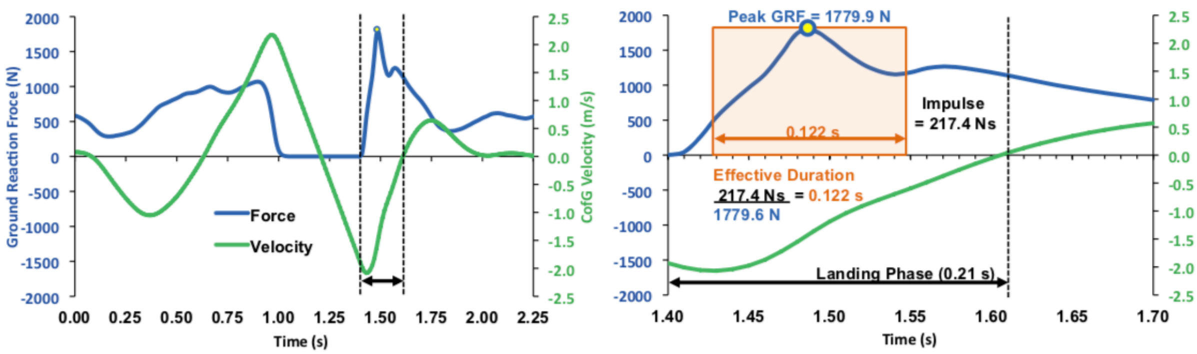

Figure 15: (left) full time-history of vertical ground reaction force and center of gravity (CofG) velocity. (right) expanded view of the landing phase from first contact to the bottom of the crouch (velocity = 0). The full duration is 0.21 s, but ED = 0.122 s, such that the area in the orange box is equal to the area under the GRF curve during the landing phase (i.e. 217.4 Ns). The participant in this example had a body weight of 582 N.

3.5 STATISTICAL ANALYSIS

Eighteen separate one-way repeated measures analysis of variance (ANOVA) were used, with %Trial (11 levels), as the independent variable. The dependent variables used in these analyses included:

- Rating of fatigue: mean rating
- Heart Rate: mean beats/minute
- hip: flexion, internal rotation, and adduction angles at peak knee flexion
- knee: flexion angle at peak knee flexion
- ankle: flexion angles at peak knee flexion
- resultant force: angle with the force plate surface, impulse, force at peak knee flexion, and effective duration
- vertical ground reaction force (Z): impulse, force at peak knee flexion, and effective duration
- medial/lateral ground reaction force (X): impulse, and force at peak knee flexion
- anterior/posterior (Y): impulse, and force at peak knee flexion

Additionally, a two-way mixed repeated measures ANOVA was performed with %Trial (11 levels) and Muscle ($n = 9$) as the independent variables, and average normalized EMG amplitude as the dependent variable (%EMG).

The significance level for all main and interaction effects was set at $p<0.05$. Effect sizes were determined using eta squared (η^2) analyses. For effects meeting significance ($p<.05$), a Tukey's HSD test were run. All data used in statistical analysis can be found in Appendix J.

CHAPTER 4

RESULTS

The results have been separated into four sections: rating of fatigue and heart rate, kinetic, kinematic, and muscle activation data. A summary of each ANOVA can be found in Table 1.

Table 1: Summary of statistical analysis from Repeated Measures ANOVA.

Dependent Variable	Effect/ Interaction	Measure	F	MSE	Effect Size (η^2)	Sig. (*p*)
Rating of Fatigue	%Trial	Mean	225.92	1.386	**0.9576**	**0.0001*** **
Heart Rate	%Trial	Mean	125.889	484.255	**0.9264**	**0.0001*** **
Resultant Force	%Trial	Angle	3.001	2.346	0.2308	0.073
	%Trial	Effective Duration	0.81	0.001	0.08	0.447
	%Trial	Impulse	2.215	729.186	0.1813	0.147
	%Trial	Peak	0.847	6597.07	0.0781	0.585
Vertical Ground Reaction Force	%Trial	Effective Duration	0.593	0.001	0.04	0.541
	%Trial	Impulse	2.042	710.861	0.1696	0.167
	%Trial	Peak	0.708	6626.212	0.0661	0.487
Anterio-posterior Shear Force	%Trial	Impulse	1.553	29.663	0.1344	0.239
	%Trial	Peak	4.778	891.209	**0.3233**	**0.026***
Medial-Lateral Shear Force	%Trial	Impulse	4.416	57.693	0.3063	0.035*
	%Trial	Peak	0.023	216.435	0.0023	0.924
Hip Flexion Angle	%Trial	Peak	4.607	47.463	**0.3154**	**0.033***
Hip Adduction Angle	%Trial	Peak	2.463	40.537	0.1976	0.116
Hip Internal Rotation Angle	%Trial	Peak	3.17	104.457	0.2407	0.076
Knee Flexion Angle	%Trial	Peak	6.194	9.128	0.3825	0.012*
Ankle Flexion Angle	%Trial	Peak	0.446	134.39	0.0427	0.535
sEMG Amplitude	%Trial	Mean	4.240	342.971	0.1798	**0.048***
	%Trial*Muscle	Mean	0.796	159.019	0.0413	0.526
	Muscle	Mean	1.640	1405.021	0.779	0.187

** significance met at the p < 0.05 level*
*** significance met at the p < 0.01 level*
**** significance met at the 0.0001 level*

4.1 RATING OF FATIGUE & HEART RATE

There was a significant main effect of %Trial, F (2.619, 26.193) = 225.92, p = 0.0001, η^2=0.9576 on mean Rating of Fatigue. There was also a significant main effect of %Trial, F (1.707, 17.074) = 125.889, p = 0.0001, η^2=0.9264 on mean Heart Rate (Figure 16).

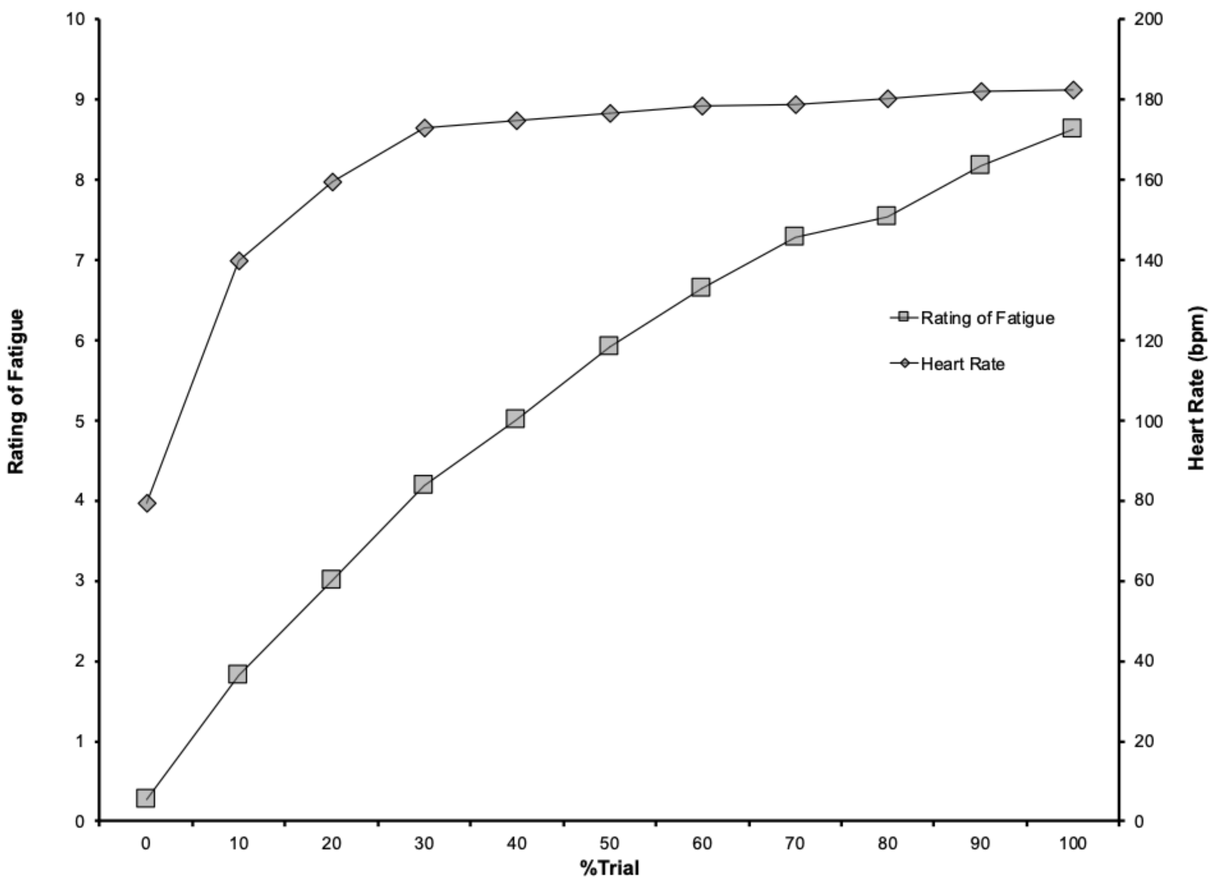

Figure 16: Average Rating of Fatigue and Heart Rate progression throughout the trial (n = 12).

4.2 KINETIC DATA

4.2.1 ANTERIO-POSTERIOR SHEAR FORCE

There was no significant main effect of %Trial on the anterio-posterior shear force impulse during weight acceptance (p = 0.239). There was a significant main effect of %Trial, F (1.757, 17.568) = 4.778, p = 0.026, η^2=0.323, on the anterio-posterior shear force at peak knee flexion during weight acceptance (Figure 17). Post-hoc comparisons revealed a steady increase in anterio-posterior shear force, becoming significant at 90% of the trial (Table 2).

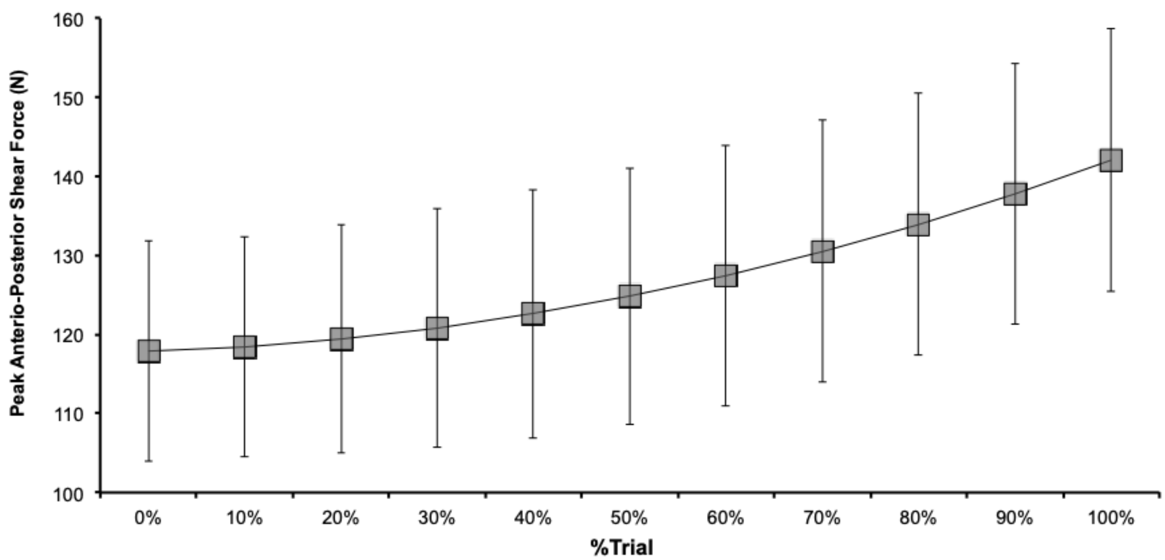

Figure 17: The effect of time on the anterio-posterior shear force at peak knee flexion during the weight acceptance phase of cutting. Standard error bars are shown (n=12).

Table 2: Summary of post-hoc comparisons from Repeated Measures ANOVA for anterio-posterior shear force at peak knee flexion

%Trial	0	10	20	30	40	50	60	70	80	90	100
0										0.041*	0.013*
10									0.041*	0.013*	0.005**
20								0.041*	0.013*	0.005**	0.005**
30							0.041*	0.013*	0.005**	0.005**	0.009**
40						0.041*	0.013*	0.005**	0.005**	0.009**	0.017*
50							0.005**	0.005**	0.009**	0.017*	0.031*
60								0.009**	0.017*	0.031*	0.048*
70									0.031*	0.048*	
80											
90											
100											

** significance met at the p < 0.05 level*

*** significance met at the p < 0.01 level*

4.2.2 MEDIAL-LATERAL SHEAR FORCE

There was a significant main effect of %Trial, $F\ (1.634, 16.343) = 4.416$, $p = 0.035$, $\eta^2=0.306$, on the impulse of the medial-lateral shear force during weight acceptance (Figure 18). Post-hoc comparisons between trial periods are shown in Table 3. There was no significant main effect of %Trial on the medial-lateral shear force at peak knee flexion during weight acceptance (p = 0.924). Post-hoc comparisons revealed a steady decrease in the impulse of medial-lateral shear, becoming significant at 60% of trial and plateauing (Table 3).

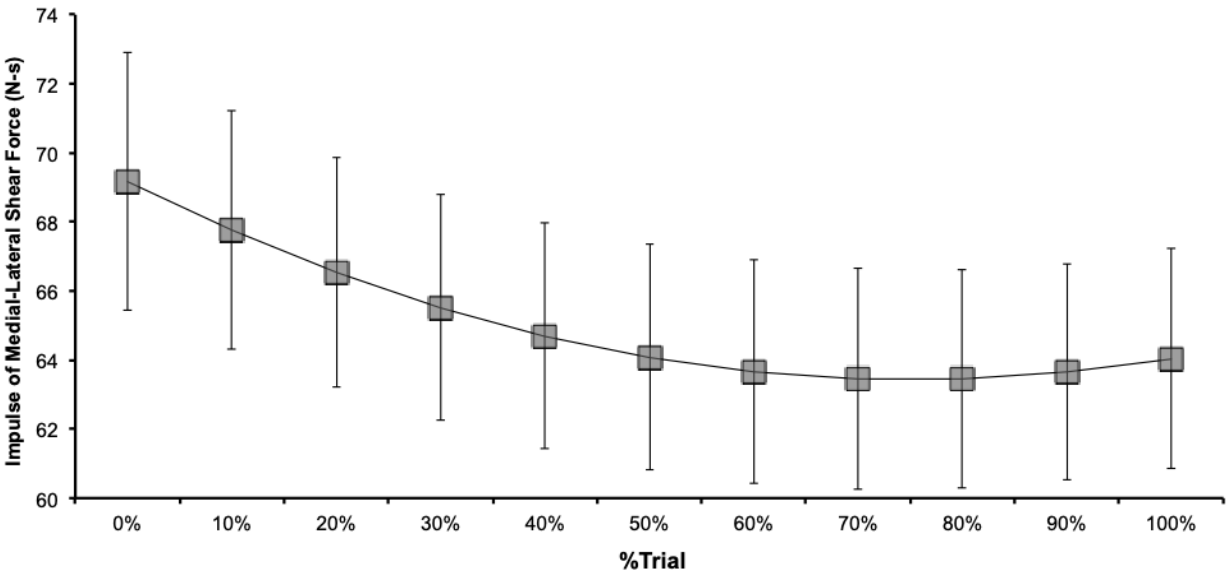

Figure 18: The effect of time on the impulse of medial-lateral shear force during the weight acceptance phase of cutting. Standard error bars are shown (n=12)

Table 3: Summary of post-hoc comparisons from Repeated Measures ANOVA for impulse of medial-lateral shear force

%Trial	0	10	20	30	40	50	60	70	80	90	100
0							0.049*	0.042*	0.035*	0.030*	0.030*
10						0.049*	0.042*	0.035*	0.030*	0.030*	0.047*
20					0.049*	0.042*	0.035*	0.030*	0.030*	0.047*	
30					0.042*	0.035*	0.030*	0.030*	0.047*		
40						0.030*	0.030*	0.047*			
50							0.047*				
60											
70											
80											
90											
100											

** significance met at the p < 0.05 level*

4.2.3 VERTICAL GROUND REACTION FORCE

There was no significant main effect of %Trial on the ground reaction force effective duration ($p = 0.541$), impulse ($p = 0.167$), or at peak knee flexion ($p = 0.487$).

4.2.4 RESULTANT FORCE

There was no significant main effect of %Trial on the resultant ground reaction force angle ($p = 0.073$), effective duration ($p = 0.447$), impulse ($p = 0.147$) or at peak knee flexion ($p = 0.585$).

4.3 KINEMATIC DATA

4.3.1 HIP ADDUCTION ANGLE

There was no significant main effect of %Trial on hip adduction angle at peak knee flexion ($p = 0.116$).

4.3.2 HIP INTERNAL ROTATION ANGLE

There was no significant main effect of %Trial on hip internal rotation angle at peak knee flexion ($p = 0.076$).

4.3.3 ANKLE FLEXION ANGLE

There was no significant main effect of %Trial on ankle flexion angle at peak knee flexion ($p = 0.535$).

4.3.4 HIP FLEXION ANGLE

There was a significant main effect of %Trial, F (1.6, 16.0) = 4.607, p= 0.033, η^2=0.315, on hip flexion angle at peak knee flexion during the weight acceptance phase (Figure 19). Post-hoc comparisons showed a steady decrease in hip flexion at peak knee flexion, becoming significant at 90% of the trial (Table 4).

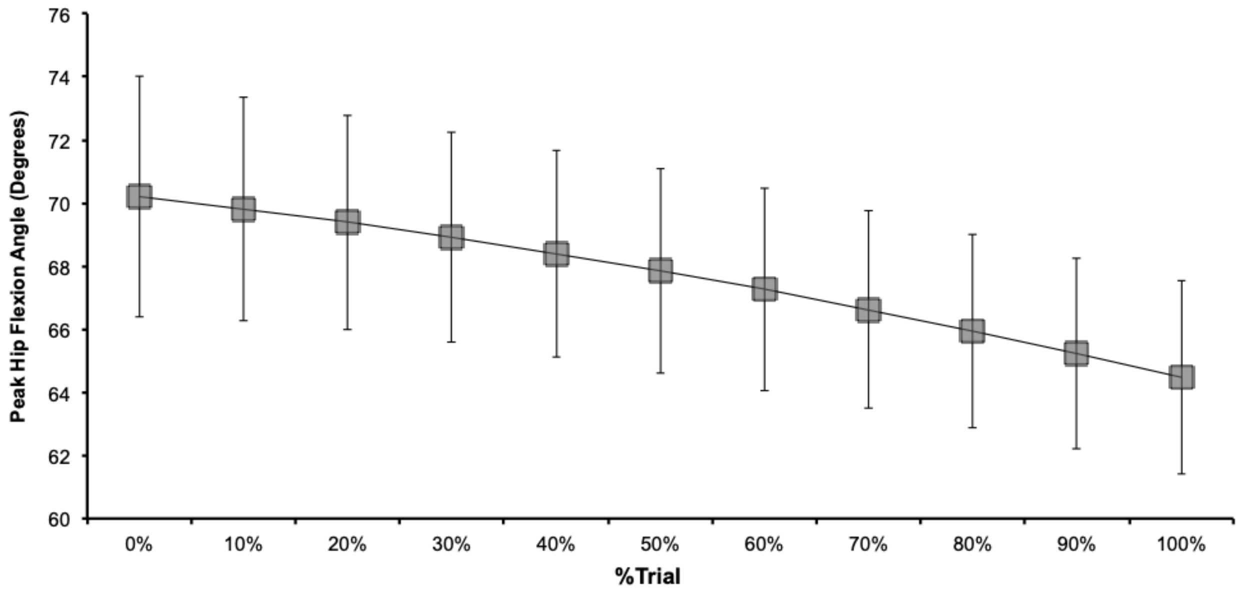

Figure 19: The effect of time on the hip flexion angle at peak knee flexion during the weight acceptance phase of cutting. Standard error bars are shown (n=12)

Table 4: Summary of post-hoc comparisons from Repeated Measures ANOVA for hip flexion angle at peak knee flexion.

%Trial	0	10	20	30	40	50	60	70	80	90	100
0										0.038*	0.016*
10									0.038*	0.016*	0.008**
20								0.038*	0.016*	0.008**	0.008**
30							0.014*	0.008**	0.008**	0.016*	0.038*
40						0.038*	0.016*	0.008**	0.008**	0.014*	0.029*
50							0.008**	0.008**	0.014*	0.029*	
60								0.014*	0.029*		
70											
80											
90											
100											

** significance met at the p < 0.05 level*

*** significance met at the p < 0.01 level*

4.3.5 KNEE FLEXION ANGLE

There was a significant main effect of %Trial, F (1.713, 17.126) = 6.194, p = 0.012, η^2=0.382, on peak knee flexion angle (Figure 20). Post-hoc comparisons showed a steady decrease in peak knee flexion angle, becoming significant at 60% of trial and then plateauing (Table 5).

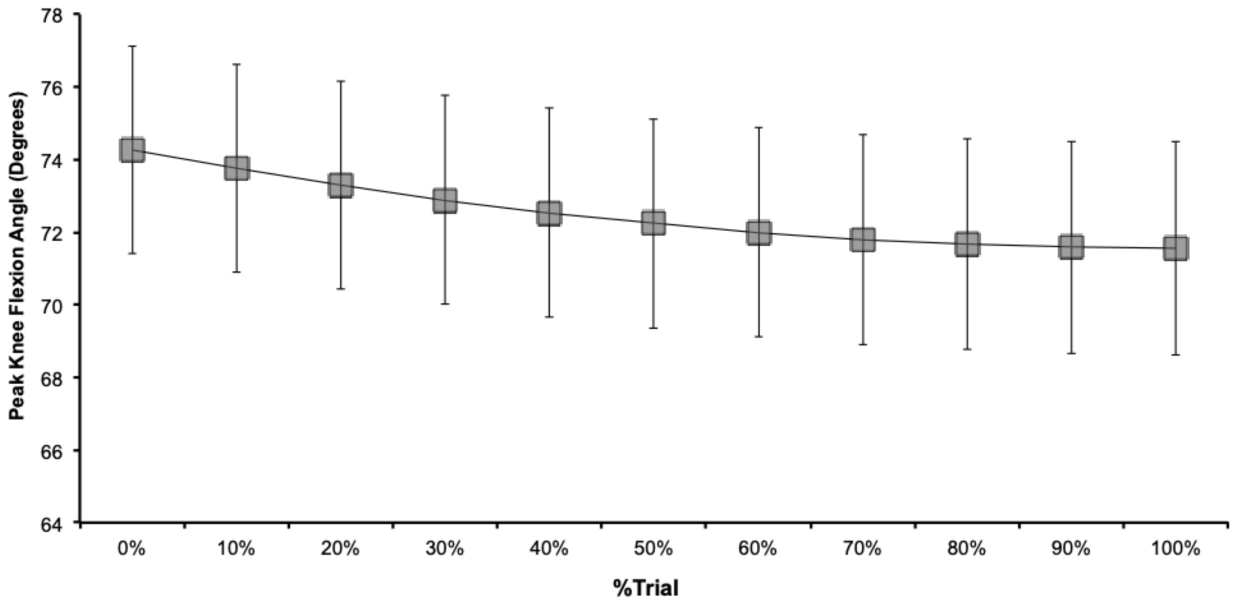

Figure 20: The effect of time on the peak knee flexion angle during the weight acceptance phase of cutting. Standard error bars are shown (n=12).

Table 5: Summary of post-hoc comparisons from Repeated Measures ANOVA for peak knee flexion angle.

%Trial	0	10	20	30	40	50	60	70	80	90	100
0							0.043*	0.025*	0.012*	0.005**	0.002**
10						0.043*	0.025*	0.012*	0.005**	0.002**	0.004**
20					0.043*	0.025*	0.012*	0.005*	0.002*	0.004*	0.020*
30					0.025*	0.012*	0.005**	0.002**	0.004**	0.020*	
40						0.005**	0.002**	0.004**	0.020*		
50							0.004**	0.020*			
60											
70											
80											
90											
100											

** significance met at the p < 0.05 level*

*** significance met at the p < 0.01 level*

4.4 MUSCLE ACTIVATION DATA

The interaction between Muscle and %Trial was not significant ($p = 0.526$), nor was the main effect of muscle on EMG amplitude ($p = 0.187$). There was a significant main effect of %Trial, F (1.388, 13.878) = 4.240, $p = 0.048$, $\eta^2 = 0.1798$, on average EMG amplitude (Figure 21). Post-hoc comparisons show a steady decrease from the start, becoming significant by 10% of trial, then an increase after 60%, becoming significant by trial end (100%) (Table 6).

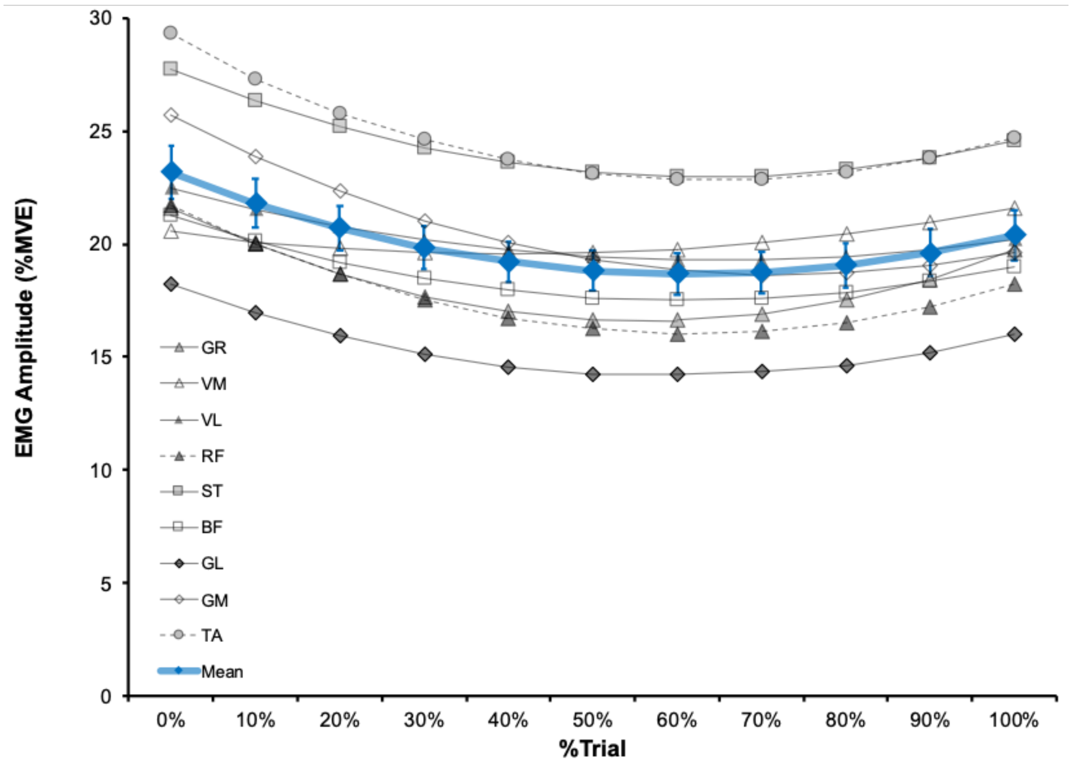

Figure 21: The effect of time on sEMG amplitude during the weight acceptance phase of cutting. Means are shown for each time point for each channel of sEMG. Standard error bars are shown the mean are shown (n=12).

Table 6: Summary of post-hoc comparisons from Repeated Measures ANOVA for mean sEMG amplitude.

%Trial	0	10	20	30	40	50	60	70	80	90	100
0		0.002**	0.002**	0.003**	0.004**	0.005**	0.007**	0.013*	0.027*		
10			0.003**	0.004**	0.006**	0.008**	0.015*	0.030*			
20				0.005**	0.008**	0.014*	0.030*				
30					0.015*	0.030*					
40											
50											
60											
70											
80											
90											0.039*
100											

** significance met at the $p < 0.05$ level*

*** significance met at the $p < 0.01$ level*

CHAPTER 5

DISCUSSION

The most important finding of this study was that, across all monitored muscles, there was a significant main effect of time on muscle activity such that the pooled average of the normalized sEMG activity demonstrated an overall decrease of 11.9% from the start to the end of the trials. However, it should be noted that, by 60% trial completion, sEMG activity began to increase and this may reflect a practice effect or a potentially harmful response to fatigue. There were significant decreases in the flexion angles at the hip and knee at peak by 8.1% and 3.7%, respectively. Lastly, anterio-posterior shear forces at peak knee flexion increased from the beginning of trial by approximately 20%, and the impulse of medial-lateral shear force decreased by about 7% during the weight acceptance phase of the cut, though there were no significant changes seen in vGRF or resultant force of this task.

The EMG results were unexpected and not consistent with many other studies showing an increase in sEMG amplitude with muscle fatigue (Rodacki, Fowler, & Bennett, 2002; Kallenberg, Schulte, Disselhorst-Klug, & Hermens, 2007; Cirfrek, Medved, Tonkovic, & Ostojić, 2009). Upon inspecting the pattern of muscle activity in more detail, it became apparent that sEMG amplitudes did decrease to a point, but then started to increase until the trial was terminated. For most muscles, this turning point was at 60% of trial completion – the exceptions were VM at 40% of trial completion and GM at 70% of trial completion. Upon further examinations of pairwise comparisons for all muscles, there were significant increases in average sEMG amplitudes from 60%-100% of trial completion.

It is possible that this resulted from a learning effect occurring as participants cycled repeatedly through the trial. Although participants were required to attend an orientation session, in which they were able to practice the exact methods to be executed during the testing session, the cutting maneuver - especially performed in isolation - is not a common movement in everyday life. It is possible that, as individuals became more familiar with and acclimated to the task, their pattern of muscle activity were becoming more optimized. This may suggest that participants were refining their recruitment strategies, limiting the amount of antagonistic muscle activity needed to perform the task, and coordinating muscle contractions in a more synergistic manner (Hobart & Vorro, 1974; Bernardi, Solomonow, Nguyen, Smith, & Baratta, 1996; Carson & Riek, 2001). This, presumably, occurred for most muscles until 60% of trial completion.

It is also possible that the decrease in sEMG activity was a reflection of a decrease in the muscle forces and their stability contributions about the knee joint. This theory can be explained with reference to the kinematic data. There was a steady decrease in both hip and knee flexion angles resulting in a straightened leg as participants continued to cycle through the trial. It is possible that, to conserve energy expenditure, participants used less muscle force when landing on the force plate and arresting their momentum before changing directions. This would mean that individuals would land with less hip and knee flexion, increasing the burden to dissipate energy on the passive structures of the knee, such as the ACL. Though this strategy may preserve energy in order to continue the activity for a longer period of time, the reduction in muscle contractions during dynamic movements, such as cutting, has the potential to increase the risk associated with injury to the ACL. As muscle activation and contractile forces decrease, there is a reduction in the capacity of the muscles to absorb energy (Mair, Seaber, Glisson, & Garrett, 1996). This decreases their contribution as dynamic stabilizers and could leave the joint’s passive stabilizers, such as the ACL, with a larger demand to dissipate energy, After the 60% point of trial, sEMG amplitudes began to increase, and it is in this portion of the data which seems as though the fatiguing state of the muscles overshadowed the initial function of decreasing EMG.

From its minimum values at 60% of the trials, to the end of the trials, overall average muscle activity increased by 9.3%. Gracilis was observed trending towards having the largest change, increasing its muscle activation by 18.8%; 5% greater than any other muscle. The Gracilis plays a role in hip adduction and knee internal rotation, which both contribute to the dynamic valgus position, compromising the safety of the ACL (Laughlin, Weinhandl, Kernozek, Cobb, & Keenan, 2011). While observing the cutting task itself it is evident that, to achieve the desired performance, slight adduction at the hip during the weight acceptance phase may be necessary to allow for more range to abduct at the hip while pushing off during the step-out phase. Therefore, the Gracilis is instrumental in controlling the speed and degree to which an individual adducts at the hip. Participants in this study did not differ significantly in the degree to which they rotated in adduction at the hip, which may suggest that the increase in EMG was a reflection of an increase in muscle force required to maintain stability at the joint, or increase activation necessary to produce the same force due to fatigue.

At the hip and knee, there were significant decreases in the angle of flexion at peak knee flexion during the cut. These results are in agreement with those of Borotikar and colleagues, which demonstrated increases in hip and knee extension during single leg drop landings, a different unilateral deceleration task (Borotikar, Newcomer, Koppes, & McLean, 2008). The average hip angle at peak from the beginning of the trial decreased by 5.7°; an 8.1% change in flexion. The average peak knee angle from the beginning of the trial decreased by 2.7°; a 3.7% change in peak flexion. This means that participants, during their initial cutting efforts, were in a more crouched position during the weight acceptance phase of the cut than when executing their later cutting efforts, where the position of their lower extremity was, relatively, more extended. This decreased the range of motion over which energy could be dissipated and, during this time of rapid deceleration and change in direction, created the potential for increasing strain on the structures of the knee and, of particular concern, the ACL (Kernozek & Torry, 2005).

As participants came into contact with the force plate and experienced weight acceptance during the cutting phase, the resultant of the ground reaction forces experienced did not vary significantly between their initial cutting efforts to those performed at the end of the trial. The average angle of the resultant force remained within a 1.04° range, suggesting that individuals did not alter the angle at which they landed on the force plate. The impulse, force at peak knee flexion and effective duration of the resultant force at this time also did not differ significantly. Though the change in the resultant remained fairly consistent in its direction and amplitude, by dissecting this force into its X, Y and Z components, it became apparent that the distribution of force was, in fact, affected by time. Such that the X and Y components, contributing to the shear forces, were more influential than the Z component.

The vertical ground reaction force at the weight acceptance phase did not change significantly from the beginning of the trial to termination. This finding is interesting considering that there was decrease in both hip and knee flexion. This change in landing form was expected to have an increase in the vertical ground reaction force, as with less rotation of the joints, the duration of the landing would be shorter and, thus, for the same momentum the force would have to be higher to arrest the motion over a shorter time. This would reduce the amount energy dissipated by the muscles and be represented as a higher vGRF. There was however a significant increase in the anterio-posterior shear force at peak by 20.5%. This increase is in agreement with the work of Chappell et al. (2005), who observed significant increases in anterior shear forces at peak knee flexion after participants completed a fatiguing protocol involving consecutive vertical jumping and sprints. The current findings are also similar to those of Ireland (2002), who observed more than a 30% increase in anterior tibial translation in response to fatigue during rapid run and stop maneuvers. This increase can be attributed to more anterior shear force occurring at the lower extremity as individuals landed on the force plate. As previously stated, one of the main functions of the ACL in joint stability is to resist anterior translation of the tibia with respect to the femur (Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007). Anterior shear is known to stress the ACL, and thus increase the risk of injury to the ligament. Though the measure of force through the force plate is not a direct estimate of the shear forces experienced at the knee, correlations between the two have been presented. For example, a study by D'Lima and colleagues, comparing internal and external measures of force at the knee, correlated peak tibiofemoral knee forces in vivo to peak ground reaction forces (D'Lima, Fregly, Patil, Steklov, & Colwell, Jr., 2012). There was also a significant decrease in the impulse of the medial-lateral shear force by 7.4%, however no significant difference in force at peak knee flexion. This can be interpreted as there being a shorter application time of lateral shear force to the lower extremity, and thus a higher rate of energy dissipation.

5.1 HYPOTHESES REVISITED

1) Participants would demonstrate a decrease in knee flexion angles and increase in knee valgus angles during the weight acceptance phase of the cutting task as fatigue progressed.

The current results reject the null hypothesis that there was no change in the knee flexion angles as fatigue progressed. Participants did have significant decreases in knee flexion angle during the weight acceptance phase. However, this study failed to reject the null hypothesis that there would not be an increase in knee valgus angles during the weight acceptance phase of the cutting task as fatigue progressed. Participants did not show significant increases in hip adduction and internal rotation from the beginning of the trial to the end, both of which are contributors to the dynamic valgus position.

2) Participants would experience greater vertical ground reaction forces and effective duration decreases as fatigue progressed.

The current results failed to reject the null hypothesis that vertical ground reaction forces would not increase as fatigue progressed throughout the trial. The study also failed to reject the null hypothesis that effective duration of the vertical ground reaction force would not decrease as fatigue progressed.

3) Fatigue progression would have a significant effect on normalized sEMG for RF, VL, VM, BF, ST, GL, GM, GR and TA.

The current results reject the null hypothesis that there would be no effect of fatigue progression on normalized sEMG for all muscles recorded. There was indeed a significant effect of time on the normalized sEMG for the recorded muscles. However, there was no main or interacting effect of muscle on the average sEMG amplitude. It is important to note that the main effect of time showed a significant decrease in sEMG amplitude from the beginning of the trial to the end. However further inspection into pairwise comparisons revealed the expected significant increase in sEMG after ~60% of the trial duration.

5.2 LIMITATIONS AND ASSUMPTIONS

In the current study, the main limitations and assumptions pertained to the relationship between the independent and dependent variables during the time between the beginning (%Trial = 0) and end (%Trial = 100) points. It is likely that the direct relationship was confounded due to either what has been identified as a possible learning effect, external motivation factors which were not measured. It is also possible that this relationship was misinterpreted, due to the measure of muscle force not being directly measured.

The cutting maneuver, especially performed in isolation from sport, is not a well-practiced skill. Participants were required to attend an orientation session in which they were able to practice this athletic task in the same way it would be performed during the testing session. For there to be an adequate amount of rest time between the two sessions, to minimize cumulative muscle fatigue, participants were made to wait at least of 72 hours before their testing day. This time gap was important to the study from the perspective of obtaining valid outcome measures, however, it may have hindered one of the objectives of the training session: to properly learn the test protocol. It is also possible that participants had external motivations while completing the trial, such as a better performance in terms of how long they lasted. Because of this they may have altered the way that they would normally complete the cutting task had they not expected to have to continuously repeat their performance. It is possible that this could have confounded the effects of fatigue on the outcome measures for this study, thus affecting the interpretation of the results.

Using surface EMG, we were able to determine the degree of activation of the muscles recorded, and although this is strongly correlated with muscle force, we were not able to predict the forces which resulted as a product of these activations. It could only be hypothesized what these magnitudes mean in terms of the force being produced by the muscles. Had force been directly measured, there may have been a different interpretation, and better understanding of each variable. Additionally, we did not record for every single muscle in the dominant leg, as this could lead to a more invasive and complicated collection, but there may have been muscles unaccounted for that contributed to the results.

It was assumed that the participants would have reached a near maximal level of muscle fatigue by the end of the trial. The termination of the trial could have resulted from any of the following: (1) The inability to reach 80% of the participant's maximal cut distance, as recorded at the beginning of the session; (2) A Rating-of-Fatigue of 9 or higher is reached, or a reading of 8 three consecutive times; (3) Noticeable change in body mechanics putting the participant at risk; (4) Volitional exhaustion by the participant. For all participants in this study, termination was a result of criteria (2) or (4). Because these two criteria represent the most subjective ratings from the participant, it is possible that there could have been other reasons for ending the trial. One possible reason is that, due to the nature of the fatiguing slideboard protocol, individuals reached a threshold of cardiovascular fatigue causing them to be unable to continue. In this study, we recorded the heart rate of participants after each cycle. Participants' heart rate, on average, quickly rose and began to plateau between around %Trial = 30. Therefore, it is possible that feelings of cardiovascular fatigue, rather than the intended muscular fatigue, contributed to the participants' ROF and when the trial was stopped. Despite this potential limitation, the safety of the participants in this study was a priority, and these precautions were set in place to minimize the increase in injury risk due to high levels of fatigue.

During the motion capture collection, there were moments when not all 45 markers were recognized in 3-dimensional space by the cameras. Prediction of the location of those markers, using the Cortex software, was required for those cases, creating the potential for some error in the exact positioning of these markers. There were also times when a retro-reflective marker would fall off the individual, either because of sweat removing the adhesive or from the participants physically knocking off the marker accidentally. This caused a very brief pause to occur before the next effort began, while the researcher rushed to re-attach the marker. Recovery during these brief pauses was a possibility.

With regard to the force plates, due to the dynamic nature of the cutting task, there were times when participants slipped upon landing on the force plate and this may have caused a different reading in the force application during that effort. As well, because we were unable to provide each participant with an identical pair of shoes, there may have been individual differences in the force impact and lateral control upon landing due to the variability in footwear composition. Participants were instructed to wear as close to a "court shoe" as possible, but this was as much as we could do to control this portion of the study.

Finally, in order to increase the control of the study, the athletic cutting task was performed in isolation and from a stationary position. During sport, this maneuver is most likely to occur in the midst of a run, and at a variety of angles from the planted foot. Therefore, it is possible that the effect during a 'game-time' situation maybe be more intense, and that our results reflect a more conservative situation.

CHAPTER 6

CONCLUSION

This study demonstrated that, as fatigue progressed during performance of repeated cutting maneuvers, there was an increase in the shear force at peak knee flexion in the anterior-posterior direction by approximately 20%, and a decrease in the impulse of the medial-lateral direction by about 7% during the weight acceptance phase. No significant difference was seen in the vertical ground reaction forces or in the resultant force of this task.

Kinetic data from this study revealed significant changes in both the hip and knee joint angles as fatigue progressed. The average hip angle at peak knee flexion from the beginning of the trial decreased by 5.7°; an 8.1% change in flexion. The average peak knee angle from the beginning of the trial decreased by 2.7°; a 3.7% change in peak flexion. These changes are indicative of a relatively more extended leg posture, which is associated with more anterior tibial translation, and increases the stress placed on the ACL during this loading task. There were no statistically significant changes in the angle of ankle flexion during the weight acceptance phase as fatigue progressed. Additionally, there were no statistically significant changes in hip adduction and hip rotation angles from the beginning of the trial to the end.

Lastly, there was a significant decrease in relative overall average muscle activation from the beginning to the end of trial by 11.9%. However, further investigation of pairwise comparisons indicated that, from 60%-100% of the trial, average muscle activation significantly increased by 9.3%. No significant difference was seen in sEMG amplitudes between muscles during this task.

6.1 FUTURE RESEARCH DIRECTIONS

It is recommended that, for future research in this area, an estimate of muscle force to accompany the monitored sEMG be calculated, in order to better understand the reasoning for the changes observed in sEMG amplitude. It may also be beneficial that a higher volume of training be done to reduce any potential effects of learning from occurring during the testing session. This can be done by adding more full testing sessions to the study, allowing the participants to return and repeat the protocol on more than one occasion. It may also be beneficial to have a different method of fatiguing participants rather than the slide board exercise which is less aerobic in nature, so that the subjective indications of fatigue from participants more reflect the extent of muscular fatigue rather than cardiovascular fatigue. These may reduce the effects of confounding factors on the data and allow for more control of the study.

Future studies should continue to examine the capacity of the structures of the lower extremity and the demands which pose the greatest risks to injury and performance. The data gained from this type of research could provide the insights needed to enhance the monitoring strategy and training techniques needed in order to identify and manage the presence of high injury risk situations in sport. In response to an editorial, calling for the exploration of 'critical mediators' serving as strong drivers of both neuromuscular fatigue and tissue damage by the British Journal of Sports Medicine, Harper and Kiely (2018) state that "empirically informed training strategies focused on increasing player resilience to the negative consequences of repeated decelerations are urgently required". By continuing to improve the methods in which we conduct these research studies, we provide the empirical information necessary to supplement the interventions necessary to aid in the reduction of the detrimental effects which follow fatiguing exercise.

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APPENDICES

APPENDIX A

PARTICIPANT DEMOGRAPHICS

Participant	Age (years)	Height (cm)	Weight (kg)	H:Q Ratio	Dominant Leg
P01	24	170	87.8	0.51	Right
P02	24	164	61.4	0.49	Right
P03	25	165.5	53.7	0.56	Right
P04	23	169	69.6	0.53	Right
P05	23	168.5	56	0.52	Right
P06	24	163	57.6	0.51	Right
P07	21	156.5	54	0.59	Right
P08	25	171	73.3	0.62	Right
P09	21	165	65.5	0.47	Right
P10	22	166	67	0.52	Right
P11	26	157	60.3	0.59	Right
P12	23	160	46.5	0.49	Left
Mean (SD)	**23.4 (1.6)**	**164.6 (4.8)**	**62.7 (10.9)**	**0.53 (0.05)**	

APPENDIX B

University of Windsor

Letter of Information

Examining the influence of fatigue on knee joint mechanics during an athletic cutting task

INVESTIGATORS: Joel Cort, Sara Santos, Jim Potvin, Chad Sutherland

PRINCIPAL INVESTIGATOR: Dr. Joel Cort –Associate Professor, Dept. of Kinesiology, University of Windsor

PURPOSE OF THIS RESEARCH

The purpose of this study is to investigate if impact forces (vertical ground reaction forces), during an athletic cutting task, increase after a fatiguing protocol. Generally, this study will investigate the effects of muscle fatigue on the performance of a cutting task. You will be asked to perform maximal side-cuts, onto a force plate, and lateral slides on a on a slide board until a certain level of fatigue is reached.

PROCEDURES INVOLVED IN THE RESEARCH

You will be invited to visit the Occupational Simulation and Ergonomics Laboratory (OSEL) at the University of Windsor (room HK209) on two separate occasions, with a minimum of 72-hours between the sessions. The first session will be treated as an orientation session, and the second session will be the testing session. You will be asked to wear athletic clothing (comfortable athletic shoes, shorts, and athletic shirt) to the orientation session. For the testing session, you are required to wear clothing that fits tightly to the skin, to accommodate the motion capture markers, inertial sensors and EMG electrodes. The methods will be outlined in detail below:

Orientation Session:

1. We will review the letter of information and consent form with you, describing all procedures in detail and outlining that you are free to withdraw from the study at any time, without penalty. After we answer any and all of your questions about the study, you will be asked to sign the Letter of Consent.
2. You will be asked to complete the lower-limb health questionnaire, which describes the health of your lower extremities (hip, knee, ankle and foot). Any answer of 'yes' on this questionnaire will cause you to be ineligible to participate in this study. Next, your height and leg length (meters), and weight (kilograms) will be measured and recorded on the questionnaire, as well as your age and gender.
3. We will guide you through a 10-minute generalized warm-up, including stationary cycling and dynamic stretching.
4. You will be seated on the Biodex (common exercise equipment used for training and rehabilitation purposes), and seat adjustments will be made so your knee, hip and ankle are all at 90 degree angles. You will then be instructed to complete a maximal leg extension effort (extending the lower leg away from the body; using your quadriceps), followed by a maximal leg flexion effort (flexing the lower leg

towards the body; using your hamstrings). This task represents a common leg extension/flexion exercise, which is typically included in all exercise training programs. We will then calculate your Hamstrings: Quadriceps force ratio (Flexion force/Extension force). If you are above 45%, you will be asked to continue with the research protocol.

5. We will instruct you on how to properly execute a maximum side-cut as well as the slide board exercise. You will then try these maneuvers for yourself, and we will give proper instructions related to body mechanics. Once you are able to properly execute a side cut and the slide board exercise, you will be given an opportunity to practice these techniques until you are comfortable with the motions, stride length and pace.
6. You will schedule your testing session, which will be a minimum of 72 hours after your orientation session. To accommodate the motion capture markers, Xsens sensors and the electromyography (EMG) electrodes, specific clothing requirements are necessary to complete this research during the testing session. For males, comfortable running shoes and tight fitting spandex shorts will be recommended. Furthermore, males will have the option of going shirtless or wearing a tight, spandex type t-shirt. For females, comfortable running shoes, tight fitting spandex shorts, and a sports bra or tight, spandex type t-shirt will be recommended. These clothing requirements are very common when completing athletic tasks using motion capture and EMG.
7. The orientation session will take no more than 1 hour and 30 minutes.

Testing Session:

1. When you return to OSEL, we will review all the procedures involved in the testing session and answer any questions you may 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. Furthermore, the questionnaire will ensure you have not completed any exercise in the previous 72 hours, and that you are not experiencing any leg muscle fatigue or delayed onset muscle soreness.
2. We will guide you through a 10-minute generalized warm-up, including stationary cycling and dynamic stretching. You will then practice maximal side-cutting (~10 cuts), similar to the orientation session. Next, you will be asked to perform 3 maximal cuts on the force plate.
3. Self-adhesive sEMG electrodes will be placed on the belly of your muscles (9): vastus lateralis (VL) vastus medialis (VM) rectus femoris (RF), biceps femoris (BF), semitendinosus (ST), gastrocnemius medial (GM), gastrocnemius lateral (GL), gracilis (GR), tibialis anterior (TA). An area (the size of a credit card) of your skin will be cleaned, at the location on the muscle, where the electrode will be placed: (1) shaving the hair with a disposable razor, (2) light skin abrasion (rubbed with cotton pad) and (3) disinfection with rubbing alcohol. This is required to maximize electrode adhesion (so they don't fall off) during the protocol, as well as to minimize the noise produced (between the electrode and skin) from skin movement. We will then use two fingers to palpate your specific muscle belly (e.g. VL) and verbally inform you where the specific sEMG electrodes will be placed.
4. After the sEMG electrodes are placed, you will then perform a 20 s-noise trial, requiring you to lay prone on a mat and remain as still as possible. Next, you will perform three 3 s isometric maximum voluntary exertions (MVEs) of the calves, quadriceps, hamstrings, gracilis and tibialis anterior:

Quadriceps (knee extensors): From a seated position (90˚ trunk, hip and knee angle), you will be instructed to maximally extend your knee against resistance.
Hamstrings (knee flexors): While lying in a prone position with the knee flexed at 90˚, you will be instructed to maximally flex your knee against resistance.
Gracilis (hip adductor): From a standing position with your hands supported, you will be instructed to maximally adduct your leg against resistance.
Calves (plantar flexors): From a seated position (90˚ trunk, hip and knee angle), you will be instructed to maximally plantar flex your ankle against resistance.
Tibialis Anterior (dorsi-flexor): From a seated position (90˚ trunk, hip and knee angle), you will be instructed to maximally dorsiflex your ankle against resistance.
Each of these efforts will be performed for 3 s, three times in succession, with 30 s of rest between efforts. The MVE will be used to determine the maximal amount of muscle activity, which is used to normalize the rectified sEMG signal of the testing data.

5. 17 inertial sensors will be placed on the following locations: hand (2), forearm (2), upper arm (2), shoulder (2), thigh (2), lower leg (2), feet (2), head (1), sternum (1) and pelvis (1). We will use two fingers to palpate the landmarks (e.g. medial and lateral elbow) and then verbally inform you where the sensor is going to be placed. The inertial sensors are secured to the body using elastic bands.
6. 45 retro-reflective motion capture markers (14 mm sphere) will be placed on your skin (or t- shirt), with an adhesive tape, to capture motion data. We will use two fingers to palpate the landmarks (e.g. medial and lateral elbow), and then verbally inform you where the marker is going to be placed. The retro-reflective markers will be placed over joints/segments, as to accurately document the movements during the testing session. With the markers placed, you will be required to complete a 5 s T-pose, which requires you to stand quietly with your arms abducted to 90˚. Next, you will be instructed through a range of motion trial (30 s), which includes body movements typically used while jumping (e.g. flexing the knees).
7. You will start the testing protocol by completing a 30 s quiet standing trial, requiring you to stand as quietly as possible. You will then begin sliding at a selected pace on the slide board, as cued by a metronome, making one stride for every beat for one minute. You will then be instructed to make 5 maximal side cuts on the force plate at a 45˚ angle. These will be demonstrated for you before you start. This procedure will be repeated until fatigue, which will be determined by the researcher based on visual and biomechanical measurements. You will receive feedback on your progress after each effort. You will be asked how you are feeling every 10 jumps and will be informed to stop if you feel your mechanics (hip, knee, ankle angles) of the cut technique has changed enough to put you at risk of an injury. Furthermore, if your cut technique becomes associated with an unacceptable risk, we will terminate the session. Lastly, you will complete a second 30 s quiet standing trial immediately following protocol termination. The maximal cut testing protocol will be video recorded to verify you are landing with your foot on the force place.
8. Surface EMG electrodes, inertial sensors and retro-reflective markers will be removed immediately following the protocol. You will then be guided through a cool-down session including stationary cycling (5-8 minutes) and a lower body static stretching routine (8 minutes), to alleviate any potential muscle fatigue symptoms. You will then be asked if you have any questions, and if not, you are free to leave the lab. You will be reminded that, if you are interested in the results of the study, you must leave your email address with us.

9. The testing session will take no more than 2 hours and 30 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 in 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. Sara Santos 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 cannot 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

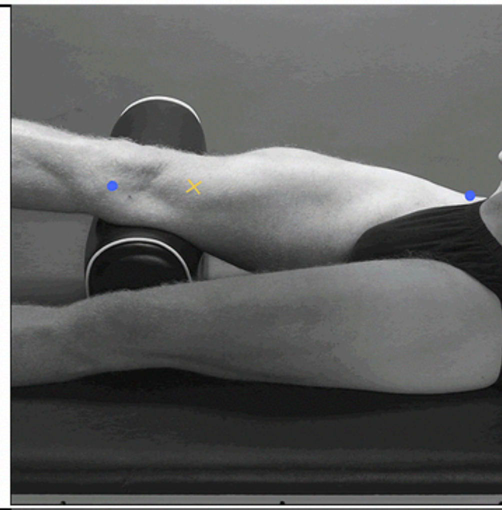
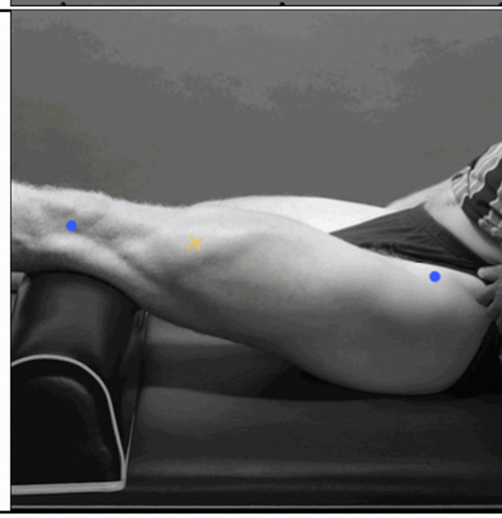
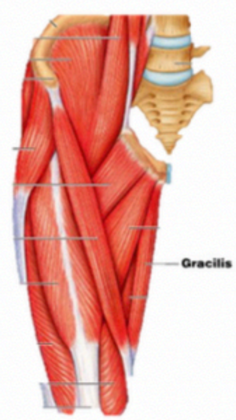
____
Date

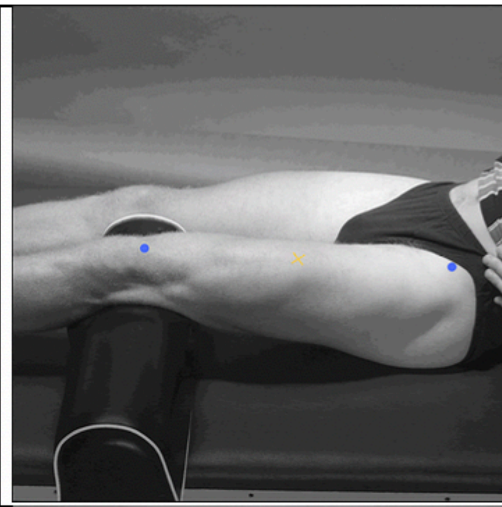
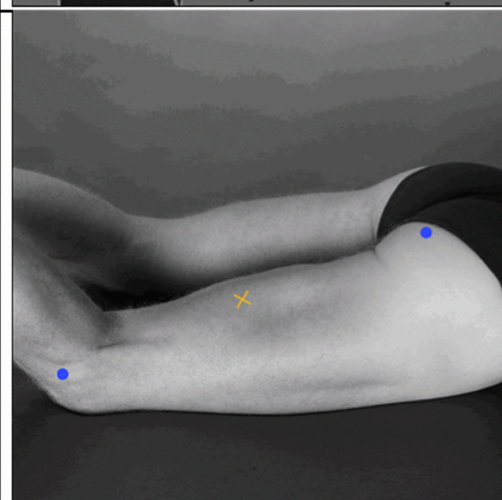
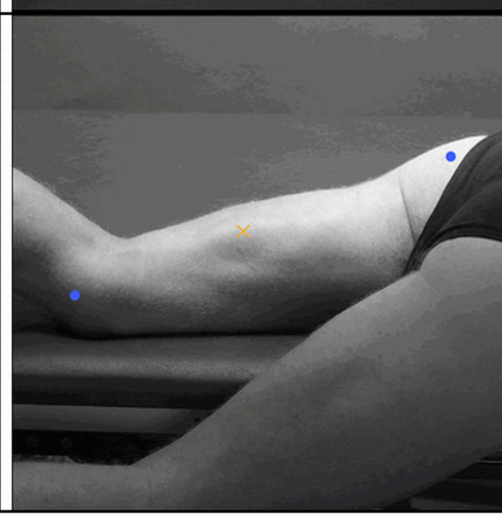
____
SIGNATURE OF INVESTIGATOR

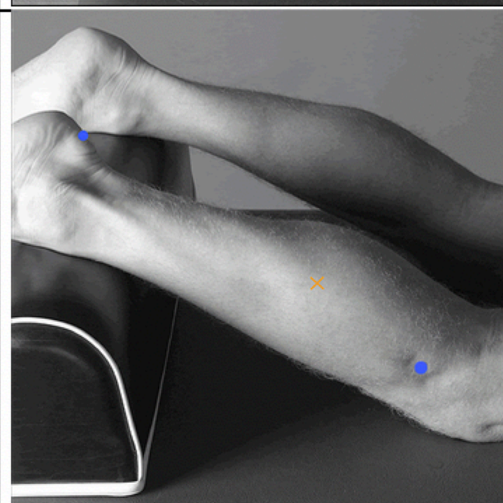
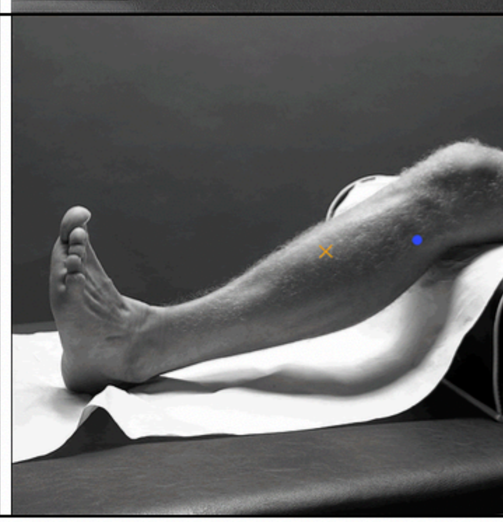
These are the terms under which I will conduct research.

APPENDIX C

EMG Sensor Placement

Vastus Medialis	80% of the line between ASIS and joint space in front of anterior border of MCL	
Vastus Lateralis	2/3 on the line from the ASIS to the lateral side of the patella	
Gracilis	1/3 of the way from the pubic tubercle to the medial edge of the knee joint	

Rectus Femoris	50% on the line from the ASIS to superior part of patella	
Biceps Femoris	50% on the line between the ischial tuberosity and lateral epicondyle of tibia	
Semitendinosus	50% on the line between the ischial tuberosity and the medial epicondyle of tibia	

Gastrocnemius Medialis	Most prominent bulge of muscle	
Gastrocnemius Lateralis	1/3 of the line between the head of the fibula and the heel	
Tibialis Anterior	1/3 on the line between the tip of the fibula and the tip of medial malleolus	

APPENDIX D

Retro-Reflective Marker Placement

Number	Segment	Label	Placement
1	Head	Top Head	Apex of skull
2		Front Head	Forehead at hairline center
3		Rear head	Occipital protuberance
4	Neck	Neck	C7
5	Back	Back Right Offset	Interior to apex of scapula at T8-10 level
6		Lower Back	L4/L5 Vertebrae
7	Chest	Sternum	Manubrium
8,9	Shoulder	R/L Shoulder	Superior aspect of the acromion process
10,11	Upper Arm	R/L Triceps	Posterior surface, midpoint of triceps
12,13	Elbow	R/L Elbow Lateral	Lateral epicondyle of the humerus
14,15		R/L Elbow Medial	Medial epicondyle of the humerus
16,17	Forearm	R/L Forearm	Lateral surface, midpoint of the forearm
18,19	Wrist	R/L Wrist Lateral	Radial styloid process
20,21		R/L Wrist Medial	Ulnar styloid process
22,23	Pelvis	R/L ASIS	Anterior superior iliac spine
24,25		R/L PSIS	Posterior superior iliac spine
26,27	Upper Leg	R/L Trochanter	Lateral surface of greater trochanter
28,29		R/L Thigh	Anterior surface, midpoint of the upper leg
30,31	Knee	R/L Knee Lateral	Lateral epicondyle of femur
32,33		R/L Knee Medial	Medial epicondyle of femur
34,35	Shank	R/L Middle leg	Lateral surface, midpoint of the shank
36,37		R/L Lower Leg	Anterior surface, distal 2/3 of the shank
38,39	Ankle	R/L Ankle Lateral	Lateral malleolus
40,41		R/L Ankle Medial	Medial malleolus
42,43	Foot	R/L Calcaneus	Calcaneus
44,45		R/L Toe	Top of shoe over the big toe

APPENDIX E

LOWER LIMB HEALTH QUESTIONNAIRE

Participant Code: ____________________

Date: ____________

Have you, at any time throughout the previous 2 years, had any discomfort, injury or disorders of the following lower extremities (including but not limited to aches, pain, general discomfort, strain, sprain, fracture, numbness):

Hip:	Yes	No
Knee:	Yes	No
Ankle:	Yes	No
Foot:	Yes	No

Have your activities of daily living been negatively affected by any of the following lower extremities, in the past 2 years?

Hip:	Yes	No
Knee:	Yes	No
Ankle:	Yes	No
Foot:	Yes	No

Have you ever required surgery on your:

Hip:	Yes	No
Knee:	Yes	No
Ankle:	Yes	No
Foot:	Yes	No

Height: __________ **Lower Limb Length:** __________ **Stride Length:** __________

Weight: __________ **Age:** _________

APPENDIX F

CSEP | SCPE
THE GOLD STANDARD IN EXERCISE SCIENCE AND PERSONAL TRAINING

Get Active Questionnaire

CANADIAN SOCIETY FOR EXERCISE PHYSIOLOGY –
PHYSICAL ACTIVITY TRAINING FOR HEALTH (CSEP-PATH®)

Physical activity improves your physical and mental health. Even small amounts of physical activity are good, and more is better.

For almost everyone, the benefits of physical activity far outweigh any risks. For some individuals, specific advice from a Qualified Exercise Professional (QEP – has post-secondary education in exercise sciences and an advanced certification in the area – see csep.ca/certifications) or health care provider is advisable. **This questionnaire is intended for all ages – to help move you along the path to becoming more physically active.**

☐ I am completing this questionnaire for myself.

☐ I am completing this questionnaire for my child/dependent as parent/guardian.

PREPARE TO BECOME MORE ACTIVE

The following questions will help to ensure that you have a safe physical activity experience. Please answer **YES** or **NO** to each question before you become more physically active. If you are unsure about any question, answer **YES**.

YES	NO	
		1 Have you experienced **ANY** of the following (A to F) **within the past six months**?
○	○	**A** A diagnosis of/treatment for heart disease or stroke, or pain/discomfort/pressure in your chest during activities of daily living or during physical activity?
○	○	**B** A diagnosis of/treatment for high blood pressure (BP), or a resting BP of 160/90 mmHg or higher?
○	○	**C** Dizziness or lightheadedness during physical activity?
○	○	**D** Shortness of breath at rest?
○	○	**E** Loss of consciousness/fainting for any reason?
○	○	**F** Concussion?
○	○	**2** Do you currently have pain or swelling in any part of your body (such as from an injury, acute flare-up of arthritis, or back pain) that affects your ability to be physically active?
○	○	**3** Has a health care provider told you that you should avoid or modify certain types of physical activity?
○	○	**4** Do you have any other medical or physical condition (such as diabetes, cancer, osteoporosis, asthma, spinal cord injury) that may affect your ability to be physically active?

NO to all questions: go to Page 2 – ASSESS YOUR CURRENT PHYSICAL ACTIVITY

YES to any question: go to Reference Document – ADVICE ON WHAT TO DO IF YOU HAVE A YES RESPONSE

Get Active Questionnaire

ASSESS YOUR CURRENT PHYSICAL ACTIVITY

Answer the following questions to assess how active you are now.

1 During a typical week, on how many days do you do moderate- to vigorous-intensity aerobic physical activity (such as brisk walking, cycling or jogging)? ______ DAYS/ WEEK

2 On days that you do at least moderate-intensity aerobic physical activity (e.g., brisk walking), for how many minutes do you do this activity? ______ MINUTES/ DAY

For adults, please multiply your average number of days/week by the average number of minutes/day: ______ MINUTES/ WEEK

Canadian Physical Activity Guidelines recommend that adults accumulate at least 150 minutes of moderate- to vigorous-intensity physical activity per week. For children and youth, at least 60 minutes daily is recommended. Strengthening muscles and bones at least two times per week for adults, and three times per week for children and youth, is also recommended (see csep.ca/guidelines).

GENERAL ADVICE FOR BECOMING MORE ACTIVE

Increase your physical activity gradually so that you have a positive experience. Build physical activities that you enjoy into your day (e.g., take a walk with a friend, ride your bike to school or work) and reduce your sedentary behaviour (e.g., prolonged sitting).

If you want to do **vigorous-intensity physical activity** (i.e., physical activity at an intensity that makes it hard to carry on a conversation), and you do not meet minimum physical activity recommendations noted above, consult a Qualified Exercise Professional (QEP) beforehand. This can help ensure that your physical activity is safe and suitable for your circumstances.

Physical activity is also an important part of a healthy pregnancy.

Delay becoming more active if you are not feeling well because of a temporary illness.

DECLARATION

To the best of my knowledge, all of the information I have supplied on this questionnaire is correct. If my health changes, I will complete this questionnaire again.

I answered NO to all questions on Page 1

Sign and date the Declaration below

I answered YES to any question on Page 1

Check the box below that applies to you:

- ☐ I have consulted a health care provider or Qualified Exercise Professional (QEP) who has recommended that I become more physically active.
- ☐ I am comfortable with becoming more physically active on my own without consulting a health care provider or QEP.

Name (+ Name of Parent/Guardian if applicable) [Please print] ______ Signature (or Signature of Parent/Guardian if applicable) ______ Date of Birth ______

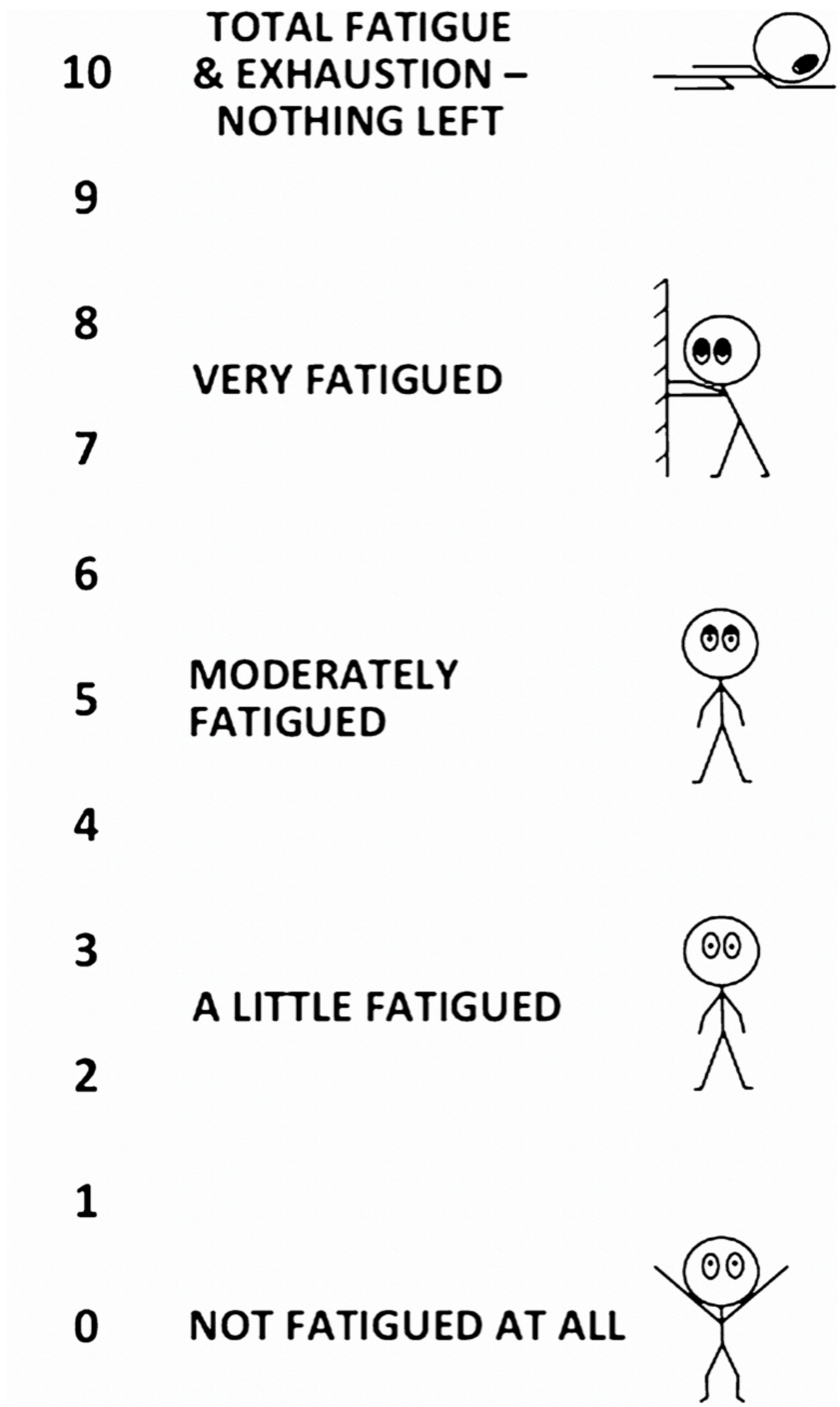
Date ______ Email (optional) ______ Telephone (optional) ______

With planning and support you can enjoy the benefits of becoming more physically active. A QEP can help.

☐ Check this box if you would like to consult a QEP about becoming more physically active. (This completed questionnaire will help the QEP get to know you and understand your needs.)

APPENDIX G

RATING OF FATIGUE SCALE



APPENDIX H

RATING OF FATIGUE SCALE INSTRUCTIONS

Familiarize yourself with the scale by looking closely at the ROF scale now. You will notice that the ROF scale consists of 11 numerical points that range from 0 to 10. There are also five descriptors and five diagrams that are intended to help you understand the scale and make your rating.

When you are presented with the ROF scale please carefully inspect the scale before giving a numerical response from 0 to 10. Always try to respond as honestly as possible giving a rating that best reflects how fatigued you feel at the time.

Try not to hesitate too much and make sure you only give ONE number as a response. For example, avoid responding by giving two numbers such as 'three or four'.

Now please read the following examples of what some of the ROF ratings mean:

A response of 0 would indicate that you do not feel at all fatigued. An example of this might be soon after you wake up in the morning after having a good night's sleep. Now try to think of a similar occasion in your past where you have experienced the lowest feelings of fatigue and use this as your reference.

A response of 10 would indicate that you feel totally fatigued and exhausted. An example of this might be not being able to stay awake, perhaps late at night but equally could include situations such as sprinting until you can no longer physically continue. Again, try to think of a similar example that you have actually experienced in the past.

APPENDIX I

RETURNING PARTICIPANT QUESTIONNAIRE

Participant Code: _________

In the past 72 hours, have you experienced any muscle, tissue or joint pain, or discomfort, in any of the following areas:

Hip: Yes No
Knee: Yes No
Ankle: Yes No
Foot: Yes No

Have you completed any strenuous physical activity (intense exercise exertions, such as athletic competition, or any personal training) in the past 72 hours?

YES **NO**

APPENDIX J

TRIAL DATA

Table 7: Heart rate data (bpm)

	%Trial 0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
P01	66	130	163	176	180	184	186	187	190	191	188
P02	78	137	181	186	190	194	194	196	198	199	200
P03	68	82	102	160	161	170	170	174	176	177	179
P04	78	143	159	163	137	162	168	171	165	161	174
P05	86	144	173	182	184	184	183	185	188	189	189
P06	78	147	156	166	167	167	172	169	169	172	171
P07	85	158	166	178	176	179	178	180	181	183	183
P08	80	135	133	157	156	154	160	154	152	158	162
P09	86	158	183	186	188	186	189	189	190	191	190
P10	71	152	160	168	165	167	169	171	171	170	171
P11	87	164	174	175	180	181	184	182	183	188	189
P12	89	130	161	168	173	178	179	181	183	184	183
Mean	**79**	**140**	**159**	**173**	**175**	**177**	**179**	**179**	**180**	**182**	**182**

Table 8: Rating of Fatigue data

	%Trial 0%	10%	20%	30%	40%	50%	60%	70%	80%	90%	100%
P01	0	2	3	4	5	6	7	7	8	8	9
P02	0	2	4	5	6	7	7	8	8	8	8
P03	0	1	2	3	4	5	6	7	7	8	8
P04	1	3	4	5	6	7	8	8	8	8	9
P05	0	1	2	3	4	5	6	7	8	8	9
P06	0	2	4	5	5	6	7	7	7	8	8
P07	0	1	3	5	6	7	8	8	8	9	9
P08	1	3	4	5	5	6	6	7	7	7	8
P09	0	1	2	3	4	5	6	7	8	9	10
P10	1	2	3	4	5	6	6	7	7	8	8
P11	1	3	3	4	5	5	6	7	7	8	9
P12	0	2	3	5	6	7	8	8	8	9	9
Mean	**0.27**	**1.82**	**3.00**	**4.18**	**5.00**	**5.91**	**6.64**	**7.27**	**7.55**	**8.18**	**8.64**

Table 9: Resultant Force Data

	%Trial										
	0%	**10%**	**20%**	**30%**	**40%**	**50%**	**60%**	**70%**	**80%**	**90%**	**100%**
Angle (Degrees)											
P01	81.7	81.2	80.8	80.4	80.1	79.9	79.8	79.7	79.7	79.8	79.9
P02	85.7	85.4	85.1	84.9	84.8	84.7	84.7	84.7	84.9	85.0	85.2
P03	80.1	80.1	80.1	80.1	79.9	79.7	79.4	79.0	78.6	78.0	77.4
P05	86.8	86.4	86.1	85.8	85.6	85.5	85.3	85.3	85.3	85.3	85.4
P06	85.0	84.5	84.1	83.6	83.2	82.8	82.4	82.1	81.8	81.5	81.3
P07	78.1	77.7	77.3	77.0	76.8	76.7	76.8	76.9	77.2	77.5	77.9
P08	84.7	85.4	85.9	86.2	86.5	86.6	86.7	86.6	86.5	86.2	85.9
P09	84.1	84.0	83.8	83.7	83.7	83.7	83.7	83.9	84.1	84.3	84.6
P10	83.3	84.3	85.0	85.5	85.7	85.8	85.6	85.3	84.7	83.8	82.7
P11	82.1	82.5	82.8	83.0	83.0	82.9	82.7	82.3	81.8	81.1	80.2
P12	69.1	69.4	69.6	69.8	69.9	69.9	69.8	69.7	69.5	69.2	68.8
Mean	**81.9**	**81.9**	**81.9**	**81.8**	**81.7**	**81.7**	**81.5**	**81.4**	**81.3**	**81.1**	**80.8**
Effective Duration (s)											
P01	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2
P02	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2
P03	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.3
P05	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2
P06	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2
P07	0.1	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.1	0.1	0.1
P08	0.1	0.1	0.1	0.1	0.1	0.1	0.1	0.1	0.1	0.1	0.1
P09	0.1	0.2	0.2	0.2	0.2	0.2	0.2	0.1	0.1	0.1	0.1
P10	0.2	0.2	0.2	0.1	0.1	0.1	0.1	0.1	0.1	0.1	0.1
P11	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2
P12	0.2	0.2	0.2	0.2	0.3	0.3	0.3	0.3	0.3	0.3	0.3
Mean	**0.2**	**0.2**	**0.2**	**0.2**	**0.2**	**0.2**	**0.2**	**0.2**	**0.2**	**0.2**	**0.2**
Impulse (N-s)											
P01	278	270	263	257	252	248	245	242	241	241	241
P02	200	203	204	205	205	204	203	200	197	193	188
P03	192	186	180	177	175	174	175	177	181	186	193
P05	181	177	174	171	168	166	165	164	163	164	164
P06	170	176	181	185	187	188	187	185	182	178	172
P07	133	132	132	131	130	130	129	128	127	126	125
P08	204	193	184	177	172	169	168	170	173	178	186
P09	168	166	164	163	161	159	158	156	154	152	150
P10	243	225	211	199	190	185	182	183	186	193	203
P11	196	185	175	167	162	158	157	158	161	167	174
P12	178	181	184	189	194	200	206	214	222	230	240

Mean	**195**	**190**	**187**	**184**	**181**	**180**	**180**	**180**	**181**	**183**	**185**
Peak (N)											
P01	1337	1368	1396	1421	1442	1460	1475	1487	1495	1499	1501
P02	884	882	882	884	888	895	904	915	929	945	963
P03	786	796	803	807	807	805	799	791	779	764	747
P05	903	902	902	904	907	910	915	921	928	936	946
P06	973	952	934	919	907	899	895	893	895	901	909
P07	888	864	847	834	828	827	832	843	859	881	909
P08	1423	1476	1520	1554	1579	1595	1600	1596	1583	1559	1527
P09	1127	1096	1071	1053	1041	1036	1037	1044	1058	1079	1106
P10	1313	1348	1377	1399	1414	1423	1424	1418	1406	1387	1362
P11	959	979	994	1005	1010	1011	1006	997	983	964	941
P12	793	781	772	765	759	756	754	755	757	762	768
Mean	**1023**	**1028**	**1033**	**1037**	**1041**	**1043**	**1046**	**1047**	**1048**	**1048**	**1047**

Table 10: FZ Force Data (Vertical Ground Reaction Force)

	%Trial										
	0%	**10%**	**20%**	**30%**	**40%**	**50%**	**60%**	**70%**	**80%**	**90%**	**100%**
Effective Duration (s)											
P01	0.20	0.19	0.18	0.17	0.17	0.16	0.16	0.15	0.15	0.15	0.15
P02	0.22	0.22	0.22	0.22	0.22	0.22	0.21	0.21	0.20	0.19	0.18
P03	0.23	0.22	0.21	0.21	0.20	0.20	0.21	0.21	0.22	0.23	0.25
P05	0.19	0.18	0.18	0.18	0.17	0.17	0.17	0.17	0.16	0.16	0.16
P06	0.16	0.17	0.18	0.19	0.19	0.20	0.20	0.20	0.19	0.19	0.18
P07	0.14	0.14	0.15	0.15	0.15	0.15	0.15	0.14	0.14	0.13	0.13
P08	0.13	0.12	0.11	0.10	0.10	0.10	0.09	0.10	0.10	0.10	0.11
P09	0.14	0.14	0.14	0.15	0.15	0.15	0.14	0.14	0.14	0.13	0.13
P10	0.17	0.16	0.14	0.13	0.12	0.12	0.12	0.12	0.12	0.13	0.14
P11	0.19	0.18	0.17	0.16	0.15	0.15	0.15	0.15	0.16	0.17	0.18
P12	0.23	0.24	0.24	0.25	0.26	0.27	0.28	0.29	0.30	0.31	0.32
Mean	**0.18**	**0.18**	**0.18**	**0.17**	**0.17**	**0.17**	**0.17**	**0.17**	**0.17**	**0.17**	**0.18**
Impulse (N-s)											
P01	265	256	248	242	236	232	228	226	225	225	225
P02	190	191	193	193	193	192	190	187	184	180	175
P03	180	173	167	163	161	160	161	164	168	174	181
P05	166	163	160	157	155	153	152	151	151	151	152
P06	159	164	169	172	174	175	175	174	171	168	163
P07	120	120	120	119	119	118	117	116	115	114	113
P08	184	174	166	159	155	152	152	153	156	162	169
P09	154	154	153	152	151	149	148	146	145	143	141
P10	226	208	194	182	174	168	166	167	171	178	188
P11	184	173	164	157	152	149	148	149	152	157	164
P12	169	172	176	180	186	191	198	205	213	221	230

Mean	**182**	**177**	**174**	**171**	**169**	**167**	**167**	**167**	**168**	**170**	**173**
Peak (N)											
P01	1323	1352	1378	1401	1421	1438	1452	1463	1471	1476	1478
P02	882	879	878	880	884	891	900	911	925	941	960
P03	774	784	791	794	795	792	786	776	764	748	729
P05	901	900	900	902	904	907	912	918	925	933	943
P06	970	947	928	913	901	892	887	885	886	891	899
P07	869	844	826	813	806	805	810	821	838	861	889
P08	1417	1471	1516	1551	1576	1592	1597	1593	1580	1556	1523
P09	1121	1090	1065	1046	1035	1029	1030	1038	1053	1074	1101
P10	1304	1342	1372	1395	1410	1419	1420	1414	1400	1379	1351
P11	950	971	986	997	1003	1003	998	988	973	952	927
P12	740	731	724	717	713	710	708	708	709	712	716
Mean	**1023**	**1028**	**1033**	**1037**	**1041**	**1043**	**1046**	**1047**	**1048**	**1048**	**1047**

Table 11: FY Force Data (Anterior-Posterior Shear)

	%Trial										
	0%	**10%**	**20%**	**30%**	**40%**	**50%**	**60%**	**70%**	**80%**	**90%**	**100%**
Impulse (N-s)											
P01	20.37	23.47	26.01	28.00	29.43	30.31	30.63	30.40	29.61	28.27	26.38
P02	9.74	10.92	11.85	12.54	12.99	13.20	13.17	12.89	12.37	11.61	10.61
P03	17.35	16.85	16.54	16.40	16.45	16.68	17.10	17.70	18.48	19.45	20.60
P05	9.14	8.65	8.28	8.05	7.94	7.97	8.13	8.42	8.84	9.39	10.07
P06	9.59	10.86	12.04	13.13	14.12	15.02	15.83	16.55	17.18	17.71	18.15
P07	17.78	17.78	17.78	17.78	17.79	17.80	17.81	17.83	17.85	17.88	17.91
P08	13.18	11.17	9.52	8.24	7.31	6.75	6.54	6.70	7.22	8.09	9.33
P09	9.79	10.58	11.18	11.58	11.79	11.81	11.64	11.27	10.71	9.96	9.01
P10	15.64	13.69	12.08	10.79	9.82	9.19	8.88	8.91	9.26	9.94	10.95
P11	15.75	14.76	14.04	13.61	13.45	13.56	13.96	14.63	15.58	16.81	18.32
P12	15.58	15.95	16.56	17.42	18.52	19.87	21.47	23.31	25.40	27.74	30.32
Mean	**13.99**	**14.06**	**14.17**	**14.32**	**14.51**	**14.74**	**15.02**	**15.33**	**15.68**	**16.08**	**16.51**

Peak (N)											
P01	190.1	207.8	223.2	236.3	247.0	255.4	261.4	265.1	266.4	265.4	262.0
P02	65.3	70.3	74.5	77.9	80.6	82.4	83.4	83.7	83.1	81.8	79.7
P03	133.1	134.1	135.6	137.4	139.6	142.3	145.3	148.7	152.5	156.7	161.3
P05	50.6	56.2	61.2	65.5	69.0	71.9	74.1	75.6	76.4	76.5	75.9
P06	85.1	90.9	96.6	102.2	107.7	113.0	118.3	123.4	128.4	133.3	138.0
P07	182.2	184.2	186.0	187.4	188.5	189.4	189.9	190.1	190.0	189.6	188.8
P08	131.9	119.3	109.0	101.0	95.2	91.7	90.6	91.6	95.0	100.6	108.5
P09	94.0	97.8	100.6	102.6	103.6	103.8	103.0	101.3	98.7	95.3	90.9
P10	153.3	133.7	118.9	108.9	103.7	103.2	107.5	116.6	130.4	149.1	172.5
P11	128.4	123.5	120.4	119.0	119.3	121.4	125.2	130.8	138.1	147.2	158.0
P12	82.9	85.1	87.8	91.0	94.7	98.9	103.5	108.7	114.4	120.6	127.2
Mean	**117.9**	**118.5**	**119.4**	**120.8**	**122.6**	**124.8**	**127.5**	**130.5**	**134.0**	**137.8**	**142.1**

Table 12: FX Force Data (Medial-Lateral Shear)

	%Trial										
	0%	**10%**	**20%**	**30%**	**40%**	**50%**	**60%**	**70%**	**80%**	**90%**	**100%**
Impulse (N-s)											
P01	83.20	83.11	83.02	82.93	82.84	82.75	82.66	82.58	82.49	82.40	82.32
P02	62.91	65.21	67.09	68.55	69.59	70.20	70.38	70.15	69.49	68.41	66.90
P03	65.40	65.52	65.58	65.58	65.53	65.43	65.27	65.05	64.78	64.46	64.08
P05	71.57	69.63	67.90	66.39	65.09	64.00	63.13	62.48	62.04	61.81	61.80
P06	61.47	63.43	64.79	65.55	65.71	65.26	64.22	62.57	60.32	57.47	54.02
P07	54.01	52.77	51.75	50.95	50.36	49.99	49.84	49.90	50.18	50.67	51.39
P08	87.14	82.87	79.32	76.49	74.38	73.00	72.34	72.40	73.18	74.69	76.92
P09	65.27	62.31	59.73	57.52	55.68	54.22	53.14	52.42	52.09	52.13	52.54
P10	89.68	85.58	82.09	79.23	76.97	75.34	74.32	73.93	74.14	74.98	76.43
P11	66.21	61.47	57.58	54.54	52.35	51.02	50.53	50.90	52.11	54.18	57.09
P12	54.13	53.43	53.04	52.96	53.18	53.71	54.55	55.69	57.14	58.90	60.96
Mean	**69.18**	**67.76**	**66.54**	**65.52**	**64.70**	**64.08**	**63.67**	**63.46**	**63.45**	**63.65**	**64.04**

Peak (N)											
P01	-28.3	-21.1	-15.3	-10.8	-7.5	-5.7	-5.1	-5.8	-7.9	-11.3	-16.0
P02	9.4	8.0	6.8	5.8	5.1	4.5	4.1	3.9	4.0	4.2	4.6
P03	-25.2	-23.9	-22.9	-22.1	-21.6	-21.4	-21.5	-21.8	-22.4	-23.2	-24.3
P05	3.2	1.3	-0.3	-1.6	-2.6	-3.4	-3.8	-3.9	-3.8	-3.3	-2.6
P06	-7.7	-6.5	-5.5	-4.6	-4.0	-3.5	-3.1	-3.0	-3.0	-3.1	-3.5
P07	-15.4	-14.9	-14.6	-14.5	-14.7	-15.2	-15.9	-16.8	-18.0	-19.4	-21.1
P08	-6.9	-9.3	-11.5	-13.5	-15.2	-16.7	-17.9	-18.9	-19.6	-20.1	-20.4
P09	-67.3	-61.5	-56.6	-52.7	-49.6	-47.4	-46.2	-45.9	-46.5	-47.9	-50.3
P10	-8.7	-11.7	-13.9	-15.2	-15.8	-15.5	-14.5	-12.6	-9.8	-6.3	-2.0
P11	-33.1	-32.0	-31.0	-30.1	-29.4	-28.9	-28.5	-28.2	-28.1	-28.1	-28.3
Mean	**-18.0**	**-17.2**	**-16.5**	**-15.9**	**-15.5**	**-15.3**	**-15.2**	**-15.3**	**-15.5**	**-15.9**	**-16.4**

Table 13: Joint Angle Data (Degrees)

		%Trial										
		0%	**10%**	**20%**	**30%**	**40%**	**50%**	**60%**	**70%**	**80%**	**90%**	**100%**
Hip Internal Rotation												
	P01	-6.92	-3.29	-0.49	1.48	2.63	2.94	2.43	1.10	-1.07	-4.06	-7.88
	P02	50.15	44.19	39.55	36.21	34.18	33.45	34.04	35.93	39.12	43.63	49.44
	P03	-0.19	2.78	5.21	7.10	8.44	9.23	9.47	9.17	8.32	6.93	4.99
	P05	16.78	20.60	23.29	24.85	25.27	24.57	22.73	19.75	15.65	10.41	4.04
	P06	24.13	23.22	22.61	22.28	22.25	22.51	23.05	23.89	25.02	26.44	28.15
	P07	4.76	5.33	5.74	6.00	6.11	6.07	5.88	5.54	5.04	4.40	3.60
	P08	34.81	35.76	36.13	35.91	35.11	33.72	31.76	29.21	26.08	22.37	18.07
	P09	6.45	6.75	6.93	7.00	6.94	6.77	6.48	6.07	5.54	4.89	4.13
	P10	6.44	9.72	12.24	14.00	15.00	15.24	14.72	13.44	11.40	8.61	5.05
	P11	-16.35	-11.24	-7.20	-4.22	-2.30	-1.44	-1.65	-2.91	-5.24	-8.62	-13.07
	P12	13.77	17.36	19.88	21.34	21.73	21.05	19.29	16.47	12.59	7.63	1.60
	Mean	**12.17**	**13.74**	**14.90**	**15.63**	**15.94**	**15.83**	**15.29**	**14.33**	**12.95**	**11.15**	**8.92**
Hip Adduction												
	P01	-5.33	-5.19	-5.03	-4.85	-4.65	-4.43	-4.20	-3.94	-3.67	-3.38	-3.07
	P02	-4.32	-3.75	-3.44	-3.40	-3.61	-4.09	-4.84	-5.84	-7.11	-8.64	-10.43
	P03	-6.31	-5.90	-5.45	-4.95	-4.40	-3.81	-3.17	-2.49	-1.76	-0.98	-0.16
	P05	-13.80	-15.22	-16.05	-16.29	-15.95	-15.02	-13.50	-11.40	-8.71	-5.43	-1.57
	P06	-0.99	-1.09	-1.07	-0.94	-0.70	-0.34	0.13	0.72	1.42	2.23	3.16
	P07	-6.96	-6.69	-6.55	-6.53	-6.63	-6.85	-7.19	-7.65	-8.24	-8.94	-9.77
	P08	-12.61	-15.02	-16.74	-17.77	-18.11	-17.75	-16.70	-14.96	-12.53	-9.40	-5.58

	P09	-4.84	-4.51	-4.26	-4.10	-4.01	-4.01	-4.10	-4.26	-4.51	-4.84	-5.26
	P10	-1.41	-0.96	-0.43	0.18	0.86	1.63	2.47	3.39	4.38	5.46	6.61
	P11	7.17	2.56	-1.20	-4.10	-6.15	-7.35	-7.69	-7.18	-5.82	-3.61	-0.54
	P12	-6.41	-5.89	-5.28	-4.57	-3.77	-2.87	-1.87	-0.78	0.40	1.68	3.05
	Mean	**-5.08**	**-5.61**	**-5.96**	**-6.12**	**-6.10**	**-5.90**	**-5.52**	**-4.95**	**-4.19**	**-3.26**	**-2.14**
Hip Flexion												
	P01	73.52	75.63	77.25	78.38	79.02	79.17	78.83	77.99	76.67	74.86	72.55
	P02	95.09	90.10	85.80	82.21	79.32	77.13	75.64	74.86	74.77	75.39	76.70
	P03	77.72	77.25	76.93	76.75	76.72	76.83	77.08	77.47	78.01	78.68	79.51
	P05	66.20	66.69	66.84	66.65	66.11	65.24	64.02	62.47	60.57	58.33	55.75
	P06	71.96	69.95	68.28	66.94	65.94	65.28	64.95	64.96	65.31	65.99	67.01
	P07	50.11	50.50	50.77	50.93	50.98	50.91	50.74	50.44	50.04	49.52	48.89
	P08	64.62	65.44	66.00	66.29	66.33	66.09	65.60	64.84	63.81	62.52	60.97
	P09	53.27	53.59	53.88	54.14	54.37	54.58	54.75	54.89	55.01	55.10	55.16
	P10	66.65	63.37	60.56	58.22	56.35	54.94	54.01	53.55	53.56	54.04	54.99
	P11	70.23	72.67	74.45	75.57	76.02	75.81	74.93	73.39	71.18	68.31	64.77
	P12	82.95	82.83	82.52	82.01	81.29	80.38	79.27	77.97	76.46	74.75	72.85
	Mean	**70.21**	**69.82**	**69.39**	**68.92**	**68.40**	**67.85**	**67.26**	**66.62**	**65.94**	**65.23**	**64.47**
Knee Flexion												
	P01	76.20	74.76	73.46	72.28	71.24	70.32	69.53	68.88	68.35	67.95	67.69
	P02	71.73	70.30	69.17	68.32	67.76	67.49	67.51	67.82	68.42	69.31	70.49
	P03	80.44	79.12	78.00	77.10	76.41	75.93	75.65	75.59	75.74	76.09	76.66
	P05	71.96	71.69	71.39	71.07	70.72	70.34	69.94	69.50	69.04	68.55	68.04
	P06	73.74	73.15	72.63	72.18	71.79	71.47	71.22	71.03	70.91	70.85	70.86
	P07	75.18	74.08	73.16	72.41	71.85	71.47	71.27	71.25	71.40	71.74	72.26
	P08	75.22	74.17	73.34	72.72	72.30	72.10	72.11	72.33	72.76	73.40	74.25
	P09	75.46	75.23	75.01	74.80	74.60	74.41	74.23	74.06	73.90	73.74	73.60
	P10	71.96	71.05	70.29	69.67	69.21	68.89	68.72	68.70	68.83	69.10	69.52
	P11	67.28	69.42	71.08	72.25	72.93	73.13	72.84	72.06	70.80	69.05	66.81
	P12	77.89	78.39	78.75	78.98	79.08	79.05	78.88	78.58	78.15	77.59	76.90
	Mean	**74.28**	**73.76**	**73.30**	**72.89**	**72.54**	**72.24**	**71.99**	**71.80**	**71.66**	**71.58**	**71.55**
Ankle Dorsiflexion												
	P01	33.76	25.89	20.17	16.59	15.15	15.85	18.68	23.66	30.77	40.02	51.42
	P02	33.07	33.10	33.19	33.33	33.52	33.77	34.07	34.43	34.84	35.31	35.83
	P03	37.83	36.84	36.10	35.58	35.31	35.27	35.47	35.90	36.57	37.48	38.63
	P05	40.91	40.68	40.46	40.25	40.03	39.83	39.62	39.42	39.23	39.03	38.85
	P06	30.45	30.65	30.57	30.21	29.57	28.65	27.45	25.97	24.22	22.18	19.86
	P07	36.13	35.86	35.58	35.30	35.01	34.72	34.41	34.10	33.78	33.46	33.12
	P08	47.13	46.53	46.02	45.60	45.27	45.04	44.90	44.86	44.90	45.04	45.27
	P09	36.44	36.39	36.32	36.25	36.17	36.08	35.97	35.86	35.74	35.61	35.47
	P10	29.94	30.03	30.07	30.06	30.00	29.88	29.72	29.50	29.23	28.91	28.54
	P11	35.71	36.53	37.12	37.50	37.66	37.60	37.33	36.83	36.12	35.19	34.05
	P12	34.94	35.70	36.34	36.84	37.20	37.44	37.54	37.51	37.35	37.06	36.63
	Mean	**36.03**	**35.29**	**34.72**	**34.32**	**34.08**	**34.01**	**34.11**	**34.37**	**34.80**	**35.39**	**36.15**

Table 14: Muscle Activation Data (%EMG)

	%Trial										
	0%	**10%**	**20%**	**30%**	**40%**	**50%**	**60%**	**70%**	**80%**	**90%**	**100%**
Rectus Femoris											
P01	22.7	23.0	23.5	24.2	25.0	25.9	26.9	28.1	29.4	30.9	32.5
P02	5.7	6.3	6.6	6.7	6.6	6.2	5.6	4.8	3.7	2.3	0.8
P03	18.9	17.5	16.3	15.3	14.6	14.1	13.8	13.8	14.0	14.4	15.0
P05	26.8	25.3	24.2	23.3	22.6	22.3	22.2	22.3	22.8	23.4	24.4
P06	15.2	13.1	11.3	9.9	8.8	8.1	7.8	7.8	8.1	8.8	9.9
P07	23.6	20.6	18.2	16.3	15.0	14.3	14.1	14.5	15.5	17.0	19.1
P08	20.9	19.6	18.5	17.5	16.7	16.1	15.6	15.2	15.1	15.1	15.2
P09	49.1	44.5	40.8	37.9	35.9	34.8	34.6	35.2	36.8	39.2	42.4
P10	32.7	28.2	24.5	21.4	19.1	17.5	16.6	16.4	16.9	18.1	20.0
P11	10.9	9.9	9.1	8.4	8.0	7.7	7.6	7.6	7.9	8.3	9.0
P12	12.6	12.3	12.1	11.9	11.8	11.7	11.6	11.6	11.7	11.8	12.0
Mean	**21.7**	**20.0**	**18.6**	**17.5**	**16.7**	**16.2**	**16.0**	**16.1**	**16.5**	**17.2**	**18.2**
Vastus Lateralis											
P01	32.4	32.1	31.9	31.9	32.1	32.4	32.9	33.5	34.3	35.2	36.2
P02	8.7	9.4	9.9	10.0	10.0	9.6	9.1	8.2	7.1	5.8	4.2
P03	25.6	25.1	24.5	24.0	23.6	23.2	22.8	22.5	22.2	22.0	21.8
P05	26.6	25.9	25.4	25.0	24.7	24.5	24.4	24.4	24.5	24.7	25.0
P06	15.9	14.1	12.5	11.2	10.2	9.4	8.9	8.7	8.7	9.0	9.5
P07	19.5	19.1	18.9	18.8	18.8	18.9	19.1	19.5	19.9	20.5	21.2
P08	21.7	20.2	19.0	18.0	17.3	16.8	16.5	16.5	16.7	17.2	17.9
P09	29.0	26.4	24.3	22.6	21.5	20.8	20.5	20.8	21.5	22.6	24.3
P10	21.4	19.6	18.1	16.9	15.9	15.1	14.6	14.3	14.2	14.4	14.9
P11	27.0	24.7	22.8	21.3	20.2	19.6	19.3	19.4	19.9	20.9	22.2
P12	19.5	20.5	21.4	22.1	22.8	23.5	24.0	24.4	24.8	25.1	25.2
Mean	**22.5**	**21.6**	**20.8**	**20.2**	**19.7**	**19.4**	**19.3**	**19.3**	**19.4**	**19.8**	**20.2**
Vastus Medialis											
P01	25.7	26.0	26.3	26.7	27.1	27.6	28.2	28.8	29.4	30.1	30.9
P02	6.9	7.7	8.3	8.6	8.6	8.3	7.8	6.9	5.8	4.4	2.6
P03	30.4	29.7	29.2	28.8	28.5	28.3	28.2	28.3	28.4	28.7	29.1
P05	31.2	30.9	30.6	30.3	30.0	29.6	29.3	28.9	28.6	28.2	27.8
P06	19.9	17.6	15.8	14.2	13.0	12.2	11.7	11.5	11.7	12.2	13.1
P07	1.9	0.5	-0.1	0.3	1.7	4.0	7.3	11.6	16.8	22.9	30.1
P08	29.0	28.1	27.1	26.2	25.3	24.4	23.5	22.7	21.8	21.0	20.2
P09	33.8	31.3	29.3	27.7	26.6	25.9	25.7	25.9	26.6	27.7	29.3
P10	-1.1	5.3	10.5	14.6	17.5	19.2	19.8	19.2	17.5	14.6	10.5
P11	26.7	23.9	21.6	19.8	18.6	17.9	17.6	18.0	18.8	20.2	22.1
P12	21.1	20.0	19.1	18.5	18.2	18.2	18.4	18.9	19.6	20.6	21.9
Mean	**20.5**	**20.1**	**19.8**	**19.6**	**19.5**	**19.6**	**19.8**	**20.1**	**20.5**	**21.0**	**21.6**

Gracilis												
	P01	12.6	13.7	15.2	17.0	19.2	21.7	24.6	27.9	31.5	35.4	39.8
	P02	3.6	4.5	5.1	5.5	5.5	5.2	4.6	3.7	2.6	1.1	-0.7
	P03	13.0	12.6	12.2	11.8	11.5	11.2	10.9	10.7	10.5	10.4	10.3
	P05	22.9	21.1	19.6	18.4	17.4	16.8	16.5	16.4	16.6	17.2	18.0
	P06	18.0	16.3	14.8	13.5	12.6	11.8	11.4	11.1	11.2	11.4	12.0
	P07	20.3	18.5	17.0	15.8	14.9	14.4	14.2	14.4	14.8	15.6	16.8
	P08	17.4	16.0	15.0	14.1	13.5	13.2	13.1	13.3	13.7	14.3	15.2
	P09	53.6	48.0	43.5	39.9	37.5	36.0	35.6	36.3	37.9	40.7	44.4
	P10	37.0	33.0	29.7	27.0	25.1	23.8	23.3	23.4	24.2	25.6	27.8
	P11	27.3	24.7	22.6	20.9	19.7	19.0	18.7	18.9	19.6	20.8	22.4
	P12	11.9	11.4	10.9	10.5	10.3	10.1	10.0	10.0	10.1	10.3	10.5
	Mean	**21.6**	**20.0**	**18.7**	**17.7**	**17.0**	**16.7**	**16.6**	**16.9**	**17.5**	**18.4**	**19.7**
Biceps Femoris												
	P01	29.2	29.9	31.0	32.4	34.1	36.2	38.7	41.4	44.6	48.1	51.9
	P02	7.2	8.2	8.9	9.3	9.5	9.3	8.9	8.2	7.2	5.9	4.3
	P03	12.9	11.9	11.0	10.3	9.6	9.1	8.8	8.5	8.4	8.4	8.5
	P05	25.4	24.0	22.8	21.8	21.1	20.5	20.2	20.1	20.2	20.6	21.1
	P06	13.4	12.7	12.0	11.3	10.8	10.3	9.8	9.4	9.1	8.8	8.6
	P07	21.3	18.9	16.9	15.3	14.1	13.2	12.7	12.6	12.9	13.6	14.6
	P08	28.8	29.3	29.7	29.9	29.9	29.8	29.5	29.1	28.6	27.8	27.0
	P09	22.8	20.1	17.8	16.0	14.6	13.6	13.1	13.0	13.3	14.1	15.3
	P10	21.6	19.6	18.0	16.6	15.5	14.6	14.0	13.6	13.5	13.6	14.1
	P11	34.5	30.4	27.1	24.5	22.6	21.5	21.1	21.4	22.5	24.3	26.9
	P12	16.9	16.5	16.1	15.9	15.7	15.7	15.7	15.9	16.1	16.4	16.8
	Mean	**21.3**	**20.1**	**19.2**	**18.5**	**17.9**	**17.6**	**17.5**	**17.6**	**17.8**	**18.3**	**19.0**
Semitendinosus												
	P01	26.0	28.8	31.5	34.1	36.5	38.9	41.1	43.2	45.2	47.0	48.8
	P02	15.1	15.5	15.7	15.7	15.6	15.3	14.7	14.0	13.1	12.1	10.8
	P03	15.6	15.0	14.5	14.1	13.8	13.6	13.4	13.3	13.3	13.4	13.6
	P05	46.8	40.7	35.5	31.3	28.0	25.7	24.4	24.0	24.6	26.1	28.6
	P06	33.4	30.8	28.6	26.7	25.2	24.0	23.3	22.8	22.8	23.1	23.8
	P07	23.7	21.5	19.8	18.5	17.7	17.3	17.4	18.0	19.0	20.4	22.3
	P08	55.6	52.5	50.0	48.0	46.7	46.0	45.9	46.3	47.4	49.1	51.3
	P09	24.0	22.6	21.4	20.3	19.3	18.4	17.6	16.9	16.4	15.9	15.6
	P10	25.0	24.6	24.3	23.9	23.6	23.3	23.0	22.7	22.4	22.1	21.9
	P11	25.0	22.4	20.2	18.6	17.4	16.7	16.5	16.8	17.6	18.9	20.6
	P12	14.7	15.1	15.5	15.6	15.7	15.6	15.4	15.0	14.6	14.0	13.2
	Mean	**27.7**	**26.3**	**25.2**	**24.3**	**23.6**	**23.2**	**23.0**	**23.0**	**23.3**	**23.8**	**24.6**

Gastrocnemius Lateral												
	P01	19.4	17.9	16.9	16.4	16.5	17.0	18.1	19.7	21.8	24.5	27.6
	P02	1.8	2.4	2.8	3.1	3.2	3.1	2.9	2.5	2.0	1.3	0.4
	P03	21.2	14.8	9.5	5.3	2.0	-0.2	-1.3	-1.4	-0.5	1.5	4.6
	P05	14.0	12.3	10.9	9.8	9.1	8.7	8.7	9.0	9.6	10.5	11.8
	P06	21.2	19.7	18.5	17.5	16.8	16.3	16.0	16.0	16.2	16.7	17.4
	P07	10.4	9.4	8.6	8.0	7.7	7.6	7.8	8.3	8.9	9.9	11.0
	P08	30.6	29.8	29.1	28.7	28.6	28.6	28.9	29.3	30.0	31.0	32.1
	P09	32.6	29.9	27.6	25.7	24.1	23.0	22.1	21.7	21.7	22.0	22.7
	P10	24.5	22.8	21.3	20.1	19.1	18.4	18.0	17.8	17.9	18.3	18.9
	P11	10.6	13.1	15.2	16.9	18.2	19.1	19.6	19.7	19.4	18.7	17.5
	P12	14.1	14.5	14.8	14.9	14.9	14.7	14.4	14.0	13.4	12.7	11.8
	Mean	**18.2**	**17.0**	**15.9**	**15.1**	**14.6**	**14.2**	**14.1**	**14.2**	**14.6**	**15.2**	**16.0**
Gastrocnemius Medial												
	P01	28.4	22.9	18.6	15.4	13.2	12.2	12.2	13.3	15.6	18.9	23.3
	P02	8.1	8.6	8.8	8.9	8.9	8.7	8.3	7.8	7.1	6.3	5.3
	P03	21.6	20.1	18.8	17.6	16.7	15.9	15.3	14.9	14.7	14.7	14.8
	P05	15.6	13.8	12.2	11.0	10.2	9.6	9.4	9.5	9.9	10.6	11.7
	P06	19.0	17.1	15.5	14.1	13.0	12.2	11.7	11.4	11.5	11.7	12.3
	P07	21.6	19.3	17.3	15.7	14.5	13.8	13.4	13.4	13.8	14.7	15.9
	P08	39.9	38.8	37.8	36.8	36.0	35.3	34.6	34.1	33.7	33.3	33.1
	P09	41.1	38.2	35.7	33.7	32.1	31.1	30.6	30.5	30.9	31.8	33.2
	P10	19.1	20.5	21.6	22.2	22.5	22.4	21.9	21.0	19.8	18.2	16.2
	P11	44.1	42.4	40.8	39.5	38.4	37.6	37.0	36.6	36.4	36.5	36.8
	P12	24.0	21.1	18.7	16.6	15.0	13.7	12.9	12.4	12.4	12.8	13.5
	Mean	**25.7**	**23.9**	**22.3**	**21.1**	**20.1**	**19.3**	**18.8**	**18.6**	**18.7**	**19.0**	**19.6**
Tibialis Anterior												
	P01	-2.5	1.5	5.5	9.5	13.5	17.5	21.5	25.5	29.5	33.5	37.5
	P02	18.2	18.6	18.7	18.6	18.1	17.3	16.3	14.9	13.3	11.4	9.1
	P03	40.7	36.6	33.2	30.2	27.9	26.1	24.9	24.2	24.1	24.6	25.6
	P05	60.9	55.6	51.0	47.3	44.3	42.0	40.6	39.9	40.0	40.9	42.5
	P06	44.0	40.1	36.8	34.1	32.1	30.7	30.0	29.9	30.5	31.7	33.5
	P07	25.2	24.7	24.3	24.1	24.0	24.0	24.2	24.5	25.0	25.6	26.3
	P08	14.8	14.6	14.5	14.4	14.4	14.5	14.7	15.0	15.3	15.7	16.3
	P09	20.9	18.9	17.3	16.0	15.2	14.6	14.4	14.6	15.2	16.1	17.4
	P10	30.6	26.2	22.5	19.5	17.1	15.3	14.2	13.8	14.0	14.9	16.4
	P11	35.1	31.3	28.0	25.2	22.9	21.1	19.9	19.1	18.9	19.2	20.0
	P12	31.8	32.0	31.9	31.8	31.5	31.1	30.5	29.8	29.0	28.0	26.9
	Mean	**29.1**	**27.3**	**25.8**	**24.6**	**23.7**	**23.1**	**22.8**	**22.8**	**23.2**	**23.8**	**24.7**

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