THE RELATIONSHIP BETWEEN THE HAMSTRINGS TO QUADRICEPS STRENGTH RATIO AND ANTERIOR CRUCIATE LIGAMENT FORCES

by

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Injury to the anterior cruciate ligament (ACL) is one of the most common knee injuries in sports and is associated with other serious health concerns like osteoarthritis (OA). Forces developed by the hamstrings and quadriceps have been found to decrease and increase forces on the ACL, respectively. Thus, some believe that the ratio of the strengths of these muscles groups (the H:Q strength ratio) is relevant when discussing ACL forces; a low H:Q ratio predisposes one to ACL injury because the weaker hamstrings cannot counteract the antagonistic quadriceps. The link between the H:Q strength ratio and ACL forces is unclear, however. The primary purpose of this study was to determine the relationship between the H:Q strength ratio and forces at the ACL during the following tasks: single and double-leg squatting, a drop jump, and walking. A secondary purpose was to determine the independent effects of the hamstrings and quadriceps on peak ACL forces during the same tasks by perturbing the maximal isometric force production of each muscle group within a subject-specific musculoskeletal model. We hypothesized that there was a relationship between the H:Q strength ratio and ACL forces. It was also hypothesized that a reduction in hamstrings strength or increase in quadriceps strength would lead to higher ACL forces, whereas an increase in hamstrings strength and a decrease in quadriceps strength would lead to decreased ACL forces. To test this, motion capture, electromyography, and ultrasound data were used to create a subject-specific model to estimate ACL forces and compare them to

each individual's H:Q strength ratio. The results indicated that the H:Q strength ratio was not related to peak ACL forces during each of the tasks together ($r = -0.12$, $p = 0.445$), or when separated by task: double leg squatting $r = -0.03$ ($p = 0.925$), single leg squatting $r = -0.52$ ($p = 0.086$), landing $r = -0.21$ ($p = 0.684$), and walking $r = 0.06$ ($p = 0.876$). Furthermore, a -10% change in hamstrings F_{max} increased peak ACL forces by 7.1% ($p < 0.001$), and a 10% increase in hamstring F_{max} decreased peak ACL forces by 5.7% ($p < 0.05$), but similar perturbations to the quadriceps strength did not have a significant effect on percent changes in peak ACL forces. When viewed in terms of bodyweights, both of the hamstrings and both of the quadriceps strength perturbations had significant effects on ACL forces, although the hamstrings still had a greater effect than the quadriceps. These results indicate that the H:Q strength ratio may not be related to peak ACL forces during submaximal tasks, but hamstring strength does appear to have an effect on modulating peak ACL forces.

The Relationship between the Hamstrings to Quadriceps Strength Ratio and Anterior Cruciate Ligament Forces

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Chapter I. Introduction

Injury to the anterior cruciate ligament (ACL) is an expensive, debilitating, and common injury that can carry significant long-term consequences. This is particularly true in young adults aged 20 to 29, in whom lesions to the ACL were found to be the most common internal knee injury and were often comorbid with medial meniscus injury (Majewski, Susanne, & Klaus, 2006). ACL injury also often requires surgery (ACL reconstruction, ACLR) (Gianotti, Marshall, Hume, & Bunt, 2009; Majewski et al., 2006). Given that approximately 50,000 ACLRs are performed yearly with an approximate cost of $17,000 per surgery, the financial burden to society is estimated to be nearly a billion dollars annually on surgeries alone (Griffin et al., 2000). More extensive cost-utility analyses estimate that the economic burden of a reconstructed ACL is $38,121 per injury, and if treated with only rehabilitation the cost increases to $88,538 per injury; annually, this equates to an annual cost of roughly $7.6 billion if the ACL is reconstructed and $17.7 billion if treated with rehabilitation (Mather et al., 2013). Rehabilitation after ACLR can take months (Beynnon et al., 2005), limiting individuals from returning to sport and completing normal daily activities. Additionally, after ACLR and return to sport, within 24 months individuals are at an increased risk for another ACL tear, in both the contralateral and ipsilateral limb (Paterno, Rauh, Schmitt, Ford, & Hewett, 2014). Furthermore, the risk of osteoarthritis (OA) has been found to increase after an ACL injury, with estimates ranging as high as half or more of all ACL injuries resulting in OA within 10 to 20 years of diagnosis (L. S. Lohmander, Englund, Dahl, & Roos, 2007). From Lohmander et al. 2007, it appears that the incidence of OA increases over time after injury as well. Similar rates of OA have been seen in both male (von Porat, Roos, & Roos, 2004) and female (L. Lohmander, Östenberg, Englund, &

Roos, 2004) soccer players who suffered from ACL injuries. Thus, ACL injury is a very serious injury that often leads to poor long-term outcomes.

Since OA and ACL injury are related, researchers have tried to determine specific risk factors associated with developing OA after ACL injury. Several have been identified, including: the severity of the ACL insufficiency, whether ACLR was performed, and the time between injury and surgery. Complete ACL tears result in a higher chance of developing OA (Kannus & Järvinen, 1989; Segawa, Omori, & Koga, 2001) compared to a partial tear (Kannus & Järvinen, 1989). ACLR patients showed higher rates of OA compared to conservatively treated ACL patients (Daniel et al., 1994; Kessler et al., 2008), and ACLR patients had even higher rates of OA in a separate study (Oiestad et al., 2010), although this was not a comparison to non-ACLR patients. Contrastingly, a meta-analysis found that the relative risk (RR) for developing any grade of OA was significantly higher in patients treated non-operatively (RR, 4.98) compared to ACLs treated with ACLR (RR, 3.62), however there was still a four-fold increased risk of developing moderate to severe OA after ACL injury regardless of treatment type (Ajuied et al., 2014). Lastly, in patients who need ACLR surgery, the longer the time before surgery is performed is associated with more degenerative changes of the articular surface of the knee (Foster, Butcher, & Turner, 2005; Jomha, Borton, Clingeleffer, & Pinczewski, 1999). Delaying ACLR surgery often necessitates meniscectomy, which appears to further increase the risk for OA (Ferretti, Conteduca, De Carli, Fontana, & Mariani, 1991; Jomha et al., 1999). Prevention of ACL injury would thus be beneficial to reduce the amount of people suffering from OA.

In order to prevent ACL injuries from occurring, the mechanism by which they occur needs to be understood. In a study utilizing questionnaires and video analysis, ACL tears were found to often occur in a noncontact mechanism (Boden, Feagin Jr, & Garrett Jr, 2000). That is, the person is not in contact with another person, but their body is in contact with the ground and is capable of producing a net force on the ACL such that it is damaged. Activities involving sudden deceleration, like landing from a jump and quick changes of direction, were the most common in noncontact ACL injuries. Furthermore, a low knee flexion angle is often associated with ACL injury (Boden et al., 2000; Yu & Garrett, 2007). While valgus moments appear to be a crucial factor in person-to-person contact ACL injuries (Boden et al., 2000), it has been suggested that they are not individually enough to cause a non-contact ACL injury (Yu & Garrett, 2007). However, a separate study found that female athletes with increased dynamic valgus and high abduction loading were at a higher risk for ACL injury (Hewett et al., 2005). Regardless, the mechanisms mentioned above appear to be the most common situations in which the ACL is torn.

Given the mechanisms of injury identified, many researchers have investigated the effects of knee muscle forces on the ACL, specifically the hamstrings and quadriceps, in order to understand how we injure and, more importantly, protect the ACL. The hamstrings reduce forces at the ACL by providing a posteriorly directed shear force onto the tibia, while the quadriceps are an antagonist of the ACL by pulling the tibia anteriorly in relation to the femur. The effects of the hamstrings and quadriceps forces are altered by the flexion angle of the knee, with the quadriceps being particularly detrimental to the ACL near full extension and the hamstrings being beneficial at almost all angles except near full extension. These findings are supported through numerous methodologies, including cadaver (DeMorat, Weinhold, Blackburn, Chudik, & Garrett, 2004; Draganich & Vahey, 1990; G. Li et al., 1999; Markolf et al., 1995; Markolf, O'Neill, Jackson, & McAllister, 2004; More et al., 1993; Pandy & Shelburne, 1997; Renstrom, Arms, Stanwyck, Johnson, & Pope, 1986; Sakane et al., 1997; Withrow, Huston, Wojtys, &

Ashton-Miller, 2008), in vivo (Beynnon et al., 1995; Henning, Lynch, & Glick, 1985; Sakane et al., 1999; Yasuda & Sasaki, 1987a; Yasuda & Sasaki, 1987b), and simulation models (Kulas, Hortobágyi, & DeVita, 2012; K. Shelburne & Pandy, 1998; Weinhandl et al., 2013; Weinhandl et al., 2014). These results are important because muscular force is a modifiable trait through training, whereas other characteristics that modify forces at the ACL (e.g., the anatomy of the knee) are more difficult or impossible to change. An assumption that seems to have been made in the literature is that muscular strength, like muscular forces, is equally able to predict both ACL loading and the risk of ACL injury.

The maximal strength of the hamstrings and quadriceps are often written as a ratio of one another: the H:Q strength ratio. Many researchers have tried to elucidate the "best" H:Q strength ratio. It has been shown that females who went on to tear their ACLs had H:Q strength ratios of ~56% compared to non ACL injured females (~60.8%) and males (~59.3%) (Myer et al., 2009). These H:Q strength ratio values were not reported to be significantly different from one another. The differences in H:Q strength ratios in the FACL group were attributed to decreased hamstring strength compared to male controls, while there were no significant differences between healthy versus injured females in either hamstring or quadriceps strength. Thus, it is unclear if lower H:Q strength ratios are predictive of ACL injury from analysis of H:Q strength ratio data alone. To date, this is the only prospective study that the current authors are aware of investigating if H:Q strength ratios are predictive of ACL injury risk. More commonly, studies investigate what happens after ACL injury. One study found higher H:Q ratios led to better functional outcomes after tearing an ACL, such as having less pain, swelling, and feelings of giving away, and increased ability to walk, climb stairs, and run (R. C. Li, Maffulli, Hsu, & Chan, 1996). Another found that after injury, the ratio of eccentric hamstrings strength to concentric quadriceps strength was similar between the injured and uninjured knees, and the ratio of concentric hamstrings to eccentric quadriceps was actually higher in the ACL deficient knee compared to the uninjured one (St Clair Gibson, Lambert, Durandt, Scales, & Noakes, 2000). Furthermore, a study found that the “optimal” H:Q strength ratio in terms of functional outcomes of the injured knee was equal to the H:Q strength ratio of the uninjured knee (less than 15% different between the knees), concluding that there may not be one specific H:Q strength ratio that is optimal when rehabilitating the ACL injured knee (Kannus, 1988). In other words, H:Q strength ratio symmetry between knees, not necessarily the magnitude of the ratio, seems to be more related to functional outcomes after tearing an ACL. However, this is not descriptive enough as the ratios of the strength could be equal, but the maximal force development could be different between the two leg’s muscles. The previous studies, aside from Myer et al., are retrospective, and it is therefore difficult to draw conclusions from these studies about the risk of low H:Q strength ratios on ACL injury risk. Despite this, it has been proposed that increasing the ratio of strength of the hamstrings to the quadriceps is desirable, to the point of saying that increasing only the hamstrings strength and not the quadriceps strength produces a favorable outcome (Tsang & DiPasquale, 2011). The clinical assumption that the H:Q strength ratio can be used as a risk factor for ACL injury appears to be unsubstantiated in the literature, as most studies focus on the effects of ACL injury on the ratio, rather than being prospective in nature. However, if the central tenet of the idea is correct, that H:Q strength ratio is related to ACL injury, then H:Q strength ratios should also be associated with ACL loading.

Purpose

The primary purpose of this study was to determine the relationship between the H:Q strength ratio and forces at the ACL during the following tasks: single and double-leg squatting, a drop jump, and walking.

A secondary purpose was to determine the independent effects the hamstrings and quadriceps strength had on peak ACL forces during the same tasks by perturbing the maximal isometric force production of each muscle group within a subject-specific musculoskeletal model.

Hypothesis

The hypothesis of the current study is that the H:Q strength ratio is a predictor of peak ACL loading during activities of daily living and common sport activities. This hypothesis is justified primarily through the research that muscular forces in the hamstrings and quadriceps modify forces on the ACL, and thus the maximal strengths of those muscle groups may also be related to ACL forces.

Furthermore, it is hypothesized that perturbations of the hamstrings and quadriceps muscle strengths within a subject-specific musculoskeletal model will cause changes in the ACL forces; specifically, a decrease in hamstrings strength or increase in quadriceps strength will increase ACL forces, and an increase in hamstrings strength or decrease in quadriceps strength will decrease ACL forces.

Significance

This study will help to determine the predictive ability of the H:Q strength ratio as a tool for assessing the forces at the ACL. Previous studies have assumed that increasing the ratio (i.e., increasing the strength of the hamstrings relative to the quadriceps) is beneficial, but this appears to be largely unsubstantiated in the literature. Thus, it is the goal of this study to conclude whether or not the ratio of the maximal strength of the hamstrings and quadriceps is predictive of peak ACL forces during the tasks aforementioned, and how perturbing each muscle group's maximal strength is related to changes in peak ACL forces.

Delimitations

1. All subjects were healthy with no self-reported previous lower extremity surgery or injury
2. The subjects were young adults (18-35 years old)
3. The subjects were recreationally active, as defined by ACSM guidelines
4. The study was delimited to analyzing only single and double-leg squatting, walking, and drop jumps
5. Optimal fiber length (FL_{opt}), a muscle force producing parameter of the subject specific model, is inherently not fully subject specific as it was calculated using resting sarcomere lengths from a cadaver study (See Ward et al., 2009, in Methods section under Subject Specific Model). There was some subject specificity, which is discussed in the Limitations section.
6. A 7° posterior tibial slope (PTS) was assumed for all individuals in the model, although this value is different for everyone (See Hudek et al., 2009, in Methods section under Estimation of ACL forces)

Limitations

1. The study utilized a musculoskeletal model that required input from numerous measures, each with their own inherent error (e.g., ultrasound, electromyography, etc.)

2. The generalizability of the results is only to healthy, young adults with no previous lower extremity injury/surgery, and not to the prevention of injury, only the forces experienced in the ACL.

Operational Definitions

Muscular strength: the net torque generated measured via dynamometry

Muscle torque: muscle force multiplied by the muscle moment arm

Strain: the relative change in shape or size of an object due to externally-applied forces

Stiffness: the resistance of a musculotendinous unit to lengthening

Optimal fiber length (FL_{opt}): the length at which a muscle fiber is capable of producing maximal force

Pennation angle (θ): the angle at which a muscle fascicle inserts into an aponeurotic tendon

Pennation angle at optimal fiber length (θ_{opt}): the angle at which a muscle fascicle inserts into an aponeurotic tendon when the muscle fibers are acting optimally

Maximal muscle force (F_{max}): the maximal amount of force generated by a given muscle

Tendon slack length (TSL): the length at which, if a tendon were stretched any more, it would begin to develop force (Delp et al., 1990)

Chapter II. Literature Review

Introduction

The primary purpose of this study was to determine the relationship between the H:Q strength ratio and forces on the ACL during the following tasks: single and double-leg squatting, a drop jump, and walking. A secondary purpose was to determine the independent effects the hamstrings and quadriceps strength had on peak ACL forces during the same tasks by perturbing the maximal isometric force production of each muscle group within a subject-specific musculoskeletal model. The rationale behind the secondary purpose was to help interpret how absolute strength (as opposed to relative strength like the H:Q ratio) is related to ACL forces. The following will be the major sections discussed in this section: the justification for ACL injury prevention research; the mechanism of ACL injury; how muscle forces at the knee affect ACL loads in various models; research on the H:Q strength ratio; and a brief summary.

Justification for ACL injury prevention research: why it is important

Epidemiology of ACL Injury and it's negative burden

Anterior cruciate ligament (ACL) injury is common in sports, and poses both immediate and chronic side effects to the injured knee. In an epidemiological study of 17,397 patients with 19,530 sports related injuries, 37% (6434) of the total patients had injuries related to the knee, with nearly 50% of internal knee injury patients being between the ages of 20 and 29, ACL lesions being the most common internal knee injury (45.4%), and medial meniscus damage often being associated with ACL injury (32.7% of multiple injured knees) (Majewski et al., 2006). A large epidemiological study over 5 years found that 65% of ACL injuries that required surgeries (3833 out of 5884 total ACL surgeries) were a result of sport and recreational activities (Gianotti et al., 2009). This means that ACL injuries are not only occurring frequently, but often in conjunction with damage to other structures as well, and often requiring surgery. An estimated 50,000 ACL reconstruction (ACLR) surgeries are performed each year with an average cost of $17,000 per ACLR, putting the cost to society at nearly a billion dollars for surgeries alone (Griffin et al., 2000). An extensive cost-utility analysis estimated that even when ACLR is performed, the economic burden is $38,121 per injury, and if treated with only rehabilitation the cost increases to $88,538 per injury; annually, this equates to roughly $7.6 billion if the ACL is reconstructed and $17.7 billion if treated with rehabilitation (Mather et al., 2013). If an individual decides to get ACLR, accelerated rehabilitation protocols can take up to 5 months, with conservative rehabilitation lasting months longer (Beynnon et al., 2005), keeping patients out of sport and limiting their functional capabilities. Lastly, ACLR is a risk factor for reinjuring the reconstructed ACL or for injuring the contralateral ACL; within 24 months after ACLR and return to sport, the incidence rate of another ACL injury was 6 times higher in ACLR patients compared to controls (Paterno et al., 2014).

ACL injury is a risk factor for developing OA

Since ACL injury is prevalent, the long-term outcomes of the injury have been investigated thoroughly. In particular, insufficiency of the ACL after injury may be a causal factor of developing osteoarthritis (OA) in the injured knee. It has been suggested that more than 50% of people who suffer an ACL injury will develop OA in the long-term (L. S. Lohmander et al., 2007). In 219 male soccer players fourteen years after an ACL tear, 78% of injured knees had radiographic changes associated with development of OA, while 41% had more advanced changes comparable with a Kellgren-Lawrence grade of 2 or more, compared to only 5% of their uninjured knees (von Porat et al., 2004). Similar values of 82% and 51% for radiographic changes and radiographic knee OA, respectively, were found in a study of female soccer players who were examined twelve years after injury (L. Lohmander et al., 2004). Because of the connection between ACL injury and OA, researchers have tried to understand what specific risk factors are involved with developing OA after ACL injury.

Several factors are important to consider with regards to development of OA in the ACL injured population. First, the severity of the ACL insufficiency may have an effect on the development of osteoarthritic symptoms. Radiographs at an average of twelve years after complete ACL injury revealed that in 89 patients, 63% had developed OA and 37% had joint space narrowing (Segawa et al., 2001), and in 77 patients who suffered from chronic partial (37) and complete (40) ACL insufficiency, 14% of partial tears and 70% of complete tears had developed OA symptoms within 8 years after ACL injury (Kannus & Järvinen, 1989). Thus, a complete tear appears to be associated with poorer long-term outcomes. Second, whether or not surgery was performed to repair the ACL may have an effect on the prevalence of OA. Even though ACLR patients had significantly better knee-stability scores compared to conservatively treated ACL patients, there was a higher rate of OA in the ACLR group (42% vs. 25%) (Kessler et al., 2008). This was further substantiated in another study that found that ACLR patients had higher levels of arthrosis after radiographs and bone scan evaluations, which they contributed to increased incidence of meniscal surgery in this group (Daniel et al., 1994). Although not in comparison to non-ACLR patients, a separate study found that 62% of ACLR patients with an isolated ACL injury developed OA after 10 to 15 years, while 82% of combined injury patients (e.g., ACL and medial meniscus) had developed OA (Oiestad et al., 2010). In contrast to these studies, a meta-analysis showed that patients treated with rehabilitation had a significantly increased relative risk (RR) of developing any grade of OA (RR, 4.98) compared to patients treated with ACLR (RR., 3.62) (Ajuied et al., 2014). Regardless of treatment type, the RR of developing minimal OA after ACL injury was 3.89, and the RR of developing moderate to severe OA was 3.84 (Ajuied et al., 2014). These studies show that even with ACLR, the long-term outcomes of an ACL injury are bleak. Lastly, the amount of time between injury and ACLR has also been established as a risk factor for OA, with increasing lengths before surgery generally leading to poorer outcomes. This is primarily due to increased degeneration of the articulating surfaces of the knee joint (Foster et al., 2005; Jomha et al., 1999). Subjects with chronically injured ACLs also required a meniscectomy significantly more often than those with acutely injured ACLs (Foster et al., 2005; Jomha et al., 1999), and having a meniscectomy alone was directly related to the development of OA (Ferretti et al., 1991; Segawa et al., 2001).

Taken together, it can be seen that ACL injury is a risk factor for developing OA, which necessitates research in both rehabilitation of ACL injuries and in prevention of such injuries. This is important because even after ACLR, the rate of OA is still high. In order to better understand factors related to ACL injury risk, a fundamental understanding of the mechanisms of ACL injury must be known.

Mechanism of ACL injury

While injury to the ACL can happen in any number of ways, research indicates that there are some shared characteristics behind many of these injuries. A study that utilized questionnaires given to athletes who tore their ACL, as well as video analysis of separate ACL tears, identified that one of the most common ways in which the ACL is injured is in a non-contact manner (Boden et al., 2000). Non-contact refers to the athlete not being in contact with another person, but their body is in contact with the ground, and through various other factors the ACL is torn. This study also found that deceleration tasks, like landing from a jump or quickly changing direction, were often being performed when the ACL was injured. ACL injury also appears to occur when the knee is at or near full extension (Boden et al., 2000; Yu & Garrett, 2007). Lastly, in cases of direct contact with another player, valgus moments appear to be a crucial factor in ACL injuries (Boden et al., 2000). It has been suggested, however, that valgus-varus and internal-external rotation moments are not individually enough to cause a non-contact ACL injury without additional forces coming from sagittal plane biomechanical factors, such as low knee flexion angle and high forces developed in the quadriceps (Yu & Garrett, 2007). Regardless, the only prospectively determined risk factor for noncontact ACL injury in females was increased knee abduction moments (Hewett et al., 2005). Thus, while the mechanism of ACL injury is multifaceted, there are some characteristics (e.g., low knee flexion angle and not in contact with another person) that appear to be common across ACL injuries.

Muscle forces at the knee affect loads on the ACL

Given the mechanisms of ACL injury, many researchers have tried to understand how forces developed in the musculature at the knee, primarily the quadriceps and hamstrings muscle groups, affect loads on the ACL. The hamstrings are protective of the ACL as they provide a posteriorly directed shear force that may reduce the loads placed on the ligament, while forces developed by the quadriceps may increase loads on the ACL by pulling the tibia anteriorly in relation to the femur. Various models have been tested, suggesting that the forces developed by the hamstrings are beneficial while the quadriceps are antagonistic with regards to protection of the ACL.

Cadaver Models

Cadaver models have been used extensively for determining how forces developed at the knee affects loads and strains on the ACL. Studies have applied quadriceps and hamstrings forces alone, as well as both quadriceps and hamstrings simultaneously.

First, the effects of the quadriceps have been studied exclusively. This muscle group acts on the tibia and produces anterior tibial forces, and thus anterior tibial translation, which has direct effects on the ACL. It has been found that a 4500N quadriceps contraction at 20° knee flexion produced significant anterior tibial translation, and thus gross ACL injury, in 6 of 11 cadaveric knees (DeMorat et al., 2004). The authors suggested that the forces developed in the quadriceps are the intrinsic mechanism responsible for noncontact ACL injury. They recognized, however, that there are various other intrinsic and extrinsic factors to consider. A separate cadaveric study applied anterior tibial loads ranging between 22N and 110N between 0^0 and 90^0 of knee flexion, and the in-situ forces in the ACL were measured (Sakane et al., 1997). ACL forces ranged from 12.8±7.3N under the 22N load at 90^0 of flexion to 110.6±14.8N under the 110N load at 15^0 of flexion, thus indicating the importance of not only the load but also the flexion angle of the knee when the load is applied (Sakane et al., 1997). Lastly, a cadaver study applied dual combinations of individual loading states on the ACL to see which combination caused the highest ACL forces (Markolf et al., 1995). The individual loading states applied were as follows: 100N of anterior tibial force, 10Nm of internal and external tibial torques, and 10Nm of varus and valgus moments. Anterior tibial force (i.e., the quadriceps force) was the most direct mechanism of ACL loading, but the highest ACL forces were generated during anterior tibial force and internal tibial torque at low flexion angles (at and near full extension). Additionally, anterior tibial force and valgus moment at flexion angles 10^0 and higher produced significantly increased ACL forces. These studies show that the quadriceps, which directly influence anterior tibial force and translation, are an important intrinsic indicator of ACL loading, particularly at low flexion angles.

Numerous other cadaver studies have involved loads applied through both the quadriceps and hamstrings both individually and simultaneously to see how these muscle groups affect ACL loads in conjunction. In one study, 100N of anteriorly directed external force was placed on the tibia, and later the researchers added 100N of quadriceps load, and then 100N of hamstrings load after that (Markolf et al., 2004). The quadriceps load increased ACL forces significantly between 10^0 and 40^0 of flexion, but the addition of 100N of hamstring loads significantly reduced ACL loads beyond 10^0 of flexion. In another study, the quadriceps were loaded with 200N of force, causing the tibia to experience increased anterior and lateral translation as well as internal rotation with respect to the femur as the knee went from full extension to 30^0 of flexion, with all these translations decreasing with further flexion (G. Li et al., 1999). Furthermore, the ACL experienced increased in-situ forces from full extension to 15^0 of flexion, decreasing thereafter. The addition of an 80N hamstring co-contraction load was able to significantly reduce the translations at all angles tested except full extension and 15^0 of flexion, as well as significantly reduce ACL forces at 15, 30, and 60^0 of knee flexion. Another study looked at the ACL strain effects of a 400N isometric quadriceps load, a 250N isometric hamstrings load, and then a simultaneous 250N hamstring load and enough load in the quadriceps to put the knee in equilibrium (Renstrom et al., 1986). When the isometric quadriceps force acted alone, it significantly increased the strain within the ACL compared to a passive normal strain between 0^0 and 45^0 of flexion. The hamstring load acting alone was able to reduce ACL strain significantly between 75 and 105^0 of flexion compared to the passive normal strain, and, although not statistically significant, reduce the ACL strain between 0 and 60^0 of flexion. When both the quadriceps and hamstrings were loaded simultaneously, the strain in the ACL was reduced significantly when compared to the isometric 400N quadriceps loading scenario between 30^0 and 90^0 knee flexion, and reduced but not statistically significant at 0^0 to 15^0 of flexion. A study using cadavers to create a sagittal-plane model of the knee showed that isolated quadriceps contractions increased forces in the ACL between 0^0 and 80^0 of flexion, but co-contraction of the hamstrings reduced ACL forces at almost all flexion angles except near full extension (Pandy & Shelburne, 1997). Again, another study found that the addition of isometric quadriceps loads of 200N significantly increased the strain on the ACL above 0% from 0^0 to 40^0 of knee flexion, but the addition of hamstrings loads to achieve equilibrium in all planes (~400N) was sufficient to significantly reduce the strain on the ACL at 10^0, 20^0, and 90^0 of knee flexion (Draganich & Vahey, 1990). Lastly, a cadaveric model of the squat exercise was developed (More et al., 1993), and it showed that increasing knee flexion angle increased the anterior tibial translation in a quadriceps stabilized knee. However, the addition of 90N of hamstring force significantly reduced anterior tibial translation, and even more so after the ACL was sectioned, providing evidence that the hamstrings may be even more important in protecting the ACL after injury to the ligament. Furthermore, internal tibial rotation was also decreased by the addition of 90N of hamstring force, becoming more noticeable at higher flexion angles. Also, the model showed that without hamstring load, the ACL (which had then been reconstructed in the experiment) experienced peak tension values at 0^0 and 30^0 of flexion, but with 90N of hamstring force the graft tension was significantly decreased, most notably between 15^0 and 45^0 of knee flexion.

To demonstrate the importance of how loads produced by the hamstrings affect ACL forces and strains, a study simulated drop-jump landing using cadavers (Withrow et al., 2008).

Adding tension to the hamstrings to mimic a lengthening muscle contraction (i.e., increase hamstring force) caused the peak relative strain in the ACL to be reduced significantly compared to both an absence of hamstring muscle tension as well as an isotonic hamstring muscle contraction. Collectively, cadaver models have repeatedly shown the antagonistic effects of the quadriceps on the ACL, particularly at low knee flexion angles and full extension, while the hamstrings act to protect the ACL, particularly at increased knee flexion angles.

In vivo models

In addition to cadaver models, in vivo studies have been performed that have shown similar effects of the quadriceps and hamstrings on the ACL.

A study of 11 patients with normal ACLs measured strain on the ACL with a Hall Effect transducer under varying conditions (Beynnon et al., 1995). A 45N weight attached to the lower leg significantly increased strain on the ACL during active extension at 10^0 and 20^0 compared to a non-weight bearing active extension movement. Furthermore, near-maximal isometric quadriceps contractions (~80% of maximal effort) increased ACL strain at 15^0 and 30^0 compared to a relaxed muscle condition, but was not significantly different at 60^0 and 90^0 of flexion. Simultaneous maximal isometric contractions of the hamstrings and quadriceps significantly increased ACL strain above a relaxed condition at 15^0, but not at 30^0, 60^0, and 90^0, and isometric hamstring contractions did not produce significantly different ACL strains at any flexion angle compared to the relaxed condition.

Another invasive in vivo study of just two patients who had a grade II sprain of the ACL compared the strain in the ACL of various exercises to the strain generated during an 80 pound Lachman test (Henning et al., 1985). They found that a quadriceps extension exercise with a 20 pound weight boot generated 87-121% of the strain that the Lachman test did between full extension and 22^0 of flexion, however at 45^0 of flexion and higher, the exercise only generated 50% as much strain in the ACL compared to the Lachman test. Other exercises requiring co-contraction of the hamstrings and higher knee flexion angles, like cycling and a single leg half squat exercise, only stressed the ACL to a small degree (7% and 21% of what the Lachman test did, respectively).

A less invasive approach to approximate ACL forces is to analyze the anterior shear forces at the tibiofemoral joint. Although not a direct measurement of ACL forces, increased anterior shear forces and anterior tibial translation are what the ACL primarily prevents, along with other ligamentous structures at knee (Sakane et al., 1999). Two studies were performed to analyze both simultaneous and separate isometric contractions of the quadriceps and hamstrings to determine the anterior/posterior drawer forces in vivo (Yasuda & Sasaki, 1987a; Yasuda & Sasaki, 1987b). When performing maximum isometric quadriceps contractions, the drawer force was found to be positive (thus indicative of ACL strain) for knee flexion angles up to $45.3^0 \pm 12.5^0$, becoming negative thereafter (Yasuda & Sasaki, 1987a). During isometric hamstring contractions, the drawer force was negative for all flexion angles tested (90^0, 75^0, 60^0, 45^0, 30^0, 15^0, and 5^0). The authors concluded that isometric contractions of the quadriceps should only be performed at 70^0 of knee flexion or more during ACL rehabilitation, while isometric hamstring contractions are safe at all angles. When performing isometric quadriceps and hamstrings contractions simultaneously, the angle at which the drawer force became negative, and was thus indicative of low or zero strain on the ACL, was at $7.4^0 \pm 5.0^0$ (Yasuda & Sasaki, 1987b). The authors concluded that simultaneous isometric contractions could therefore be performed at significantly lower degrees of knee flexion (~20^0) compared to when performing quadriceps contractions alone during ACL rehabilitation.

Simulation studies

Lastly, musculoskeletal models have been used to indirectly measure forces at the knee based off actual data that is directly measured.

One study calculated the relative displacements of the tibia, femur, and patella using a three degrees of freedom, sagittal plane model of knee during a squat (K. Shelburne & Pandy, 1998). The tibia could experience proximal-distal translation with respect to the femur, anteroposterior translation relative to the femur, and the knee was a hinge joint so it could flex and extend. The ACL was only loaded between full extension and 10^0 of flexion, after which it became unloaded; the authors attributed this to the co-contraction of the hamstrings that was necessary during the movement, thus limiting the harmful anterior pull of the quadriceps (K. Shelburne & Pandy, 1998).

When females performed a side-step cutting maneuver after a hamstring fatiguing protocol to simulate a reduction in hamstring force output, a musculoskeletal model predicted that the ACL experienced higher forces in the sagittal (44% increase in ACL forces) and frontal (24% increase in ACL forces) planes when compared to trials where the hamstring was not fatigued (Weinhandl et al., 2014). Furthermore, the model predicted a lower anterior tibiofemoral shear force in the fatigued trial, which would theoretically lower the force on the ACL, however the reduced force output of the hamstrings combined with a lower flexion angle at the time of the cutting maneuver caused higher forces on the ACL. Using the same model, the authors completed another study where the cutting maneuver was unanticipated rather than anticipated, and found that during the unanticipated trial, the ACL experienced an overall 13% higher force compared to the anticipated trial, 62% of which was because of an increase in the sagittal plane forces caused by high anterior shear forces from the quadriceps pulling the patellar tendon (Weinhandl et al., 2013).

A musculoskeletal model found that during a single leg squat, increasing forward trunk lean was able to reduce peak ACL forces by 24% and strains by 16% compared to a minimal forward lean single leg squat (Kulas et al., 2012). By leaning the trunk forward more, hamstring muscle forces were increased throughout the majority of the squat which helped to lower ACL forces and strains. The ratio of the hamstrings to quadriceps force (not strength) developed during the task was able to explain 72% of the total variance in the peak ACL forces with both moderate and minimal forward trunk lean during the single leg squat. A force ratio of ~0.4-0.5 of hamstrings to quadriceps was able to reduce the force in the ACL to 0.1 body weights, with further increases in the ratio seeing little improvement in terms of ACL force reduction. This study suggests that during activities like squatting, there may be an optimal level of force development between the hamstrings and quadriceps, which is modifiable through trunk lean, such that the ACL is loaded less.

It can be seen that dependent on the knee-flexion angle, forces developed in the quadriceps increase loads on the ACL while hamstrings forces protect the ACL. This has been supported in cadaver models, in-vivo models, and in simulated musculoskeletal models. The significance associated with these results is that muscular force, unlike other characteristics that may modify ACL loads like gross anatomy, is relatively easily modifiable. This is important because researchers can have an effect on the loads placed on the ACL by altering the contribution of the hamstring and quadriceps muscle groups during activity with an intervention, such as an exercise regimen. Because of this, many clinicians have adopted the idea of altering the maximal strength of the hamstrings and quadriceps relative to one another to better protect the ACL. There exists a clinical assumption that the maximal strength of these muscles can also be used as a means of determining the loads on the ACL.

Previous research on the H:Q strength ratio

The maximal strength of the hamstrings to quadriceps muscles, typically written as a ratio of the two (H:Q strength ratio), has therefore also been examined by researchers in an attempt to determine the “best” ratio to prevent injury. There is not yet a defined, optimal H:Q strength ratio, but many researchers have suggested that the lower the ratio the greater the risk for ACL injury. A lower ratio would imply that the hamstrings may not be able to significantly counteract the anterior pull of the stronger quadriceps.

In a prospective study that analyzed the H:Q strength ratio of females who went on to tear their ACL (FACL) compared to male and female controls who did not have an injury, the FACL group (H:Q strength ratio of ~56%) exhibited decreased relative hamstring strength but similar relative quadriceps strength compared to male controls (~59.3%), while female controls (~60.8%) exhibited reduced quadriceps strength but similar hamstring strength to male controls (Myer et al., 2009). Thus, the authors concluded that females with relatively low hamstring strength compared to quadriceps strength (i.e., lower H:Q strength ratio) were at an increased risk for injury, and that “improving” the H:Q strength ratio means to increase it, or to increase the hamstring strength relative to the quadriceps (Myer et al., 2009). This is the only prospective study that the current investigators are aware of that analyzed how the H:Q strength ratio and ACL injury risk were related.

In a study of ACL deficient knees during isokinetic strength trials performed at 180°/sec, the H:Q strength ratio extracted at 30° of knee flexion was significantly correlated with better scores on a functional ability test (R. C. Li et al., 1996). All characteristics of the hamstrings tested (peak torque, endurance ratio, total work output, and explosive power) were also correlated with improved functional scores, but none of those same characteristics of the quadriceps were correlated with increased scores. Examples of what was scored included pain, swelling, feelings of giving away, and the ability to walk, run, and climb stairs. Thus, while the authors concluded that increasing the H:Q strength ratio was correlated with functional improvements, it appears that these correlations are driven by hamstring performance.

Despite little evidence to suggest there is an optimal H:Q strength ratio, training regimens have been created around the idea of "improving" the ratio. Again, this is thought to be achievable by increasing the strength of the hamstrings relative to the quadriceps so that the anterior pull of the quadriceps is not as antagonistic to the ACL. Specifically, a plyometric training study with women participants increased the hamstrings average power development at 120^0/sec and hamstring strength in the second through fourth weeks of the program, but maintained the strength of the quadriceps near the baseline level (Tsang & DiPasquale, 2011). Thus, the authors concluded that they improved the H:Q strength ratio (Tsang & DiPasquale, 2011). Focusing on only improving the hamstrings strength may lead to functional deficits in the quadriceps however, which may be more detrimental after ACLR (Lewek, Rudolph, Axe, & Snyder-Mackler, 2002). In this study, those with weaker quadriceps strength (<80% compared to their uninvolved leg) after ACLR had altered gait patterns compared to ACLR subjects with stronger quadriceps (>90% compared to their uninvolved leg). ACLR subjects with weaker quadriceps had lower knee flexion angles and internal knee extensor moments than did uninjured subjects, similar to that of ACL deficient individuals, whereas the group with stronger quadriceps did not exhibit such deficits. Thus, although it seems beneficial to increase the strength of the hamstrings, strictly focusing on that muscle group without regard to the strength of the quadriceps is not appropriate post-ACLR.

A number of studies have focused on what happens to the H:Q strength ratio after injury to the ACL, and some have tried to draw conclusions from that data about the risk for ACL injury. After suffering from an ACL injury, one study found that the strength of the hamstrings and quadriceps were reduced significantly both concentrically and eccentrically, with the greatest deficits being seen in the eccentric peak torque of the quadriceps (decrease of 38%), followed by a 16% reduction in concentric peak torque of the quadriceps, 15% reduction in eccentric peak torque of the hamstrings, and 8% reduction in concentric peak torque of the hamstrings (St Clair Gibson et al., 2000). The ratio of eccentric hamstring peak torque to concentric quadriceps peak torque was similar between the ACL deficient and uninvolved limbs, but the ratio of concentric hamstrings peak torque to eccentric quadriceps peak torque was significantly higher in the ACL deficient compared to the uninvolved limb. Lastly, however, the study found that there were no significant differences between uninvolved and ACL deficient knees in concentric hamstring to concentric quadriceps peak torque, but the eccentric hamstring to eccentric quadriceps peak torque ratio was significantly higher in the ACL deficient knee. From these results it appears that the differences in ratios were highly dependent on contraction type (concentric vs. eccentric). This study only focused on what happens after an ACL injury, but not how the H:Q strength ratio prospectively relates to ACL injury. A study found that the optimal H:Q strength ratio of an ACL injured knee, as defined by better scores on a functional measure (Lysolm Knee-Scoring Scale), a roentgenogram, and a scale developed by another study (Marshall, Fetto, & Botero, 1977), was simply the absolute ratio for the uninjured knee, not an improved or higher ratio (Kannus, 1988). Patients with less than a 15% difference in their H:Q strength ratio of the ACL injured knee and the uninvolved knee had better outcomes in the above measures compared to those with greater than 15% difference in their H:Q ratio between knees (Kannus, 1988). Additionally, the study found that H:Q strength ratio was highly variable among subjects' injured and uninjured knees when tested at both 60°/sec and 180°/sec (H:Q ratios ranging from 0.23 in an isometric test to 1.80 in a total work test in the uninjured knee, and 0.23 in an isometric test and 2.05 in a total work test in the injured knee, with other tests in the study showing similarly high variation). The authors concluded that there may not be a specific strength ratio goal for rehabilitation, and that absolute H:Q strength ratios do not correlate with long-term outcomes.

Summary

Injury to the ACL is common and it poses long-term health consequences, most notably an increased risk for OA. Researchers have studied the mechanisms behind ACL injury, of which a few have been identified, including low knee flexion angles coupled with a non-contact movement. There is a plethora of research on how forces produced by the muscles at the knee relate to ACL forces: studies have repeatedly shown that the quadriceps develop forces that cause increased loads and strains on the ACL particularly at low flexion angles, while the hamstring muscles produce forces that reduce such forces and strains. The ability of the H:Q strength ratio to predict forces at the ACL, and how it relates to prevention of ACL injury, however, is not very well understood. Thus, the clinical relevance of the H:Q strength ratio as it relates to risk for, and prevention of, injury must be substantiated, as there is not yet a definitive way to improve it. If there is a relationship, however, then H:Q strength ratios should inherently be related to the forces placed on the ACL. The primary purpose of the study was to determine the relationship between the H:Q strength ratio and forces at the ACL during the following tasks: single and double-leg squatting, landing from a jump, and walking. A secondary purpose was to determine the independent effects of the hamstring and quadriceps on ACL forces during the squatting, landing and walking tasks. The hypothesis of the current study is that the H:Q strength ratio is associated with peak ACL forces during activities of daily living and common sport activities. It was also hypothesized that an increase in hamstrings strength or decrease in quadriceps strength would result in lower ACL forces, and a decrease in hamstrings strength or increase in quadriceps strength would increase ACL forces.

Chapter III. Methods

Design

The aim of this study was to determine how the H:Q strength ratio was related to forces in the ACL during single and double leg squatting, walking, and during a drop jump, as well as to determine the effects of perturbing the hamstrings and quadriceps maximal isometric force producing properties on ACL forces. We hypothesized that the H:Q strength ratio would be related to ACL forces during a single and double-leg squat, walking, and drop jump. Furthermore, we hypothesized that a reduction in hamstrings strength or increase in quadriceps strength would increase ACL forces, whereas an increase in hamstrings strength or decrease in quadriceps strength would decrease ACL forces. Subjects reported to the lab to have ultrasound images taken of their hamstrings and quadriceps on one day, then came back on a second day to complete dynamometer and motion capture protocols while electromyography data was collected. A musculoskeletal model that imbeds subject-specific quadriceps and hamstrings muscle architecture and strength, as well as muscle activations, was utilized to estimate muscle and joint reaction forces which were used to estimate subject specific ACL forces.

Subjects

Six young adults between the ages of 18 and 35 years old were recruited for this study (3 males: height = 1.79 (.06) m, mass = 74.7 (10.5) kg, age = 22.7 (3.2) yrs.; 3 females: height = 1.63 (.02) m, mass = 56.0 (0.6) kg, age = 21.0 (1.0) yrs.). Subjects were recreationally active, as defined by ACSM's standards for physically active adults. Subjects had no self-reported previous lower extremity surgery, pain, or muscle strain. All subjects read and signed a document of informed consent that was approved by the University Internal Review Board

(Appendix A). Criteria for inclusion included: no previous lower extremity injury or pain, recreationally active, and comfortable when performing a squat and drop landing.

Procedure

This study was conducted in the Biomechanics Lab, Ward Sports Medicine Building on East Carolina University campus. Subjects reported to the lab for data collection on two separate days. The first day consisted of anthropometric and ultrasound data collection and the second day consisted of a dynamometer protocol and a motion capture protocol while electromyography data was collected during both.

Anthropometrics

Each subject's height and weight were recorded using a Seca 703 digital scale (Seca GmbH & Co. KG, Hamburg, Germany).

Ultrasound Imaging

Ultrasound images were taken of the quadriceps, hamstrings, and gastrocnemii muscle groups of the right leg using an ultrasound unit (SuperSonic Imagine, Aixplorer, Bothell, WA). Aquasonic Ultrasound Gel (Parker Laboratories, Aquasonic 100, Fairfield, NJ) was used to enhance the image quality as well as provide a lubricant during measurement. The images were taken with the subject laying prone for the hamstrings and gastrocnemii and supine for the quadriceps on a standard treatment table (hip and knee flexion at 0°).

Multiple cross-sectional area (CSA) images of the vastus lateralis (VL) and biceps femoris long head (BFLH) were taken, as well as images of fascicle lengths and pennation angles of the VL, vastus intermedius (VI), vastus medialis (VM), rectus femoris (RF), BFLH, biceps femoris short head (BFSH), semitendinosus (ST), semimembranosus (SM), medial gastrocnemius (MG), and lateral gastrocnemius (LG). To take CSA measurements, the VL was divided into eleven equidistant points from the greater trochanter of the femur to where it inserts at the patellar tendon. For the BFLH, the muscle was divided into eleven equidistant points from the gluteal fold to where it inserts at the fibula. The distances between these landmarks in both the VL and BFLH were recorded and used to estimate volume, explained later. The ultrasound probe was held perpendicular to the BFLH and VL's long axes at each of the eleven points to capture the CSA (Figure 1A and 1C, respectively). To capture an image of the muscle fascicle length of each of the quadriceps, hamstrings, and gastrocnemii muscles, the probe was held parallel to each of the muscle's long axes and was run along the axis using a panoramic feature of the ultrasound unit (Figure 1B and D). A muscle thickness measurement of the gastrocnemii was taken at 30% of the distance from the articular cleft of the femur and tibia condyles to the lateral malleolus (Miyatani, Kanehisa, Ito, Kawakami, & Fukunaga, 2004). Every measurement had two separate images taken so that averages could be calculated for CSAs, fascicle lengths, and pennation angles.

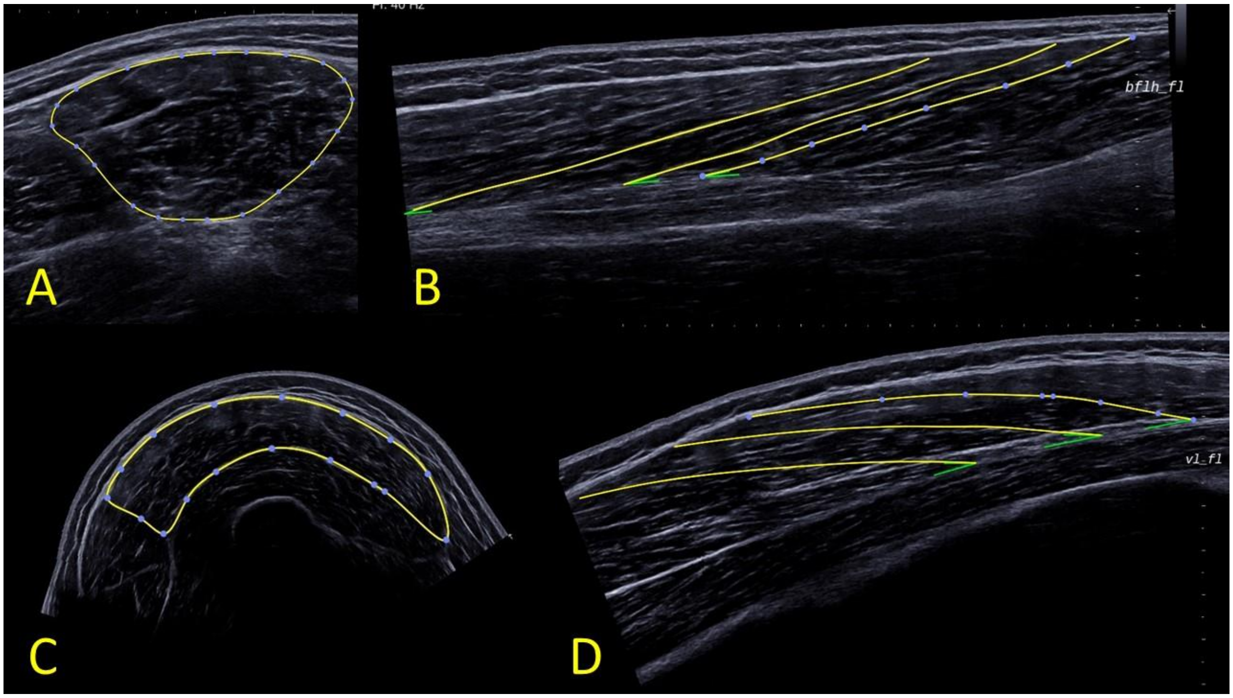

FIGURE 1. A) CSA measurement of the BFLH. B) Muscle fascicle lengths and pennation angles of the BFLH. C) CSA measurement of the VL. D) Muscle fascicle lengths and pennation angles of the VL.

Electromyography

A sixteen-channel wireless electromyography (EMG) system (Delsys Trigno^{TM} Wireless Systems, Delsys®, Natick, MA) collected EMG data of the VL, RF, VM, medial hamstrings, lateral hamstrings, MG, LG during the dynamometer and motion capture protocols explained below. Each subject's thighs and shanks were prepared using standard EMG protocols, including: shaving any hair over the muscle belly, using lemon prep abrasive skin prepping lotion to abrade the skin and lower skin impedance, and cleaning the area with alcohol prep pads.

Passive and Peak Torque Measurement

A HUMAC NORM Dynamometer (CSMI, model 502140, Stoughton, MA) was utilized to test both passive and maximal torque of both the hamstrings and quadriceps. Subject's hips were flexed to 90º, and the dynamometer was adjusted to their height and specific needs, such that the lateral epicondyle of the knee was lined up with the axis of rotation of the dynamometer. The protocol included: five repetitions of consecutive passive torque measurement where the dynamometer moved the knee through a 0º-100º range of motion at a rate of 5º/sec, five back-to-back repetitions of maximal concentric knee extension-flexion contractions at 60º/sec through the same 100º range of motion, three repetitions each of maximal isometric torque for the quadriceps at 60º, 80º, and 100º knee flexion, and three repetitions each of maximal isometric torque for the hamstrings at 30º, 45º, and 60º knee flexion.

Motion Capture Protocol

Eight Qualisys ProReflex MCU 240 cameras (Qualisys Medical AB, Gothenburg, Sweden) were used to capture each subject's motion over top of either a large AMTI force plate (AMTI Model BP6001200-2K, Watertown, MA) or a small AMTI force plate (AMTI Model OR6-6-2000, Watertown, MA), or both (e.g., during double-leg squatting and drop jump landing trials). Each subject performed five repetitions of each of the following dynamic trials: double-leg squatting, single-leg squatting on the right foot, single-leg squatting on the left foot, walking at a self-selected pace, and landing from a jump off a 30 cm box followed by a quick rebound jump for maximal height. For the static motion capture trial, reflective markers were placed on the top of the head, lateral epicondyles of the humeri, styloid processes of the radii, anterior superior iliac spines (ASIS), posterior superior iliac spines (PSIS), top of the iliac crests, greater trochanters of the femur, medial and lateral knee at the tibiofemoral joint, at the ankle on the lateral and medial malleoli, first and fifth metatarsal heads, heels, lateral rear of feet, medial side of both feet, and triad marker plates were placed on the subjects' thighs and shanks. During both static and dynamic trials, a vest was worn with reflective markers placed over the shoulders, sternum, and upper back region. For dynamic trials, the following markers were removed: top of the head, lateral epicondyles of the humeri, styloid processes of the radii, iliac crests, ankle lateral and medial malleoli, first metatarsal heads, and medial sides of the feet.

Data Reduction

Ultrasound Reduction

The CSA of the VL and BFLH, as well as fascicle lengths and pennation angles of all the quadriceps, hamstrings, and gastrocnemii were analyzed on OsiriX DICOM Viewer software (Pixmeo, Bernex, Switzerland), see Figure 1 above. Three measurements of fascicle length and pennation angle were taken in each longitudinal image (distal, middle, and proximal of each image) and averaged for analysis. Two cross-sectional images at each of the eleven equidistant points were taken of the VL and BFLH, and the CSA of each muscle was averaged for each section. This average CSA value was used to calculate muscle volume for both these muscles. Volumes were calculated by integrating the CSA versus muscle length curve, with the distance between the CSA scans being used in that integration calculation.

EMG Reduction

EMG data was high-pass filtered at 30Hz, full-wave rectified, low-pass filtered at 6 Hz, then normalized to the peak EMG signal for each muscle. Peak EMG signal could occur during either the HUMAC protocol or a motion capture trial, whichever yielded the maximum EMG activation.

Dynamometer Reduction

The strength data from the dynamometer was processed offline, and gravity corrections were made on the 60º/sec isokinetic trials. This was accomplished by using the length of the shank and the subject's mass to determine the mass of the lower leg and foot, the location of the center of mass of the lower leg, and therefore the torque generated by the weight of the lower leg at all knee angles (Kellis & Baltzopoulos, 1996). Then, each isokinetic trial was normalized to knee angle to determine the torque generated at each specific degree in either flexion or extension. The peak torque developed by both the hamstrings and quadriceps during the isokinetic trials, the isometric torque at each of the angles listed above for the quadriceps and hamstrings, and the passive torque were all determined. The H:Q strength ratio was calculated using the average of the three peak torques that occurred during second, third, and fourth flexion and extension torques generated during the isokinetic dynamometer trials (i.e., middle three repetitions of the five repetition set).

Kinematic and Kinetic Reduction

Qualisys Track Manager (QTM, Innovision Systems, Columbiaville, MI) was used to label each individual marker in every subject's trials, which was exported to Visual 3D software (V3D, C-Motion Inc., Rockville, MD) to build a model of each individual. Kinetic data were low-pass filtered using a 4^{th} order Butterworth filter with a cutoff frequency of 50Hz, and kinematic data were low-pass filtered using a 4^{th} order Butterworth filter with a cutoff frequency of 6Hz.

Creating the Subject Specific Model

Ultrasound and static motion capture data were collectively used to derive a subject-specific musculoskeletal model in Software for Interactive Musculoskeletal Modeling (SIMM, Musculographics Inc., Santa Rosa, CA). The model used in SIMM has 25 total degrees-of-freedom (DOFs), including three rotations about the lower torso, and for both right and left arms and legs: arm adduction, arm rotation, arm flexion, elbow flexion, pronation/supination of the forearm, hip adduction, hip rotation, hip flexion, knee flexion, ankle dorsiflexion/plantarflexion, and subtalar inversion/eversion.

First, a generic model that came with the distribution with SIMM (Delp et al., 2007) was scaled to match the anthropometrics of each subject. This was accomplished via static calibration using the motion capture system, and by setting the mass and height of the model equal to the individual. If there were visible errors in the wrapping objects of the model (e.g., a muscle's musculotendinous length changed drastically throughout the range of motion in a non-physiological manner), the wrapping objects were slightly modified so that they did not cause unrealistic changes in musculotendon lengths. Unrealistic changes included sudden, drastic changes in the musculotendinous lengths that were caused by the muscles interacting improperly with the wrapping objects built into SIMM. This was corrected with small (order of millimeters) changes in the positioning of the wrapping objects. These errors were noticeable whenever a muscle improperly moved over the bones or wrapping objects in a joint, such as the insertion of the BFLH briefly snapping to a non-physiological place behind the knee before returning to normal. Improper wrapping generally occurred due to non-uniform scaling of the bone segments relative to the generic model. For example, if the generic model was scaled up to match a larger individual, the pelvis may have been scaled up by a factor of 10% but the femur only by 2%. Thus, the wrapping objects associated with the pelvis that interacted with the femur would likely act non-physiologically.

The subject specific model also included various muscle force producing parameters, which were later used in the calculation of muscle forces using a Hill-type muscle model. Fascicle length and pennation angle were estimated based on ultrasound data for the aforementioned muscles, and muscle volume of the VL and BFLH were estimated based on the

CSA and muscle length measurements. From these variables, it was possible to determine optimal muscle fiber length (FL_{opt}), pennation angle at optimal fiber length (θ_{opt}), maximal isometric force (F_{max}), and tendon slack length (TSL), which were used to make the model subject specific in terms of muscle force producing parameters (Equations 1-4 below).

Since the volumes of only the VL and BFLH were estimated using the slice-CSA integration method outlined above, the volumes of the other quadriceps and hamstrings were estimated based off their relative contributions towards overall muscle group volume. To accomplish this, several studies were pooled together to derive the relative contribution of each muscle's volume to the overall volume of the muscle group (see Appendix B). For example, the VL comprised 33% of the total quadriceps muscle volume, the VI 28%, the VM 25%, and the RF 14% when averaged across several studies. Thus, after estimating the VL volume through the slice-CSA method, the rest of the muscle group's volumes could be determined by each muscle's relative contribution to the whole. This was repeated for the hamstring group after estimating the BFLH volume using the slice-CSA method. The BFLH comprised 27% of the total hamstring volume, the BFSH 14%, the ST 26%, and the SM 33% when averaged across several studies. The MG and LG muscles' volumes were estimated using regression equations that are based on leg length and the thickness of the calf muscle from the superficial muscle to the tibia, again, expressed as a relative contribution to the whole plantar flexor muscle volume (Miyatani et al., 2004). The soleus comprised 54.09% of the total plantar flexor volume, the LG 12.20%, the MG 22.26%, and the tibialis posterior 11.45%.

(1) $FL_{opt} = FL_{raw} (S_{opt} / S_{rest})$

(2) $\theta_{opt} = sin^{-1} ((FL_{raw} * sin(\theta_{raw}/57.3)) / (FL_{opt}))*57.3$

(3) $F_{max} = (Volume / (FL_{opt} * cos(\theta_{opt}))) * Specific\ Tension$

$(4)\ TSL = MT_{length} - (FL_{opt} * \sqrt{(FL_{norm}^2 - \sin^2(\theta_{opt}))})$

EQUATIONS 1-4: Muscle force producing parameter equations.

First, optimal fiber length (FL_{opt}, equation 1) was determined by multiplying the average of six muscle fascicle lengths (three from each of the two images taken of each muscle) by a ratio of the optimal sarcomere length (S_{opt}) of 2.7µm (Lieber, Loren, & Friden, 1994; Walker & Schrodt, 1974) to resting sarcomere lengths (S_{rest}) for each individual muscle determined by a cadaveric study (Ward, Eng, Smallwood, & Lieber, 2009). FL_{opt} is the length of the muscle fiber when it is at its optimal length for producing muscle force (i.e., can produce its maximal amount of force).

Second, pennation angle at optimal fiber length (θ_{opt}) is shown in equation 2 (Garner & Pandy, 2003). This relationship provided the pennation angle at which the muscle fibers were acting when the fiber was at an optimal length. In this calculation, it is assumed the muscle's thickness remains constant when it contracts. The value of θ_{raw} is divided by 57.3 to convert to radians, but the entire value calculated within the $\sin^{-1}$ function is multiplied by 57.3 to convert to degrees overall.

Third, F_{max} was determined by multiplying the physiological cross-sectional area (PCSA) by the specific tension of human muscle. Specific tension was estimated to be $35 N/cm^2$ for the purposes of this study (Erskine, Jones, Williams, Stewart, & Degens, 2010). PCSA is calculated by dividing the volume of the muscles by the FL_{opt} corrected for θ_{opt}.

Tendon slack length (TSL, equation 4) is the last muscle force producing parameter estimated for each subject-specific model. To calculate TSL, the following variables were needed: musculotendinous length (MT_{length}), FL_{opt}, θ_{opt}, and normalized fiber length (FL_{norm}, a value between 0.5 and 1.5, or within physiologically operable ranges). The equation for TSL comes from a generic form of equations 3 and 4 from Garner & Pandy, 2003, rearranged to solve for TSL instead of MT_{length}. TSL was calculated by estimating the MT_{length} of the hamstrings, quadriceps, and gastrocnemii with the hip and ankle in anatomical zero position in SIMM, similar to the position each subject was measured in when undergoing the ultrasound procedure. The calculation of TSL allowed for a variable pennation angle. This method works well for all muscles except for the ST, SM, and BFLH. For these muscles, when the hip is at 90º and knee is at 0º of flexion (i.e., fully extended), FL_{norm} (which was calculated within SIMM) can exceed physiological values, ranging from 1.5-2.0 (Arnold & Delp, 2011). Thus, an adjustment of TSL for these muscles was performed using the MT_{length} and a peak FL_{norm} of 1.5 with the hip at 90º and knee at 0º. Once the TSL's for all the muscles were calculated, the final muscle operating ranges were checked to ensure the normalized operating ranges for each muscle were within 0.5 and 1.5. We visually checked this in SIMM by plotting the FL_{norm} of each muscle and confirmed that all muscles calculated FL_{norm} were within physiologically reasonable ranges.

Lastly, the model incorporated muscle activations, obtained from EMG measurements during the motion capture trials. Some of the muscles' EMG were modeled according to other muscle's activations; the activation of the VI was equal to the average of the VL and VM, the ST was equal to the SM, and the BFSH was equal to the BFLH (Lloyd & Besier, 2003).

Estimation of Muscle Forces

Muscle forces were estimated using a hybrid static optimization procedure. The motions of the model segments were prescribed according to the kinematic experimental data. The hybrid static optimization procedure was then applied in SIMM, which optimized the experimental and modeled EMG signals such that the net knee muscle moments matched the inverse-dynamics based knee moments. In the presence of EMG, the hybrid optimization procedure attempts to preserve the experimental EMG as much as possible, rather than utilize a sum of muscle activations squared objective function which is used as the objective function in the absence of EMG. This static optimization procedure generally maintained the overall shape of the curve of muscle activations over time, although it did change the magnitude. The range of r values comparing the optimized and experimental EMG for the VL, BFLH, and LG for each task are as follows: VL r = 0.67 to 0.99, BFLH r = 0.61 to 0.76, and LG r = 0.21 to 0.95. While the VL had the overall highest correlations, the BFLH and LG still had moderate-high correlations and the patterns appear similar overall (see Appendix C), thus we felt confident using these optimized EMG data. Using this procedure, muscle forces were estimated for each muscle at each point in time using a Hill type muscle model.

The Hill type muscle model assumed that the force being produced in the musculotendinous unit (F^{mt}) was equal to the force being produced in the tendon (F^t, i.e., all force was transferred through each muscle's tendon). F^t was equal to the F_{max} of the muscle as a function of generic force-length (f(l)), force-velocity (f(v)), and parallel passive elastic force-length (f_p(l)) curves built into SIMM, modified by the current activation state of the muscle (a(t), which was the result of the optimized EMG data being fed through a first order differential equation) and the pennation angle at optimal fiber length (ϕ(t)) (Lloyd & Besier, 2003). In general form, the equation is as follows (Lloyd & Besier, 2003):

$$(5)\ F^{mt}(t) = F^t = F^{max}[f(l)f(v)a(t) + f_p(l)]\cos(\phi(t))$$

EQUATION 5: Estimation of muscle force equation.

Estimation of ACL Forces

Muscle forces estimated during the dynamic trials through the subject specific model were combined with the joint reaction forces at the knee to estimate ACL forces (Kulas et al., 2012) (Figure 2).

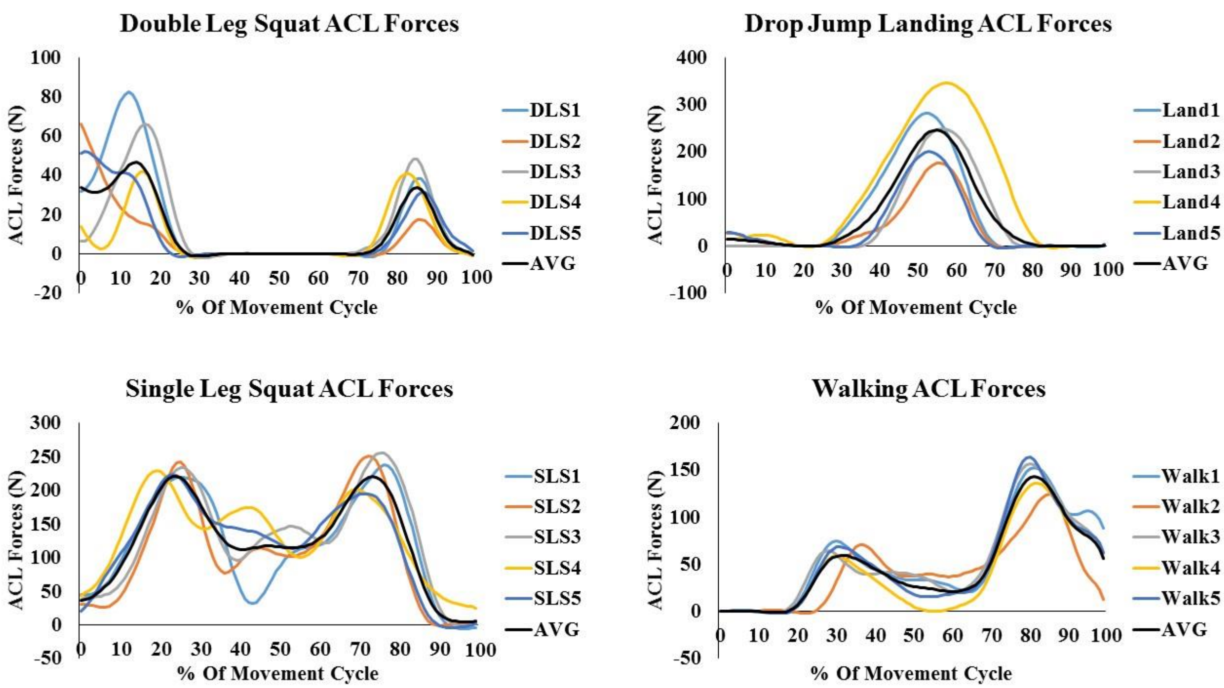

FIGURE 2. Estimation of ACL forces during five trials (and average) of double leg squatting, single leg squatting, landing, and walking from one subject's data.

To estimate ACL forces, the component quadriceps muscle forces were summed (i.e, VL, VI, VM, RF added together) and translated into patellar tendon force based on the quadriceps/patellar tendon relationship according to knee flexion angle (Van Eijden, Kouwenhoven, Verburg, & Weijs, 1986). Each of the hamstrings muscle forces were also summed, and their orientations relative to the tibia were expressed as polynomial functions to then estimate the shear components of each of these muscles. The same was done for the gastrocnemii muscles using a linear function. The orientations of the hamstrings and gastrocnemii muscles relative to the tibia were pulled directly from SIMM. Once all muscle and joint reaction forces were resolved into both compression and shear components relative to the surface of the tibia, a 7° posterior tibial slope (PTS) was taken into account (Hudek, Schmutz, Regenfelder, Fuchs, & Koch, 2009). It is understood that PTS varies between people and it also influences ACL forces (Marouane, Shirazi-Adl, Adouni, & Hashemi, 2014; K. B. Shelburne, Kim, Sterett, & Pandy, 2011), so a sensitivity analysis was performed (See Appendix C). Because of the PTS, a small portion of the muscle induced compressive forces also added to the total shear forces along the tibia. Anterior shear forces (ASFs, forces parallel to the surface of the tibia acting anteriorly) were assumed to be acting primarily on the ACL with negligible contributions from the posterior capsule, menisci, and other ligaments. Lastly, the summed total shear forces acting along the surface of the tibia were adjusted to account for the ACL elevation angle, which varies across knee flexion angle during weight bearing flexion (Jordan et al., 2007). A 10Hz Butterworth filter was then applied to the ACL forces.

Only peak ACL forces were used for comparison. Two peaks were taken from each double leg and single leg squat trials, one during the descent phase and one during the ascent phase of the squat. Squat trials started before noticeable descent had occurred and ended shortly after returning to full, upright stance. Two peaks were taken during each walking trial as well, between ground contact and toe-off (one in the first 50% of time spent in contact with the ground and one in the last 50% of time spent in contact with the ground). One peak was taken from each landing trial. Landing trials began 50 milliseconds prior to ground contact and lasted until 150 milliseconds after ground contact.

The graphs in Figure 2 illustrate ACL forces across the four tasks being performed in the study for one subject. The ACL forces, either in magnitude, pattern, or both, found in our study compare well to others found in the literature. During double-leg squatting trials in the present study, peak ACL forces generally occurred in the descent phase of the squat and near full knee extension, with slightly lower ACL peaks during the ascent phase, which corresponds well with a study investigating ACL strain during squatting (Beynnon et al., 1997). For single-leg squatting, ACL forces in the current study had peaks during both the descent and ascent phases and generally occurred at knee angles near full extension, which matches well with another study examining one-leg squats (R. Escamilla et al., 2009), although the magnitudes in the present study were higher. ACL forces during walking were bimodal with the second peak (i.e., near toe off) often being a little larger, whereas the peaks occurred during contralateral toe-off (i.e., shortly after ground contact) in a separate study (K. B. Shelburne, Pandy, Anderson, & Torry, 2004), but the overall magnitude of ACL forces were similar (less than ½ BW). In both the present study and in Shelburne et al., 2004, ACL forces increased in the early stages of walking, decreased during mid-stance, and increased again during toe-off, so the pattern of ACL forces throughout the gait cycle was similar. During drop-jump landings, peak ACL forces occurred around peak vertical ground reaction forces, which is similar to another study investigating drop-jumps from a 60cm platform (Pflum, Shelburne, Torry, Decker, & Pandy, 2004), and the peak ACL force in Pflum 2004 was similar to the present study (~.4 BW vs ~.56 BW). Similarly, in a single-leg hop task, peak ACL strain occurred at peak ground reaction force as well (Cerulli, Benoit, Lamontagne, Caraffa, & Liti, 2003). Because our peak ACL forces and the curves of ACL forces throughout the motions match well with other studies, we have reason to believe the model is producing reasonable ACL forces.

Data Analysis

All analyses were performed on the right leg, thus all single leg squats on the left leg were left out of the analysis.

The primary purpose of this study was to test the hypothesis that the H:Q strength ratio is a predictor of peak ACL loading during activities of daily living or common sport activities. A correlation analysis was performed to determine the relatedness between these two variables during double leg squatting, single leg squatting, walking, and landing from a jump. This analysis was then further separated by task to determine if there were relationships between H:Q strength ratio and peak ACL forces during a specific task.

A secondary purpose was to determine the independent effects the hamstrings and quadriceps strength had on peak ACL forces during the same tasks by perturbing the maximal isometric force production of each muscle group within a subject-specific musculoskeletal model. The F_{max} of the hamstrings and quadriceps muscle groups were perturbed by +10% or -10% within SIMM, one muscle group and one perturbation at a time, to simulate either an increase or decrease in strength of that muscle group, respectively. This effectively created two different H:Q strength ratios for each muscle group, one decreased and one increased from the original ratio by 10%, so four new H:Q strength ratios per subject. Once each muscle's F_{max} within the muscle group was perturbed, static optimization was performed on each perturbed condition for each dynamic trial per subject (6 subjects, 4 new H:Q ratios, 20 trials, so 480 simulations). This effectively changed the muscle forces during each of the trials while retaining the same joint moment conditions, so peak ACL forces were estimated and normalized again for each subject.

Statistical significance was set at $P < 0.05$ for all analyses.

Chapter IV. Results

Primary Purpose – H:Q strength ratio vs. ACL forces across all tasks

Average joint angle and joint torque data (with standard deviation lines) for each of the tasks across the right hip, knee, and ankle for all subjects are shown below in Figures 3 and 4, respectively. For kinematic and kinetic analyses, landing trials are shown for the entire trial, whereas ACL and muscle forces were analyzed from 50 ms prior to ground contact to 150 ms post ground contact.

The overall pattern of the joint angles and joint torques match well with other studies. For double leg squats, one study had male subjects with a mean mass of 93 kg performing squats at their 12-repetition maximum, and had a similar knee ROM and extension torque pattern as in the present study, although the peak extension torque in that study was higher at 175 Nm, or roughly 1.9 Nm/kg compared to the present study value of ~0.6 Nm/kg, since they were lifting weights (R. F. Escamilla et al., 1998). Powerlifters also completed the double leg squat movement with similar hip and knee ROM's, although it is difficult to compare the torques at the different joints because they were lifting close to their 1 repetition maximum. For single leg squats, subjects performed the movement through roughly 62° and 66°of peak knee flexion for men and women, respectively, which is similar to the present study's average peak knee flexion angle of ~69° (Claiborne, Armstrong, Gandhi, & Pincivero, 2006). For comparisons to drop jump landing, two studies showed roughly similar hip, knee, and ankle angles at contact position and total ROM for those joints (Decker, Torry, Wyland, Sterett, & Steadman, 2003; Fowler & Lees, 1998). In comparison to studies that investigated walking, the hip, knee, and ankle joint positions and joint torques matched well with others (DeVita & Hortobágyi, 2003; Kadaba et al., 1989), although it's important to note that Kadaba et al. analyzed the full gait cycle while in the present study we only analyzed the time in contact with the ground.

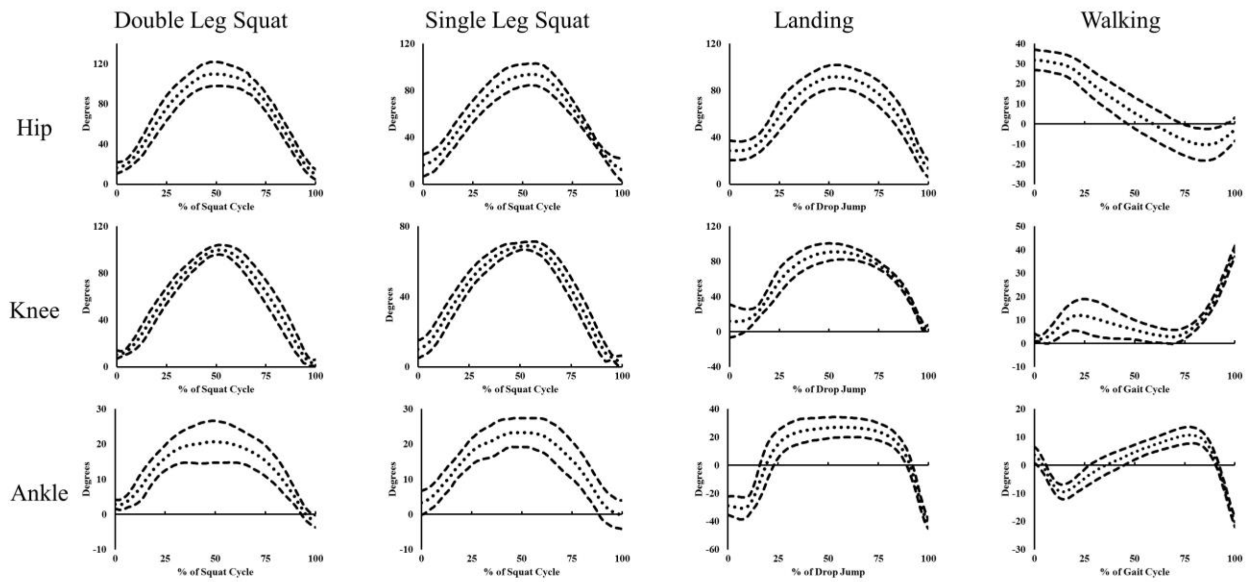

FIGURE 3. The average right hip, knee, and ankle joint angles are shown above for all subjects across the four tasks analyzed. Dashed lines represent +/- one standard deviation from the average (dotted lines). Positive angles represent hip flexion, knee flexion, and ankle dorsiflexion.

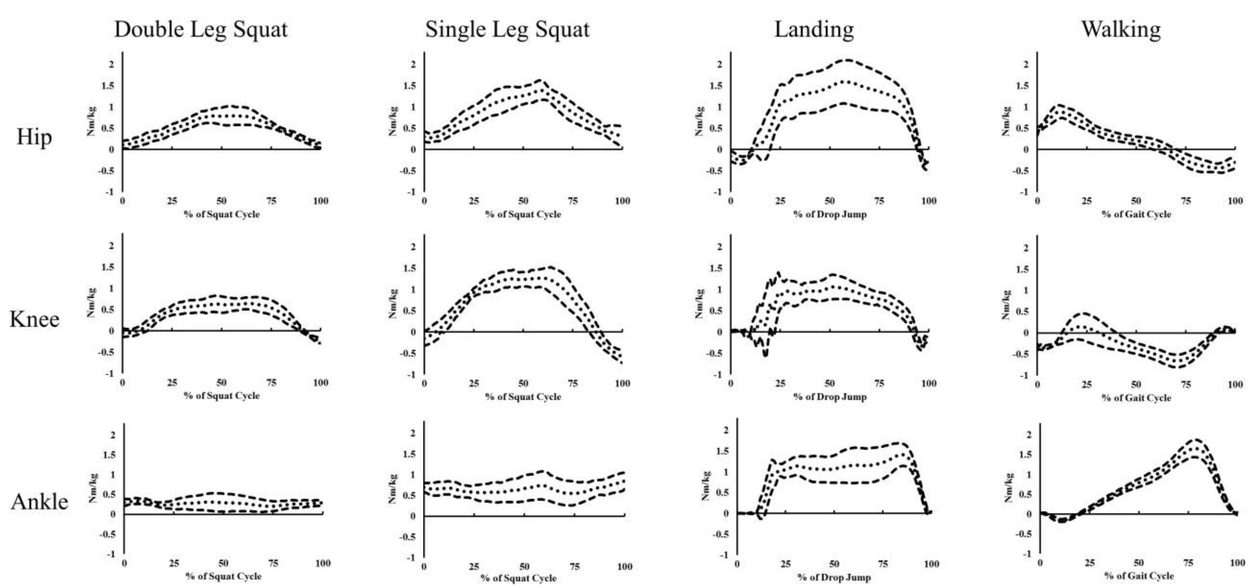

FIGURE 4. The average right hip, knee, and ankle joint torques are shown above for all subjects across the four tasks analyzed. Dashed lines represent +/- one standard deviation from the average (dotted lines). Positive torques represent hip extension torque, knee flexion torque, and plantarflexion torque.

One representative subject's average muscle forces during each of the tasks are shown below in Figure 5.

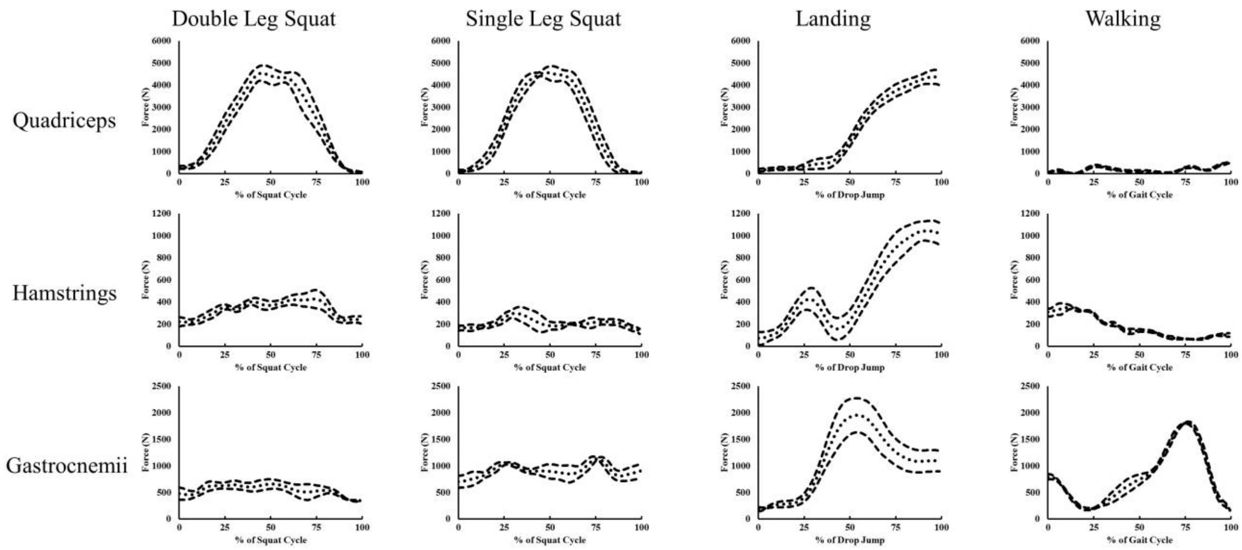

FIGURE 5. Average quadriceps, hamstrings, and gastrocnemii forces for one representative subject during each of the four tasks analyzed. Dashed lines represent +/- one standard deviation from the average (dotted lines). Each muscle's vertical axis was held constant across the different tasks to visualize the difference in magnitudes in muscle forces (Quadriceps 6000 N, Hamstrings 1200 N, and Gastrocnemii 2500 N).

It was hypothesized that the H:Q strength ratio would be correlated with peak ACL forces during the four different tasks. To address this research question, a correlational analysis was run. The Pearson correlation between the H:Q strength ratio and the peak ACL forces normalized to each subject's body weight for all tasks combined was insignificant (Figure 6, $r = -0.12$, $p =$

0.445). Note that for tasks which involved two ACL peaks (i.e., all squat and walking trials), there two data points per subject.

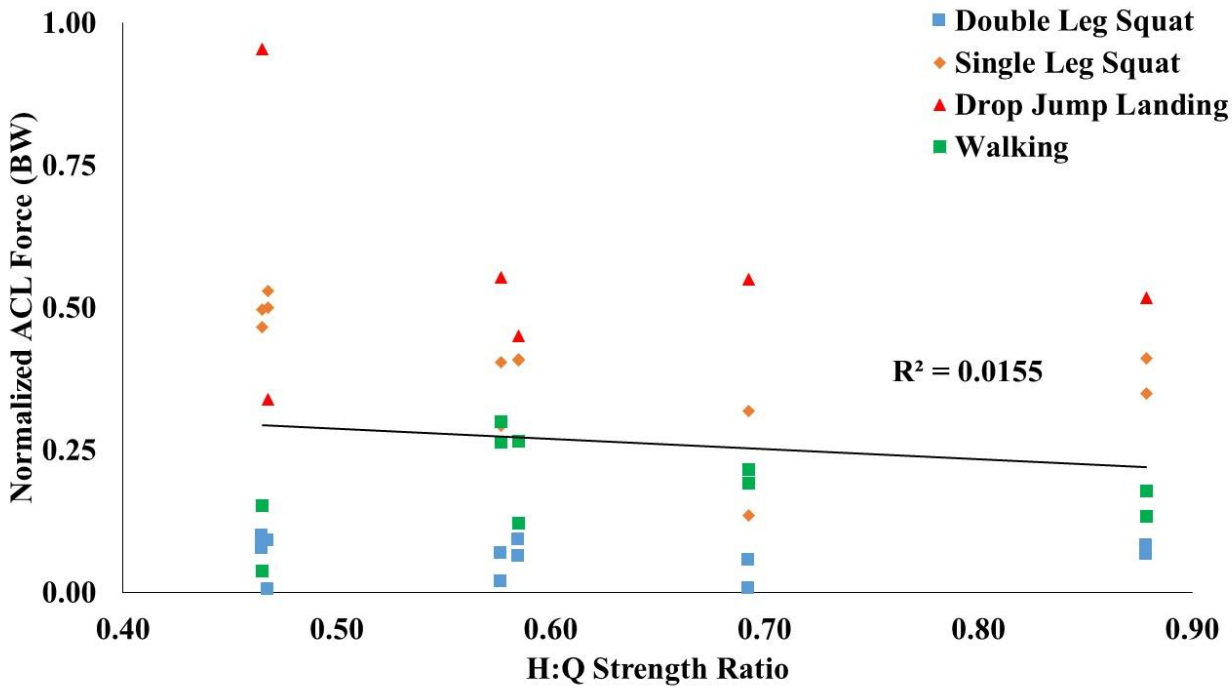

FIGURE 6. Average of peak ACL forces normalized to each subject's body weight in each of the four tasks. Note: X-axis starts at 0.4.

To further investigate this, the different tasks were separated to see if the H:Q ratio was related to normalized, peak ACL forces of a specific task. All relationships were insignificant. Each of the different tasks Pearson correlation and p values are as follows (Figure 7): double leg squatting r = -0.03 (p = 0.925), single leg squatting r = -0.52 (p = 0.086), landing r = -0.21 (p = 0.684), and walking r = 0.06 (p = 0.876).

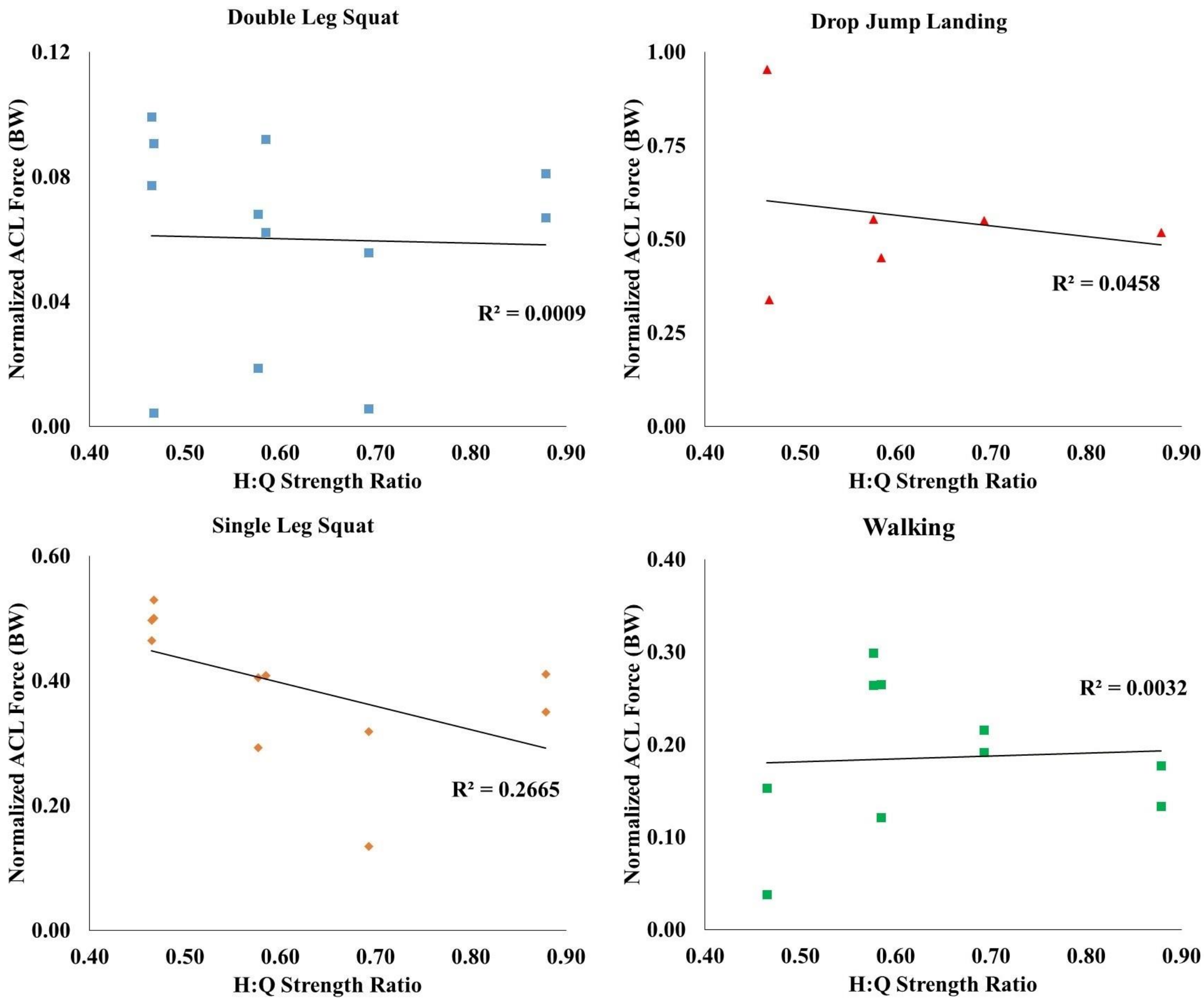

FIGURE 7. Average of peak ACL forces for all tasks normalized to each subject's body weight, separated by the four tasks analyzed in the study. Note: Y-axes have different magnitudes across tasks; this was done to better visualize the correlations between the H:Q strength ratio and the ACL forces. X-axes start at 0.4.

Secondary Purpose – Independent effects of each muscle group

The secondary purpose of this study was to analyze the independent effects of each muscle group's contribution to the peak, normalized ACL forces during each of the tasks. After perturbing the hamstrings or quadriceps muscle peak strength (F_{max}) by +10% or -10% (inherently manipulating the H:Q strength ratio), ACL forces were estimated again for each participant in each trial. Each perturbation's change in ACL forces was compared to the original ACL forces, and confidence intervals were created. This was done in two separate ways: (1) by analyzing the percent change in ACL forces when the muscles' strengths were perturbed by +/- 10% (Figure 8), and by analyzing the absolute change in ACL forces (measured in bodyweights) when the muscles' strengths were perturbed by +/- 10% (Figure 9). By decreasing the strength of the hamstrings by 10% (see Hams -10%), the percent change in ACL forces with corresponding 95% CI showed the ACL forces were significantly increased above the original ACL forces an average of 7.1%, $p < 0.001$ [2.972, 11.265]. Conversely, by increasing the strength of the hamstrings by 10% (see Hams +10%), the percent change and 95% CI indicated that the ACL forces were significantly decreased below the original ACL forces by an average of 5.7%, $p < 0.05$ [-0.091, -8.384]. The -10% change in the quadriceps muscle group's strengths did not cause significant percent changes in the ACL forces, $p = 0.602$ [-3.048, 5.245], and similarly was true for the +10% change in quadriceps strength, $p = 0.489$ [-5.605, 2.687].

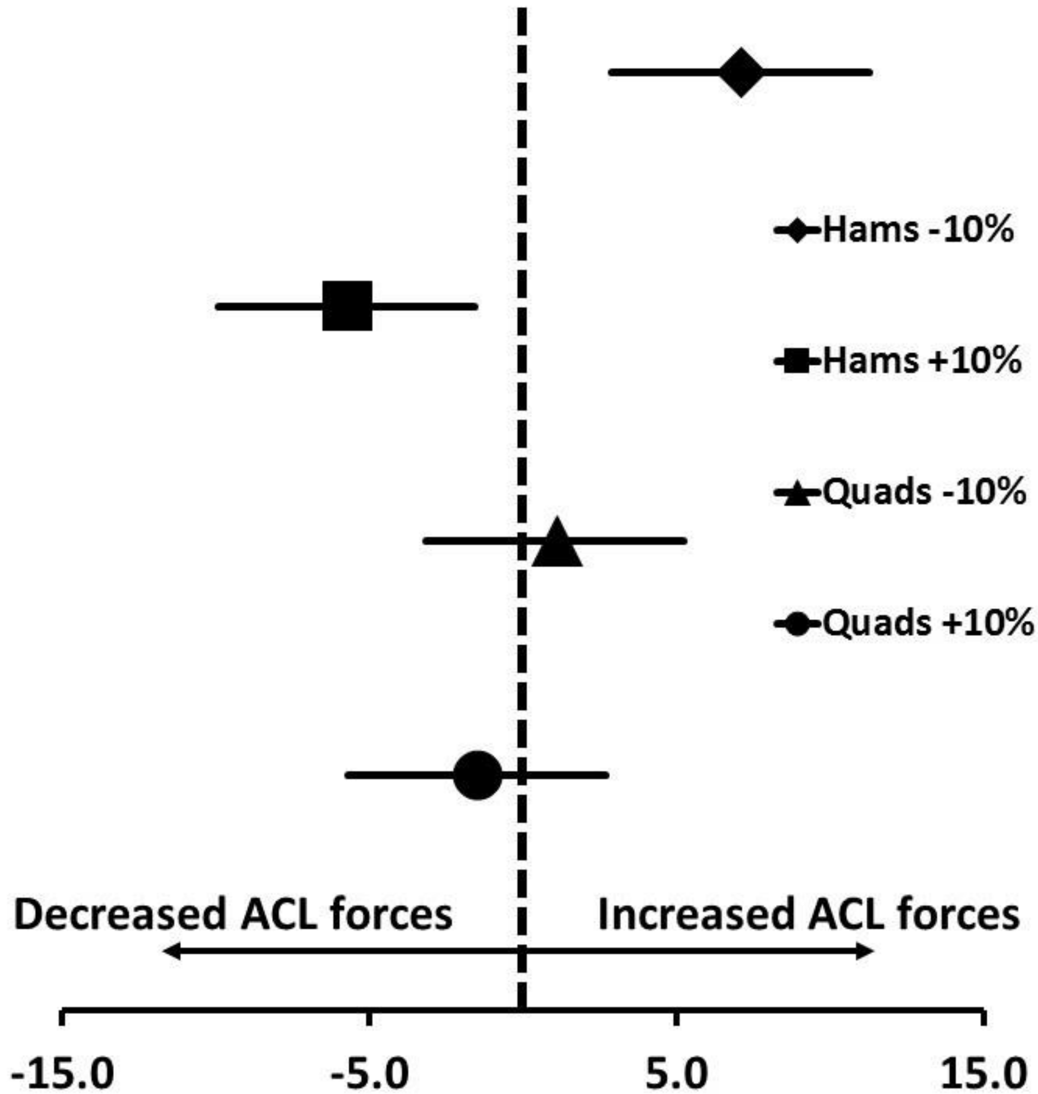

FIGURE 8: Effect of muscle group strength (F_{max}) on percent changes in peak ACL forces for all tasks combined. Hams = hamstring muscle maximal isometric forces perturbed, Quads = quadriceps muscle maximal isometric forces perturbed.

Because the ACL forces were relatively low for some tasks, small changes in ACL forces could have yielded high percent changes in ACL forces from the unperturbed to perturbed strength conditions. This necessitated viewing the changes in ACL forces in absolute terms. Thus, the same analysis as above was performed again, but the ACL forces were converted to bodyweights (BW) and the CI's were recreated (Figure 9). When the hamstrings strength was decreased by 10%, there was a significant increase in ACL forces of 0.00589 BW, $p < 0.05$ [0.00507, 0.00671]. When the hamstrings strength was increased by 10%, there was a significant decrease in ACL forces of -0.00692 BW, $p < 0.05$ [-0.00775, -0.00610]. These results are similar to the above analyses performed with percent changes. When the quadriceps strength was decreased by 10%, there was a significant increase in ACL forces of 0.00143 BW, $p < 0.05$ [0.00042, 0.00244]. And lastly, when the quadriceps strength was increased by 10%, there was a significant decrease in ACL forces of -0.00376 BW, $p < 0.05$ [-0.00490, -0.00263]. These results were significant, unlike the percent changes in ACL forces after the quadriceps strength perturbations mentioned above. It is worth noting that the effect of the quadriceps strength changes were still smaller in absolute terms, similar to the percent change analysis performed above. To give the reader an idea of the magnitude of these changes, the average mass of the males in the present study was 74.7 kg, or roughly 730 N, so a 0.005 BW change in ACL force would equate to ~4 N.

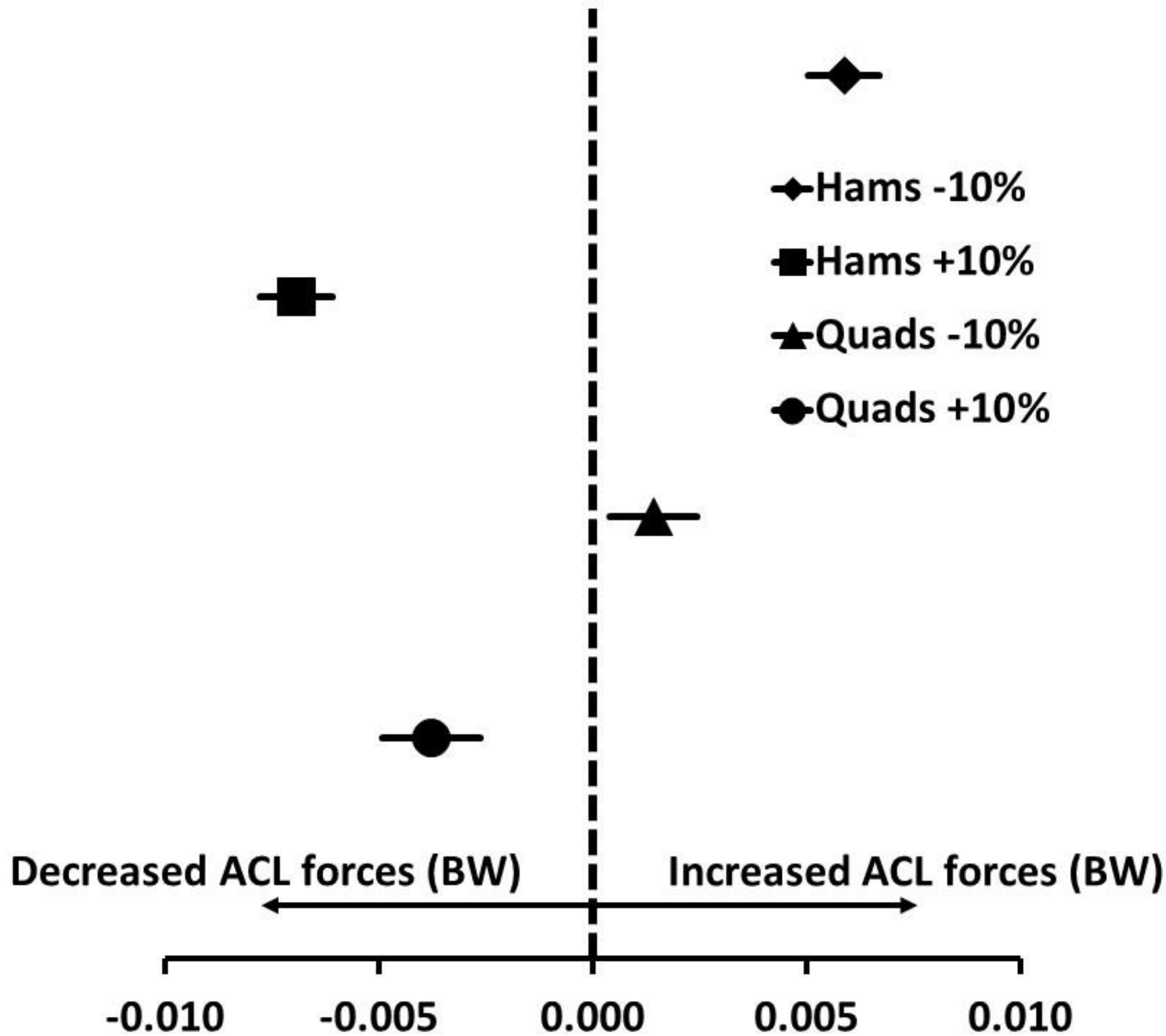

FIGURE 9: Effect of muscle group strength (F_{max}) on changes in peak ACL forces (measured in bodyweights [BW]) for all tasks combined. Hams = hamstring muscle maximal isometric forces perturbed, Quads = quadriceps muscle maximal isometric forces perturbed. For a 75 kg person, a 0.005 BW change in ACL force would equate to roughly a 4 N difference.

To confirm the consistency of these results across subjects, subject-by-subject analyses were also conducted (Figures 10 and 11). A Pearson correlation was run on the percent change in ACL forces and the percent change in the H:Q strength ratio given the perturbation either occurring to the hamstrings or quadriceps, separated by subject (Figure 10 for hamstrings, Figure 11 for quadriceps). It was found that changing the hamstrings F_{max} was significantly correlated with changes in ACL forces ($r = -0.45$, $p < 0.001$), however changing the quadriceps F_{max} was not significantly correlated with changes in ACL forces ($r = -0.17$, $p = 0.122$).

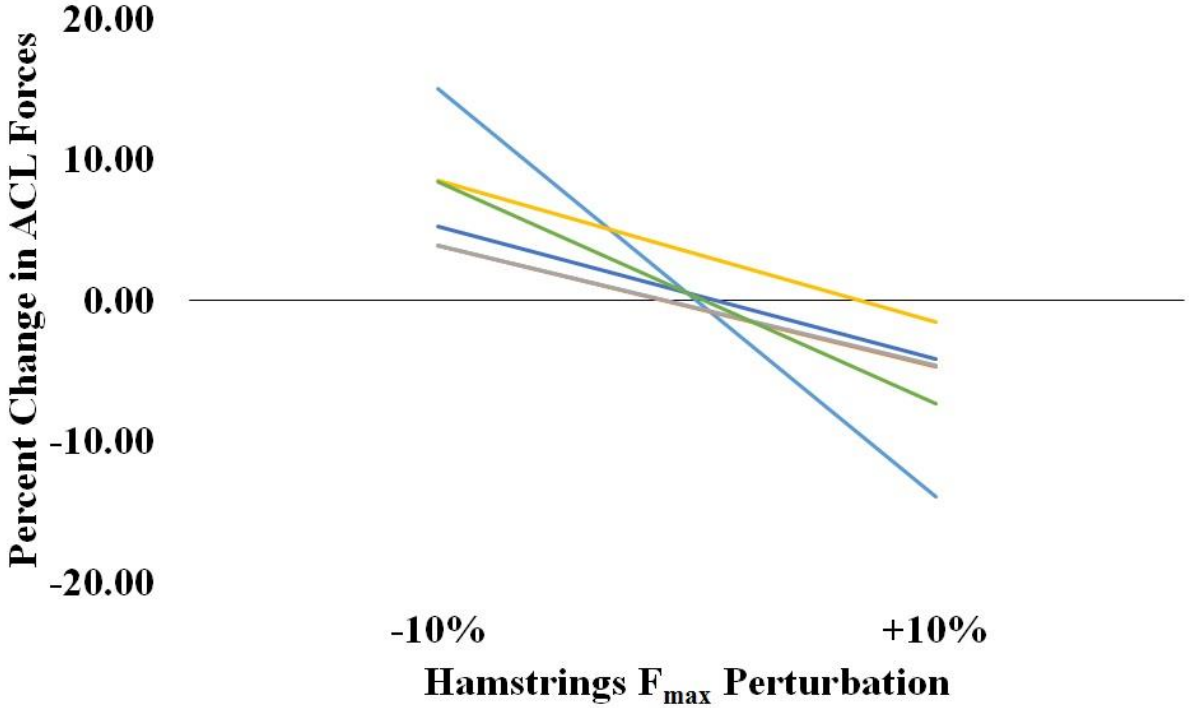

FIGURE 10. Effect of F_{max} perturbations of the hamstrings muscle group on peak ACL forces by subject. Each line represents a different subject (two subjects were very similar thus making it seem like one is missing).

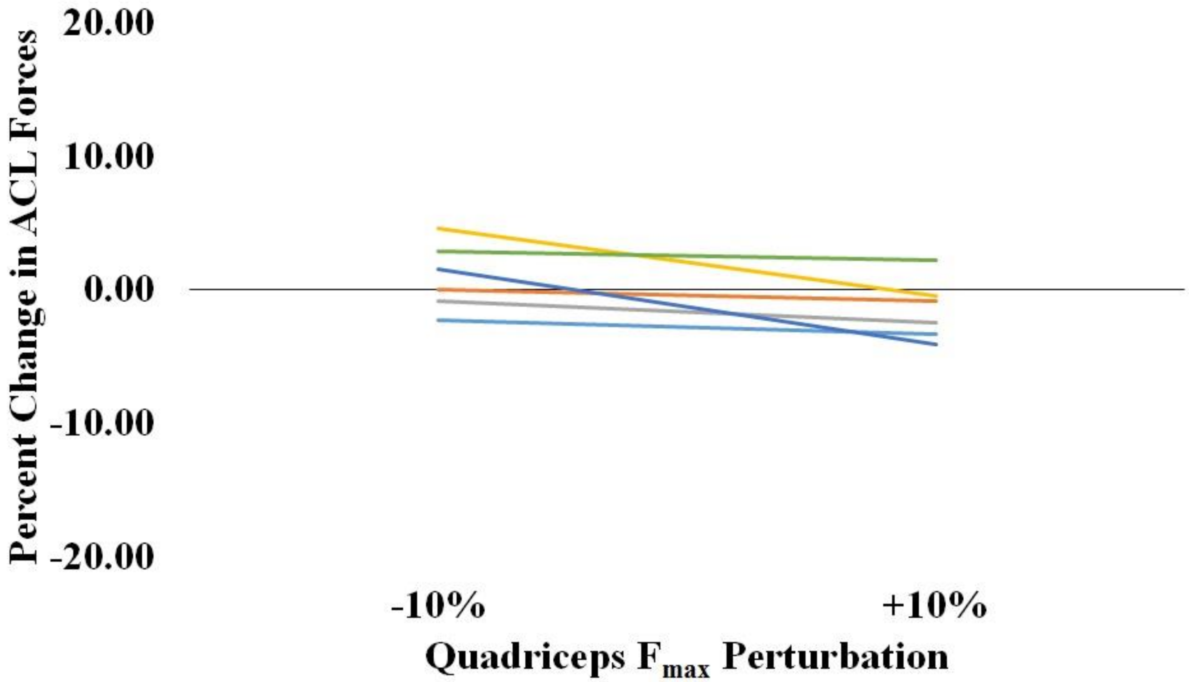

FIGURE 11. Effect of F_{max} perturbations of the quadriceps muscle group on peak ACL forces by subject. Each line represents a different subject.

Chapter V: Discussion

The primary purpose of this study was to determine the relationship between the H:Q strength ratio and peak ACL forces during the following tasks: single and double-leg squatting, a drop jump, and walking. The secondary purpose was to determine the independent effects the hamstrings and quadriceps strength had on peak ACL forces during the same tasks by perturbing the maximal isometric force production of each muscle group within a subject-specific musculoskeletal model. It was hypothesized that the H:Q strength ratio would be related to peak ACL forces because of previous research showing that each muscle group can influence forces on the ACL (Markolf et al., 2004). It was also hypothesized that a reduction in hamstring strength or increase in quadriceps strength would increase peak ACL forces, while an increase in hamstrings strength and reduction in quadriceps strength would decrease peak ACL forces.

The results of this study indicate that the H:Q strength ratio is not related to peak ACL forces for all tasks combined, or when separated by task. Furthermore, perturbing the hamstrings F_{max} by -10% caused a significant increase in peak ACL forces by 7.1%, and perturbing the hamstrings F_{max} by +10% caused a significant decrease in ACL forces by 5.7%. The -10% and +10% quadriceps F_{max} perturbations did not cause a significant percent change in peak ACL forces. Lastly, the +10% and -10% changes in hamstrings strength were associated with ACL force changes ($r = -0.45$, $p < 0.001$), but quadriceps strength changes were not ($r = -0.17$, $p = 0.122$). When analyzed in terms of bodyweights, the -10% hamstrings strength perturbations caused a significant increase in ACL force of ~0.006 BW, and the +10% hamstrings strength perturbation caused a significant decrease in ACL force of ~0.007 BW. The -10% quadriceps strength perturbations caused a significant increase in ACL force of ~0.001 BW, and the +10% quadriceps strength perturbation caused a significant decrease in ACL force of ~0.004 BW.

While the primary hypothesis was not supported, there are several potential reasons explaining this lack of association between the H:Q strength ratio and ACL forces: 1) the maximal strength of the hamstrings and quadriceps did not occur at the same knee angle and therefore the H:Q strength ratio may be inappropriate to infer directly to ACL loading, 2) it is reasonable that the positions the hip and knee were strength tested in are not directly relatable to normal activities of daily living, and 3) maximal effort (i.e., strength) does not relate to performance in submaximal effort tasks.

The H:Q strength ratio is defined as the ratio of the maximal strength of the hamstrings to the maximal strength of the quadriceps, and is clinically measured using a dynamometer. From our dynamometer data, the angles at which peak flexion (i.e., hamstrings and gastrocnemii) torque and peak extension (i.e., quadriceps) torque occurred were not the same for any subject. During the 60º/sec maximal effort knee extension-flexion isokinetic dynamometer test, maximal flexion torque was generated at 27.8º +/- 11.5º of knee flexion and maximal extension torque was generated at 67.2º +/- 7.8º of knee flexion. These angles compare well to two previous studies investigating isokinetic strength: 1) adolescent female soccer players whose isokinetic strength was measured at 30º/sec found peak hamstrings torque occurred at 38º and peak quadriceps torque occurred at 73º (Costain & Williams, 1984), and 2) healthy adult males and females whose isokinetic strength was measured at 60º/sec found peak hamstrings torque occurred at 33º for men and 37º for women, and peak quadriceps torque occurred at 54º for both sexes (Kannus & Beynnon, 1993). The differences in angles where peak torque occurs makes sense given the muscle force-length relationship. The maximal amount of force (i.e., torque) that a muscle can generate occurs at an optimal length, and this is influenced by the flexion angle of the joint. Thus, since the maximal torque generated in each muscle group did not occur at the same knee flexion angle, the clinical H:Q ratio does not seem to be an appropriate measure to make inferences about ACL loading at specific knee angles. This is particularly true for low knee flexion angles, where peak ACL forces typically occur and where the hamstrings are capable of producing closer to their optimal amount of force compared to the quadriceps, which occurs at larger knee flexion angles.

Because the H:Q strength ratio was comprised of peak torques that occurred at different knee angles, we investigated whether the H:Q strength ratio taken at knee angles where peak ACL forces occurred would correlate with peak ACL forces. The average knee angle where peak ACL force occurred across all subjects for all tasks was 28.6º ± 14.1º, thus we pulled the H:Q strength ratio at a knee flexion angle of 29º and correlated it with peak ACL forces normalized to body weight across all subjects and all tasks' averages; this correlation was not significant ($p = 0.558$, $r = -0.096$). The correlation was still not significant when done across all subjects and all trials rather than averages of trials peak ACL forces ($p = 0.220$, $r = -0.090$). Because the knee flexion angle where peak ACL forces occurred depended on the task being performed, we also correlated peak ACL forces at other knee flexion angles depending on the task. Peak ACL forces occurred at the following knee angles for each task: first double leg squat peak = 22.7º ± 10.7º, second double leg squat peak = 31.3º ± 6.1º, first single leg squat peak = 39.6º ± 6.9º, second double leg squat peak = 45.9º ± 5.4º, landing peak = 38.9º ± 6.5º, first walk peak = 11.5º ± 4.9º, and second walk peak = 10.3º ± 3.2º. For double leg squats, the H:Q strength ratio at 29º was used (HQ29), for single leg squats and landing the H:Q strength ratio at 40º was used (HQ40), and for walking the H:Q strength ratio at 11º was used (HQ11). Using the averaged peak ACL forces normalized to body weight from each task, the correlations between the H:Q strength ratios at specific angles and peak ACL forces were as follows: double leg squat and HQ29 was not significant ($p = 0.814$, $r = -0.76$), single leg squat and HQ40 was not significant ($p = 0.168$, $r = -0.426$), landing and HQ40 was not significant ($p = 0.666$, $r = -0.227$), and walking and HQ11 was not significant ($p = 0.586$, $r = 0.197$).

The positions the subjects were strength tested in also do not translate directly to most athletic activities or activities of daily living. The subjects were seated in the dynamometer with their hips flexed at 90º, and concentrically contracted their quadriceps and hamstrings through a 100º range of motion (i.e., flexed to fully extended for quadriceps, and in the reverse direction for hamstrings). When the hips are at 90º, and especially when combined with an extended knee, the hamstrings are in a lengthened position such that they can generate more torque than if the hips were more extended (Lunnen, Yack, & LeVeau, 1981). High hip flexion coupled with extended knee positions are uncommon in most tasks. This has been demonstrated in a 60cm drop-jump landing task with a preferred amount of trunk flexion; at the instant of ground contact the knee angle was 6±7º and hip angle was 14±12º, and during the loading phase (i.e., the interval between ground contact and peak knee flexion) the maximal knee flexion angle was 69±16º and the maximal hip flexion angle was 40±20º (Blackburn & Padua, 2008).

To further support the idea that the position that we strength tested the subjects in does not relate to positions of activities of daily living, we analyzed the operating ranges of the BFLH and VL in the present study (Figure 12 for BFLH, Figure 13 for VL). A typical muscle force-length curve is on each figure for comparison. Each figure depicts the operating ranges of normalized fiber lengths for the BFLH or VL, respectively, during the tasks analyzed in the study in addition to model predicted normalized fiber lengths under isometric conditions. The modeled normalized fiber lengths were created by putting each subject specific model's hips at 90º of flexion and plotting the normalized fiber lengths between 0 and 100º of knee flexion with an activation level of 1.00. The angle at which each subject achieved maximal flexion or extension torque during the isokinetic testing was matched to the normalized fiber length at that same angle and plotted on both graphs as a colored circle for each person. The model predicted normalized fiber lengths for the BFLH and VL during walking match well with a previous study investigating operating fiber length ranges during walking (Arnold & Delp, 2011). It can be seen that for the BFLH during the dynamometer testing, the typical operating range begins in the latter parts of the ascending limb, extends through the plateau region, and goes deep into the descending limb of the muscle force-length curve. Also, each subject's maximal torque production occurred at low knee flexion angle and therefore at a high normalized fiber length (average normalized fiber length of ~1.29). In other words, the position the subjects are in when using the dynamometer typically stretches the BFLH into the descending limb of the muscle force-length curve, where it can generate higher amounts of torque from the passive elements of the muscle. The BFLH normalized fiber lengths during each task were typically in the ascending limb (walking), plateau region (landing), or extended briefly into the descending limb (double and single leg squatting), but not to the same extent as the dynamometer trial. Therefore it is unlikely that someone can match the same peak torque during our investigated tasks as compared to the torque generated on a dynamometer unless their hips are flexed to such large angles as when using the dynamometer. The VL operating ranges during each task suggest there is more overlap of operating ranges between the tasks and the dynamometer, potentially indicating that the maximal quadriceps strength may relate more to the tasks in this study. Regardless, the H:Q strength ratio is a ratio of both muscle groups' strengths, and since the hamstrings' operating ranges estimates strength at a position not relatable to the tasks analyzed, the ratio likely has little translatable value. Ultimately, during a typical movement (even if it is maximal effort) it is unlikely that someone will be able to match their H:Q strength ratio in terms of how much torque is being generated in each muscle group at any instant in time.

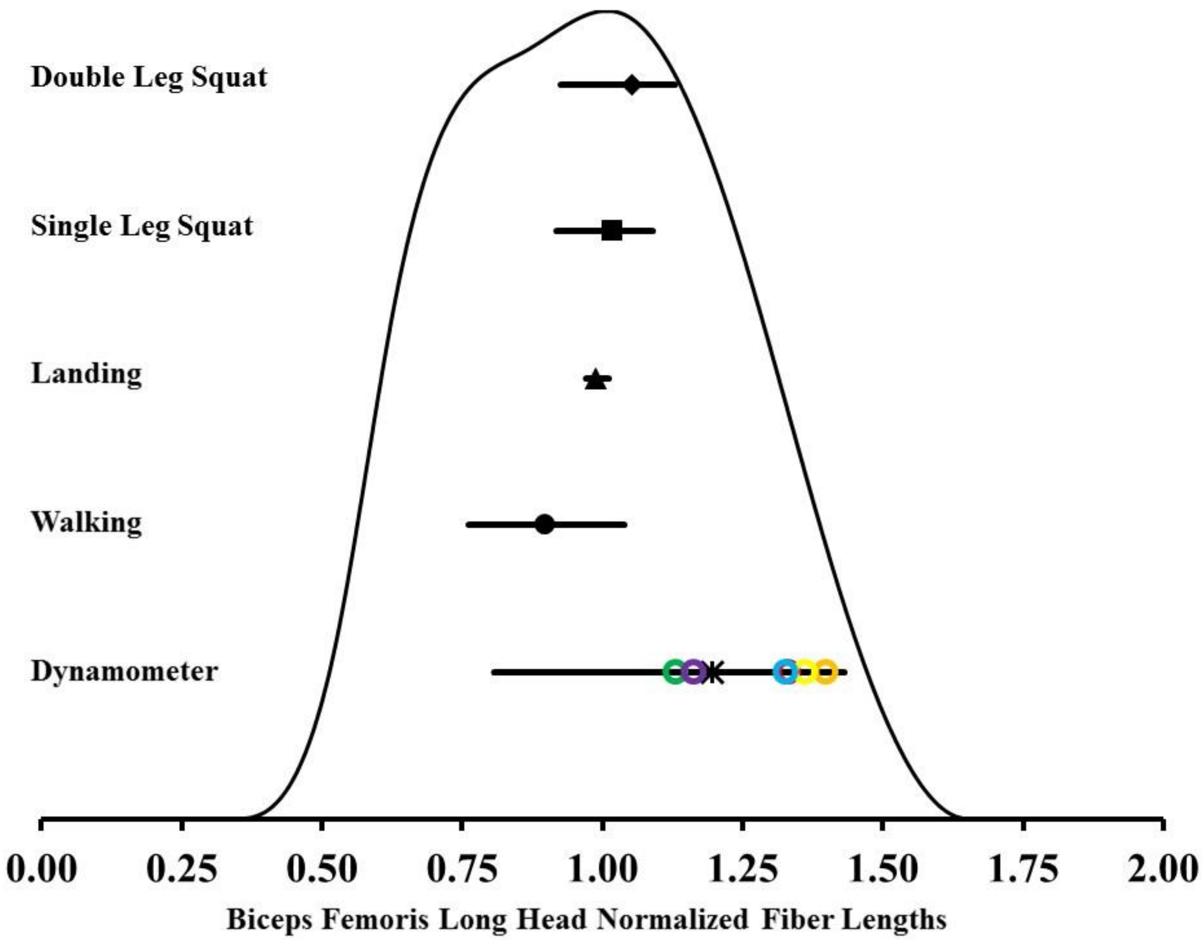

FIGURE 12: BFLH operating ranges of normalized fiber lengths during the separate tasks in the study and during the isokinetic dynamometer trial. Each bar is the averaged normalized fiber length across all subjects for each task, going from minimum to maximum with the average depicted with the shape in the middle. The dynamometer bar includes color coded circles that match the angle to the normalized fiber length where each subject achieved maximal torque output during the isokinetic trial.

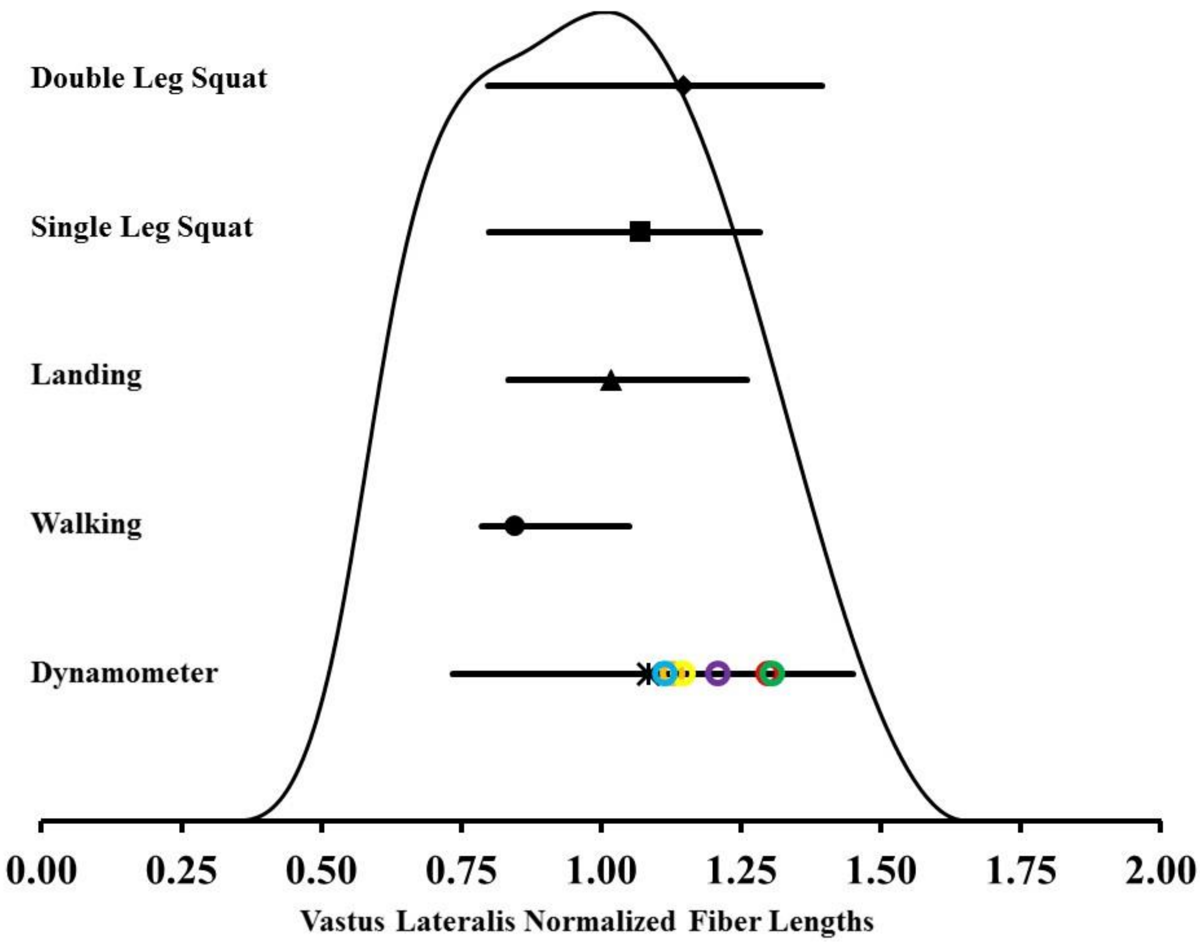

FIGURE 13: VL operating ranges of normalized fiber lengths during the separate tasks in the study and during the isokinetic dynamometer trial. Each bar is the averaged normalized fiber length across all subjects for each task, going from minimum to maximum with the average depicted with the shape in the middle. The dynamometer bar includes color coded circles that match the angle to the normalized fiber length where each subject achieved maximal torque output during the isokinetic trial.

Lastly, how maximal strength relates to submaximal tasks, and thus forces on the ACL during submaximal tasks, is not clear. The results of the present study match well with a previous study investigating the effects of eccentric quadriceps strength and concentric hamstrings strength on anterior tibial shear forces during a similar drop jump landing task (a movement the authors deemed submaximal); it was found that neither of those strength measurements, nor the ratio of the two, were significant predictors of anterior tibial shear force (Bennett et al., 2008). Although anterior tibial shear forces are not ACL forces, they are thought to be surrogate measures of ACL loading, thus showing the weak link between thigh muscle strength and potential ACL injury. It is also important to consider that even though a task may be considered maximal effort, such as a vertical jump, sprint, or isometric/isokinetic dynamometer trial, it is unlikely that a person is activating a muscle group maximally during the task, especially any two muscle groups simultaneously as implied by the H:Q strength ratio. It has been shown that normal, submaximal activities of daily living are not performed using a young, healthy adult's maximum strength or maximal activation (Hortobagyi, Mizelle, Beam, & DeVita, 2003), like the subjects and the movements in the present study. Young subjects in this study ascended and descended stairs as well as rose from a chair, and the peak knee joint moments and EMG activity during these movements were compared to the maximal isometric knee joint moments and maximal EMG activity during a maximal effort leg press. Relative effort needed to complete each task was as follows: 54±16% for stair ascent, 42±20% for stair descent, and 42±19% for rising from a chair. Furthermore, the VL was only activated to 28±20% when ascending stairs, 33±21% when descending stairs, and 29±22% when rising from a chair. In a separate study analyzing a drop jump task, the maximal strength of the hamstrings and quadriceps were only poor to moderate predictors of their activation levels, and were not predictors of knee range of motion or knee moments during the task (Shultz, Nguyen, Leonard, & Schmitz, 2009). Therefore using the H:Q strength ratio seems inappropriate when discussing tasks that do not require maximal activation or strength of both muscle groups simultaneously.

The secondary hypothesis was partially supported: there was a significant relationship between the -10% and +10% changes in hamstring strength and percent changes in ACL forces, however this was not the case with the quadriceps strength perturbations. When analyzed in terms of bodyweights rather than percent changes, the results of the hamstrings perturbations were still significant and in the same direction, but the quadriceps strength perturbations became statistically significant, and in the opposite way than previously hypothesized. The effect of the quadriceps perturbations were still lower than the hamstrings, however. Because of this, it appears that hamstring strength has a more direct influence on ACL forces and therefore potentially ACL injury risk. An explanation for why there was a low effect overall is that the -10% and +10% perturbations of hamstrings and quadriceps F_{max} within SIMM had very little effect on the muscular forces being produced during the submaximal movements studied. Given that the hybrid static optimization procedure used matched the joint torques being produced in the movements and these joint torques did not change, the activation level by each muscle did not have to change much in magnitude. Furthermore, the tasks in the present study produced relatively low ACL forces, far below what is necessary to cause damage to the ligament. The maximal amount of estimated ACL force was 614N during a male subject's drop jump landing trial (~.72 BW), and the maximal amount of estimated ACL force for females was 564N (~1.03 BW), again during a drop jump landing trial. In cadaveric ACLs, the male ligament could withstand up to 1818±699N before failure and the female ligament could withstand up to 1266±527N before failure (Chandrashekar, Mansouri, Slauterbeck, & Hashemi, 2006). Thus the ACL forces in the present study most likely never reached 50% of the force needed to cause injury to the ACL in either sex. Despite the fact that the ACL forces in the present study were low in comparison to what is necessary to cause ligament damage, the significance of lowering

ACL forces should not be undermined, especially for an ACL rehabilitation setting. It has been shown that cyclical loading of different ACL graft types with physiological forces (50-250N, comparable to the ACL forces in the present study across all of the tasks) caused migration of the graft within the femoral tunnel (Staerke, Möhwald, Gröbel, Bochwitz, & Becker, 2010). Furthermore, the ultimate failure load of numerous types of ACL grafts are below the failure loads presented by Chandrashekar et al., with 7 of 8 graft techniques failing at ACL forces as low as 572-977 N, and with one technique failing at 1345 N (Brown, Wilson, Hecker, & Ferragamo, 2004). Therefore, any reduction in ACL force could be clinically relevant to an ACL reconstructed knee.

The perturbations to hamstrings strength did have an effect on peak ACL forces in both percent changes and in bodyweight changes, thus absolute hamstring strength may have more of an influence on ACL forces and injury than absolute quadriceps strength. In females, it has been demonstrated that hamstring strength less than 75% of their quadriceps strength (i.e., H:Q ratio under 0.75) increased lower extremity injury risk by 1.6 times (Knapik, Bauman, Jones, Harris, & Vaughan, 1991). Also, females with a 15% stronger right leg hamstring muscle group compared to their left suffered lower extremity injuries 2.6 times more often than females without the strength imbalance, with injury occurring more often in the weaker left side (Knapik et al., 1991). It is important to note that this study was of female collegiate athletes, whose isokinetic strength was measured at 180º/sec (hence the high H:Q strength ratio of 0.75), and injuries included “any traumatic event or overuse impairment that occurred during practice or competition for which the athlete or her coach referred the athlete to the athletic trainer or outside medical care and for which there was some time lost from practice or competition” (Knapik et al., 1991). Regardless, because injury risk was higher when there was a hamstring strength imbalance between right and left legs, it appears that absolute hamstring strength (or a deficit in hamstring strength), not the ratio of the hamstring strength to quadriceps strength, is more related to injury risk.

Since the H:Q strength ratio was not related to peak ACL forces across the different tasks, we investigated how the muscle forces were related to ACL forces; we did this because the muscle forces are inherently task dependent, unlike someone's maximal strength. Specifically, we looked at the average force the hamstrings and quadriceps were producing at the times of peak ACL forces (i.e., the H:Q force ratio at each of the two peaks in double and single leg squats, the single peak in landing, and the two peaks in walking) and plotted it against the peak ACL force in bodyweights (Figure 14). The relationship between the H:Q force ratio and the peak ACL force was non-linear in nature, and best represented with a logarithmic fit (R^2 = 0.7829). This relationship implies that the muscular forces at each instant in time are having an effect on the ACL forces, and that after reaching a H:Q force ratio of roughly 0.5, the effect diminishes fairly rapidly. This relationship is important because it contrasts the non-significant relationships between muscular strength and peak ACL forces. The ACL forces during the tasks analyzed are not dependent on the ratio of the strength of the hamstrings and quadriceps, but are dependent on the amount of force being produced in those muscle groups.

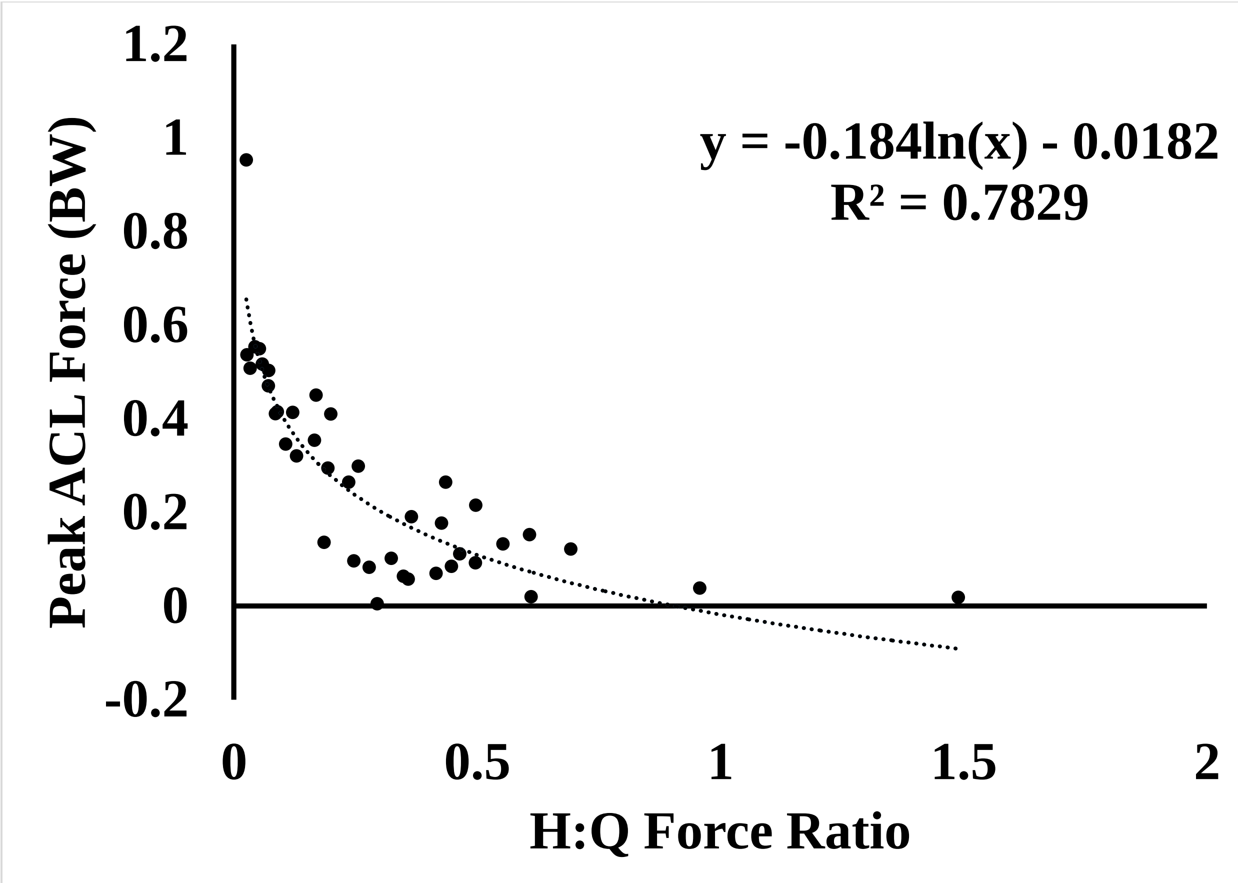

FIGURE 14. The relationship between the H:Q force ratio and peak ACL forces across all subjects and tasks for each of the seven ACL force peaks analyzed (two from each double and single leg squat, one from landing, and two from walking).

Limitations

There are certain limitations to the present study worth mentioning. The model within SIMM was edited to be subject specific, but some characteristics were either not directly changed or were not changed at all. For example, the moment arms of each of the muscles analyzed in the study were simply scaled up or down from the generic model built into SIMM according to the subject's anthropometrics. This may have led to longer or shorter moment arms for certain muscles, which has a direct influence on the torque that the muscle produces. Another limitation is how FL_{opt} was determined and edited within SIMM for each subject's muscles. It was estimated by multiplying the measured muscular fascicle length (assumed to be equal to the muscle fiber length) by a ratio of the optimal sarcomere length of 2.7μm (Lieber et al., 1994) to the measured sarcomere lengths from a cadaveric study (Ward et al., 2009). Again, these values are inherently variable for each subject's individual muscles, but the model still allowed for subject-to-subject variations in the location of optimal fiber length because the fascicle length (measured via ultrasound) to muscle length ratios (model predicted) were different from subject to subject. Certain other assumptions were made that influenced the estimation of both muscular and ACL forces. It was assumed that each subject had a -7º PTS, although the actual slope for each individual may have been much different (Hudek et al., 2009). This is discussed in Appendix D. Also, the activation level of the VI was assumed to be equal to the average of the VL and VM, the ST activation level was equal to the SM, and the BFSH was equal to the BFLH, although other authors have used this method before (Lloyd & Besier, 2003). Despite the assumptions made, we believe the peak ACL forces produced within the study are reasonable, as outlined above in the Methods section.

There are also limitations with regards to strength testing. Although we attempted to capture each individual's true maximal strength, it is likely we may have underestimated their strength. This could be partially explained by the central activation ratio (CAR). The CAR implies that individuals often are not producing the maximal amount of muscular force possible during a maximal voluntary contraction, and are actually capable of producing more muscular force if the muscle is electrically stimulated (Kent-Braun & Le Blanc, 1996). Thus, even when a person is asked to exert a maximal amount of effort, researchers may not be measuring a person's true maximal potential unless the muscle is electrically stimulated. Furthermore, the motivation level of a subject can potentially influence the maximal amount of torque generated during a dynamometer test (McNair, Depledge, Brettkelly, & Stanley, 1996). Subjects were verbally encouraged during the dynamometer protocol to put forth as much effort as possible to counteract this effect, but it is still likely that some subjects did not produce a true maximal amount of torque in either muscle group while on the dynamometer, or during any of the movements analyzed. The CAR and the motivation level of each individual make it difficult to make statements about the accuracy of strength measurements, and especially how the H:Q ratio relates to submaximal tasks.

Conclusion

The results of the present study did not support the primary hypothesis that the H:Q strength ratio was related to peak ACL forces during the tasks analyzed. Furthermore, the secondary hypothesis was partially supported: perturbations to the hamstrings F_{max} were significantly correlated with percent changes in ACL forces, however perturbations to the quadriceps F_{max} were not. Specifically, a decrease in hamstrings strength was related to an increase in ACL force, and an increase in hamstrings strength was related to a decrease in ACL force. In terms of bodyweight changes in ACL forces, the same was true for the hamstrings as it was in the percent change analysis, but the quadriceps became statistically significant as well with a decrease in quadriceps strength leading to an increase in ACL forces, and an increase in quadriceps strength leading to a decrease in ACL forces. The effect the quadriceps had was smaller than the effect of the hamstrings. These results suggest the possibility of the H:Q strength ratio not being indicative of ACL forces during submaximal tasks, and limiting its predictive ability of ACL injury risk. The maximal strength of the hamstrings, not the quadriceps or the ratio of the strengths of the two muscle groups, may be a driving factor behind ACL forces during such tasks. It is more likely that the muscular forces at each instant in time, modulated by the knee flexion angle and current muscle activation states, are more related to ACL forces (and hence injury risk) than the maximal strengths of either muscle group.

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Appendix A: Informed Consent Form and IRB Approval Letter

Study ID:UMCIRB 15-000127 Date Approved: 8/5/2015 Expiration Date: 8/4/2016

East Carolina University

Informed Consent to Participate in Research

Information to consider before taking part in research that has no more than minimal risk.

Title of Research Study: Effect of Previous Hamstring Strains on Anterior Cruciate Ligament Loading

Principal Investigator: Anthony S Kulas
Institution, Department or Division: ECU, Department of Health Education and Promotion Address: 249 Ward Sports Medicine Building
Telephone #: 252-737-2884

Researchers at East Carolina University (ECU) study issues related to society, health problems, environmental problems, behavior problems and the human condition. To do this, we need the help of volunteers who are willing to take part in research.

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

The purpose of this research is to determine how previous hamstring strain injury affects loading on the Anterior Cruciate Ligament (ACL) in the knee. You are being invited to take part in this research because you qualify into one of two groups, 1) are recreationally active, healthy, and have never experienced a hamstring strain injury, or 2) have a past history of hamstring strain injury, but are otherwise healthy now. The decision to take part in this research is yours to make. By doing this research, we hope to learn how the quadriceps and hamstrings affect loads placed on the ACL, specifically after hamstring strain injury.

If you volunteer to take part in this research, you will be one of about forty (40) people to do so, 20 with a past of hamstring strain injury and 20 without any previous injury.

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

I understand I should not volunteer to be in this study if I am under 18 years of age, have any known allergies to hypoallergenic gel commonly used with ultrasound imaging, or I have not participated in any resistance training exercise within the past year. If I qualify as a member of the group with a past history of a hamstring strain, but this injury is still causing me pain or discomfort with any activity, I should not volunteer.

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

You can choose not to participate.

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

The research will be conducted at Ward Sports Medicine Building at East Carolina University .You will need to come to the Biomechanics Laboratory located in room 332 in the Wards Sports Medicine Building twice during the study. The total amount of time you will be asked to volunteer for this study is roughly three (3.5) hours (1.5 hours on day 1, and 2 hours on day 2.

What will I be asked to do?

Consent Version # or Date: ____________

____
Participant's Initials

Title of Study: Effect of Previous Hamstring Strain Injury on Anterior Cruciate Ligament Loading

On day 1, you will be asked to do the following: lie on a standard treatment table while ultrasound images three different muscle groups crossing your knee. On day 2, you will have non-invasive surface electrodes placed on your thigh muscles, you will be seated in a chair and then asked to perform several knee extension and flexion contractions while your muscle strength is measured and the activity of your muscles recorded. Afterwards, reflective markers will be placed on your feet, legs, pelvis, and torso and you will be asked to perform several different movement tasks (squats, drop jumping, and walking) while a motion capture system captures your movements. All electronic data collected during these procedures is for research purposes only and your name, age, face or other visibly identifiable information will not be included or recorded.

What might I experience if I take part in the research?

We don't know of any risks (the chance of harm) associated with this research. Any risks that may occur with this research are no more than what you would experience in everyday life. We don't know if you will benefit from taking part in this study. There may not be any personal benefit to you but the information gained by doing this research may help others in the future.

Will I be paid for taking part in this research?

We will not pay you for the time you volunteer while being in this study.

Will it cost me to take part in this research?

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

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

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

Any agency of the federal, state, or local government that regulates human research. This includes the Department of Health and Human Services (DHHS), the North Carolina Department of Health, and the Office for Human Research Protections.

The University & Medical Center Institutional Review Board (UMCIRB) and its staff have responsibility for overseeing your welfare during this research and may need to see research records that identify you.

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

If you elect to enroll in this study by signing this informed consent document, you will be assigned an alphanumeric code. Only this alphanumeric code, not your name, will appear on the saved ultrasound images or any other electronically saved measurements. All data collected from you will only have this alphanumeric code associated with it and this data will be backed up on a network server in this lab. The only person to have access to the master list of names which link your name to your alphanumeric code will be the principal investigator identified above, Dr. Anthony S. Kulas or other key investigators, Mr. Jeffrey Patterson or Mr. Alex Geronimo. All paperwork and forms linking you to the study will be kept in the PIs office (Ward Sports Medicine – room 249) which remains locked except when in use. Your ultrasound images or the data collected from you in this study may be used for manuscript/presentation purposes. If used for these reasons, no information identifying you (your name or alphanumeric code) will be on any images/figures used for research purposes.

What if I decide I don't want to continue in this research?

You can stop at any time after it has already started. There will be no consequences if you stop and you will not be criticized. You will not lose any benefits that you normally receive.

Consent Version # or Date: ____________

____
Participant's Initials

Title of Study: Effect of Previous Hamstring Strain Injury on Anterior Cruciate Ligament Loading

Who should I contact if I have questions?

The people conducting this study will be able to answer any questions concerning this research, now or in the future. You may contact the Principal Investigator (Dr. Anthony S. Kulas) at (252) 737-2884 (days, between 8am-5pm), or Mr. Jeffrey Patterson and Mr. Alex Geronimo at (252) 737-4616.

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

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

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

I have read (or had read to me) all of the above information.
I have had an opportunity to ask questions about things in this research I did not understand and have received satisfactory answers.
I know that I can stop taking part in this study at any time.
By signing this informed consent form, I am not giving up any of my rights.
I have been given a copy of this consent document, and it is mine to keep.

Participant's Name (PRINT)	**Signature**	**Date**

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

Person Obtaining Consent (PRINT)	**Signature**	**Date**

EAST CAROLINA UNIVERSITY
University & Medical Center Institutional Review Board Office
4N-70 Brody Medical Sciences Building · Mail Stop 682
600 Moye Boulevard · Greenville, NC 27834
Office **252-744-2914** · Fax **252-744-2284** · **www.ecu.edu/irb**

Notification of Initial Approval: Expedited

From:	Biomedical IRB
To:	Anthony Kulas
CC:	
Date:	8/6/2015
Re:	UMCIRB 15-000127 Effect of previous hamstring strains on anterior cruciate ligament loading

I am pleased to inform you that your Expedited Application was approved. Approval of the study and any consent form(s) is for the period of 8/5/2015 to 8/4/2016. The research study is eligible for review under expedited category #4,6,7. The Chairperson (or designee) deemed this study no more than minimal risk.

Changes to this approved research may not be initiated without UMCIRB review except when necessary to eliminate an apparent immediate hazard to the participant. All unanticipated problems involving risks to participants and others must be promptly reported to the UMCIRB. The investigator must submit a continuing review/closure application to the UMCIRB prior to the date of study expiration. The Investigator must adhere to all reporting requirements for this study.

Approved consent documents with the IRB approval date stamped on the document should be used to consent participants (consent documents with the IRB approval date stamp are found under the Documents tab in the study workspace).

The approval includes the following items:

Name	Description
Hamstring Function Survey	Surveys and Questionnaires
INformed consent	Consent Forms
Recruitment Announcement	Recruitment Documents/Scripts
Study Protocol	Study Protocol or Grant Application

The Chairperson (or designee) does not have a potential for conflict of interest on this study.

Appendix B: Procedure used to Estimate Muscle Volume

Only the VL and BFLH muscles volumes were directly estimated using the CSA integration method outlined in the Methods section. To estimate muscle volumes of the other quadriceps and hamstrings muscles, several studies were pooled together that measured the volumes of these muscles (Akima et al., 2007; Akima et al., 2000; Belavý, Miokovic, Armbrecht, Rittweger, & Felsenberg, 2009; Erskine et al., 2010; Friederich & Brand, 1990; Miokovic, Armbrecht, Felsenberg, & Belavy, 2011; Morse, Degens, & Jones, 2007; Narici, Landoni, & Minetti, 1992; O'Brien, Reeves, Baltzopoulos, Jones, & Maganaris, 2009; Tate, Williams, Barrance, & Buchanan, 2006; Voronov, 2003; Ward et al., 2009; Wickiewicz, Roy, Powell, & Edgerton, 1983). The volumes were markedly different in absolute terms due to different populations in the studies (e.g., cadavers vs. young, healthy adults, males vs. females, etc.) and methods used to determine volume (e.g., MRI, cadaver analysis, etc.), which led to coefficients of variation (CV) of the absolute muscle volumes of 0.47-0.54 for the quadriceps and 0.29-0.34 for the hamstrings. When each muscle's volumes were expressed as percentages of the relative contribution of each muscle volume to the whole muscle group volume, the CVs were lower (0.02-0.17 for the quadriceps and 0.05-21 for the hamstrings). Each of the studies and their muscle volume estimates are presented in Figures 15 and 16 (quadriceps and hamstrings, respectively).

The average (standard deviation) of each of the muscles' volumes (cm^3) using this technique were as follows: BFLH = 181.2 (62.2), BFSH = 94.0 (32.3), ST = 174.5 (59.9), SM = 221.5 (76.1), VL = 628.7 (232.5), VI = 533.5 (197.3), VM = 476.3 (176.2), and RF = 266.7 (98.6). The volumes of the MG and LG, estimated using the regression equations from Miyatani et al., 2004, were 206.5 (63.6) and 113.2 (34.8), respectively.

Quadriceps Volume Estimates												
			Absolute Muscle Volumes					Relative Muscle Volume				
Authors	Subject characteristics	Age	VL	VI	VM	RF	Total	VL	VI	VM	VastiiTot	RF
Narici et al., 1992	6 males	34.0±4.7	5.86	5.74	4.25	2.39	18.23	32.13	31.49	23.29	86.91	13.09
Erskine et al., 2010	17 untrained males	21.3±3.4	677.00	586.00	466.00	345.00	2074.00	32.64	28.25	22.47	83.37	16.63
	17 trained males	21.3±3.4	704.00	609.00	500.00	375.00	2188.00	32.18	27.83	22.85	82.86	17.14
Obrien et al., 2009	10 untrained males	28.2±3.6	691.00	558.00	523.00	280.00	2052.00	33.67	27.19	25.49	86.35	13.65
	10 untrained females	27.4±4.2	456.00	374.00	350.00	179.00	1359.00	33.55	27.52	25.75	86.83	13.17
Morse et al., 2007	18 men recreationally active	23.9±3.4	702.00	604.00	468.00	266.00	2040.00	34.41	29.61	22.94	86.96	13.04
Wickiewicz et al., 1983	Cadaver specimen #1	NR	220.18	293.57	235.49	109.30	858.53	25.65	34.19	27.43	87.27	12.73
	Cadaver specimen #2	NR	339.50	98.10	221.55	96.94	756.10	44.90	12.97	29.30	87.18	12.82
	Cadaver specimen #3	NR	135.59	113.84	98.42	60.72	408.57	33.19	27.86	24.09	85.14	14.86
Ward et al., 2009	21 subjects (M:F = 9:12)	83.9	396.95	181.53	252.81	116.79	948.08	41.87	19.15	26.67	87.68	12.32
Tate et al., 2006	6 males	19.4	692.00	538.50	447.50	280.50	1958.50	35.33	27.50	22.85	85.68	14.32
	4 females	18	372.00	330.00	248.50	153.00	1103.50	33.71	29.90	22.52	86.14	13.86
Akima et al., 2007	6 males (training group)	23.3±4.9	458.70	381.80	344.60	198.60	1383.80	33.15	27.59	24.90	85.64	14.35
	6 males (control group)	22.7±3.9	552.40	464.80	423.00	274.90	1715.20	32.21	27.10	24.66	83.97	16.03
Akima et al., 2000	5 males (training group)	24.0±4.7	706.70	594.60	477.20	294.50	2073.00	34.09	28.68	23.02	85.79	14.21
	4 males (control group)	19.5±1.7	621.80	546.00	453.00	259.80	1899.30	32.74	28.75	23.85	85.34	13.68
Friederick and Brand, 1990	Cadaver Specimen #1	37	514.00	606.00	555.00	238.00	1913.00	26.87	31.68	29.01	87.56	12.44
	Cadaver Specimen #2	63	133.00	135.00	123.00	60.00	451.00	29.49	29.93	27.27	86.70	13.30
		Grand Mean	465.48	390.03	343.96	199.47	1399.99	33.43	27.62	24.91	85.96	13.98
		Grand SD	226.55	207.47	162.07	107.08	689.47	4.43	4.72	2.23	1.41	1.40
		Grand CV	0.49	0.53	0.47	0.54	0.49	0.13	0.17	0.09	0.02	0.10
						Percentages Used:		33	28	25	86	14

FIGURE 15: Absolute and relative muscle volumes of the quadriceps across several studies with the means, standard deviations (SD), and coefficients of variation (CV) reported. The percentages used in estimating the other quadriceps muscle volumes are listed at the bottom.

Hamstring Volume Estimates												
			Absolute Muscle Volumes					Relative Muscle Volume				
Authors	Subject characteristics	Age	SM	ST	BFLH	BFSH	Total	SM	ST	BFLH	BFSH	BF Total
Voronov et al., 2003	19 athletic males	19-35	243.60	157.10	184.00	89.90	674.60	36.11	23.29	27.28	13.33	40.60
Belavy et al., 2009	10 males (controls at baseline)	33.4	273.60	250.10	232.80	123.70	880.20	31.08	28.41	26.45	14.05	40.50
	10 males (controls at 56wks bed rest)	33.4	239.95	224.09	203.70	114.67	782.41	30.67	28.64	26.04	14.66	40.69
	10 subjects (exp. at baseline)	32.6	274.20	240.20	229.00	119.20	862.60	31.79	27.85	26.55	13.82	40.37
Ward et al., 2009	21 subjects (M:F = 9:12)	83.9	126.96	92.64	110.29	56.25	386.14	33.00	24.50	28.56	14.57	43.13
Tate et al., 2006	6 males	19.4	209.50	208.50	202.50	76.00	696.50	30.08	29.94	29.07	10.91	39.99
	4 females	18	135.50	100.50	114.50	35.00	385.50	35.15	26.07	29.70	9.08	38.78
Akima et al., 2007	6 males (training group)	23.3±4.9	180.80	135.90	133.40	70.00	520.10	34.76	26.13	25.65	13.46	39.11
	6 males (control group)	22.7±3.9	208.40	170.70	175.20	84.20	638.50	32.64	26.73	27.44	13.19	40.63
Akima et al., 2000	5 males (training group)	24.0±4.7	248.50	176.70	201.80	84.00	711.00	34.95	24.85	28.38	11.81	40.20
	4 males (control group)	19.5±1.7	234.20	167.00	184.00	89.40	674.60	34.72	24.76	27.28	13.25	40.53
Friederick and Brand, 1990	Cadaver Specimen #1	37	347.00	212.00	217.00	100.00	876.00	39.61	24.20	24.77	11.42	36.19
	Cadaver Specimen #2	63	75.00	45.00	60.00	52.00	232.00	32.33	19.40	25.86	22.41	48.28
Miokovic et al., 2011	7 males (resistance w vibration group)	NR	300.40	220.90	235.10	140.10	896.50	33.51	24.64	26.22	15.63	41.85
	8 males (resistance only group)	NR	292.10	236.10	245.60	134.80	908.60	32.15	25.99	27.03	14.84	41.87
	9 males (control)	NR	256.10	207.70	209.90	128.80	802.50	31.91	25.88	26.16	16.05	42.21
		Grand Mean	227.86	177.82	183.67	93.63	682.98	33.40	25.70	27.03	13.90	40.93
		Grand SD	70.47	59.30	52.85	31.39	206.35	2.41	2.48	1.34	2.90	2.52
		Grand CV	0.31	0.33	0.29	0.34	0.30	0.07	0.10	0.05	0.21	0.06
						Percentages Used:		33	26	27	14	41

FIGURE 16: Absolute and relative muscle volumes of the hamstrings across several studies with the means, standard deviations (SD), and coefficients of variation (CV) reported. The percentages used in estimating the other hamstrings muscle volumes are listed at the bottom.

It has been established that muscle volume and strength are highly correlated (Fukunaga et al., 2001). The estimated muscle volumes used in the present study correlated well with the respective strengths of the muscle groups (i.e., the maximal flexion and extension torques developed on the dynamometer correlated with the hamstrings and quadriceps strength, respectively), thus we felt confident in our estimations of muscle volumes. The correlation between each subject's maximal flexion torque and their total hamstrings volume was significant ($p < 0.05$, $r = 0.959$, as was the correlation between their maximal extension torque and total quadriceps volume ($p < 0.05$. $r = 0.877$). Both of these correlations are shown in Figure 17 below.

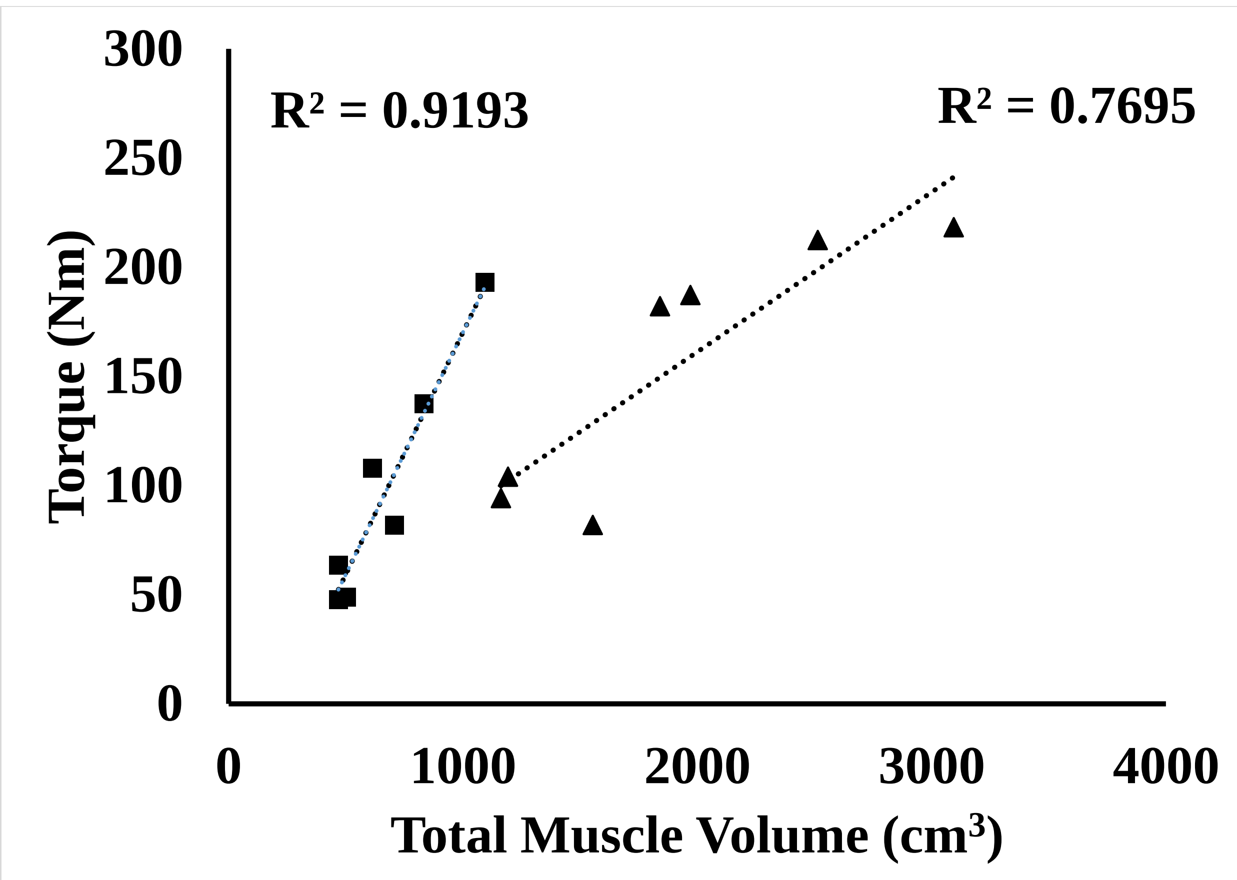

FIGURE 17. Correlations between hamstrings muscle volume and maximal flexion torque produced on the dynamometer (squares), and quadriceps muscle volume and maximal extension torque produced on the dynamometer (triangles).

Appendix C: Correlations of experimental and static optimization EMG

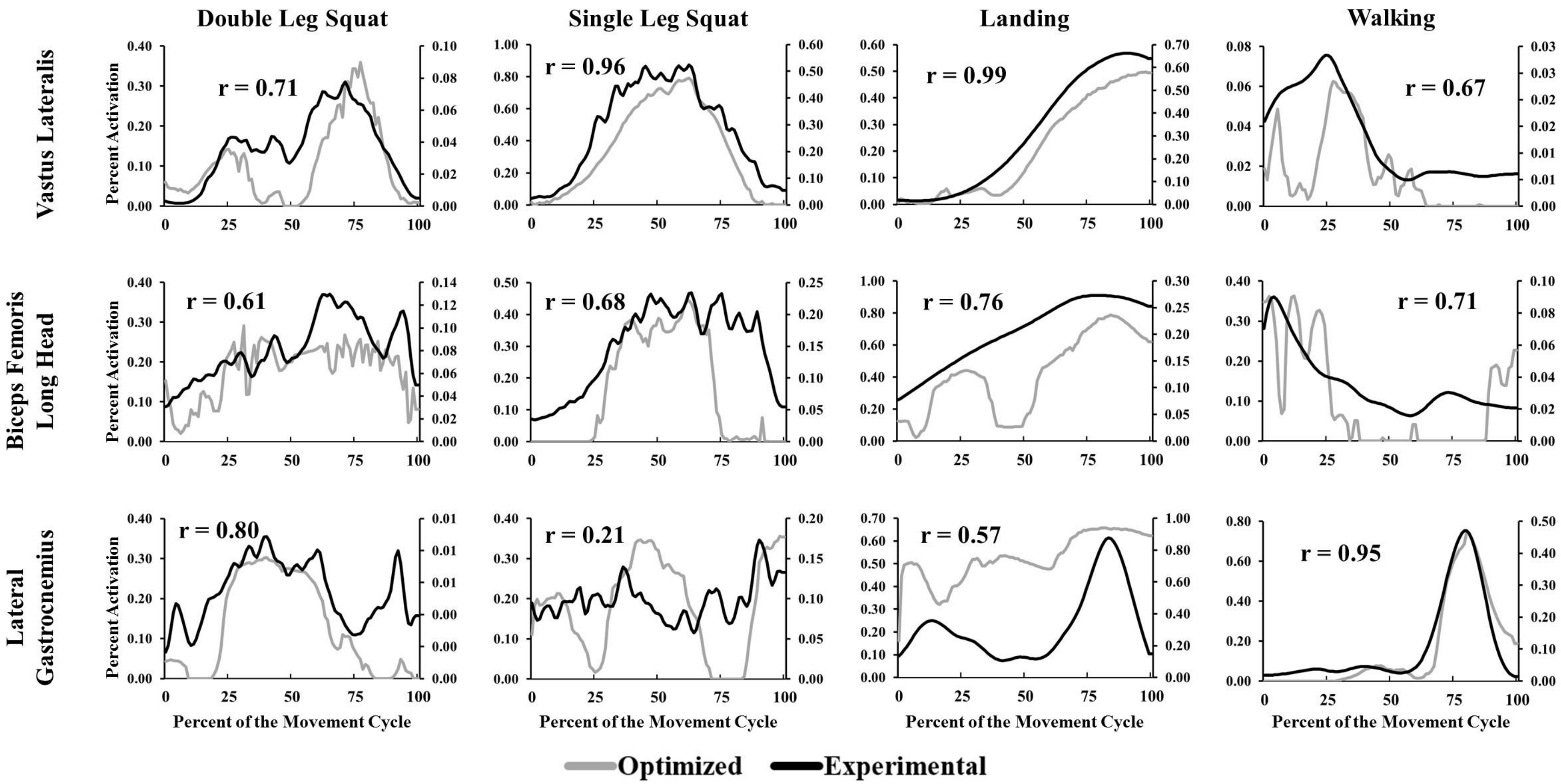

FIGURE 17: Correlations between the experimental EMG and optimized EMG of the VL, BFLH, and LG in each of the tasks analyzed. The vertical axes represent the percentage activation of the muscles (compared to their maximums) with the left vertical axis being optimized EMG, and right axis being experimental EMG. These are ensemble curves of one representative subject's optimized and experimental EMG, but show that the overall shape and pattern of muscle activation was upheld after the static optimization procedure despite changes in the magnitudes of the activations.

Appendix D: Tibial slope sensitivity analysis

For all subjects and for all trials (including the hamstrings and quadriceps +/- 10% F_{max} perturbations), the posterior tibial slope (PTS) was set to 7°. PTS is known to range across different people (Hudek et al., 2009), and PTS has been found to have effects on ACL forces (Marouane et al., 2014; K. B. Shelburne et al., 2011). Increasing PTS (i.e., tilted more posteriorly) causes the ACL to experience higher forces in standing and walking (K. B. Shelburne et al., 2011), and in active gait and in a passive knee joint under compression between 0 and 45° knee flexion (Marouane et al., 2014). For simplicity, 7° was chosen as a representative value for our modeling, but it was important to consider how PTS could be manipulating the ACL forces observed in our tasks and in the perturbation trials. Thus, a sensitivity analysis was performed on one representative subject's normal and perturbed data.

The ACL forces were estimated again after changing the value of the PTS within the ACL model by 3°, to 4° and 10°. This is visualized in Figure 18 which shows one subject's ACL forces across each of the tasks and an average of all tasks, with the PTS at 4° (decreased), 7° (normal), and 10° (increased). Similar to Shelburne et al., 2011 and Marouane et al., 2014, with increasing PTS the ACL experienced higher forces.

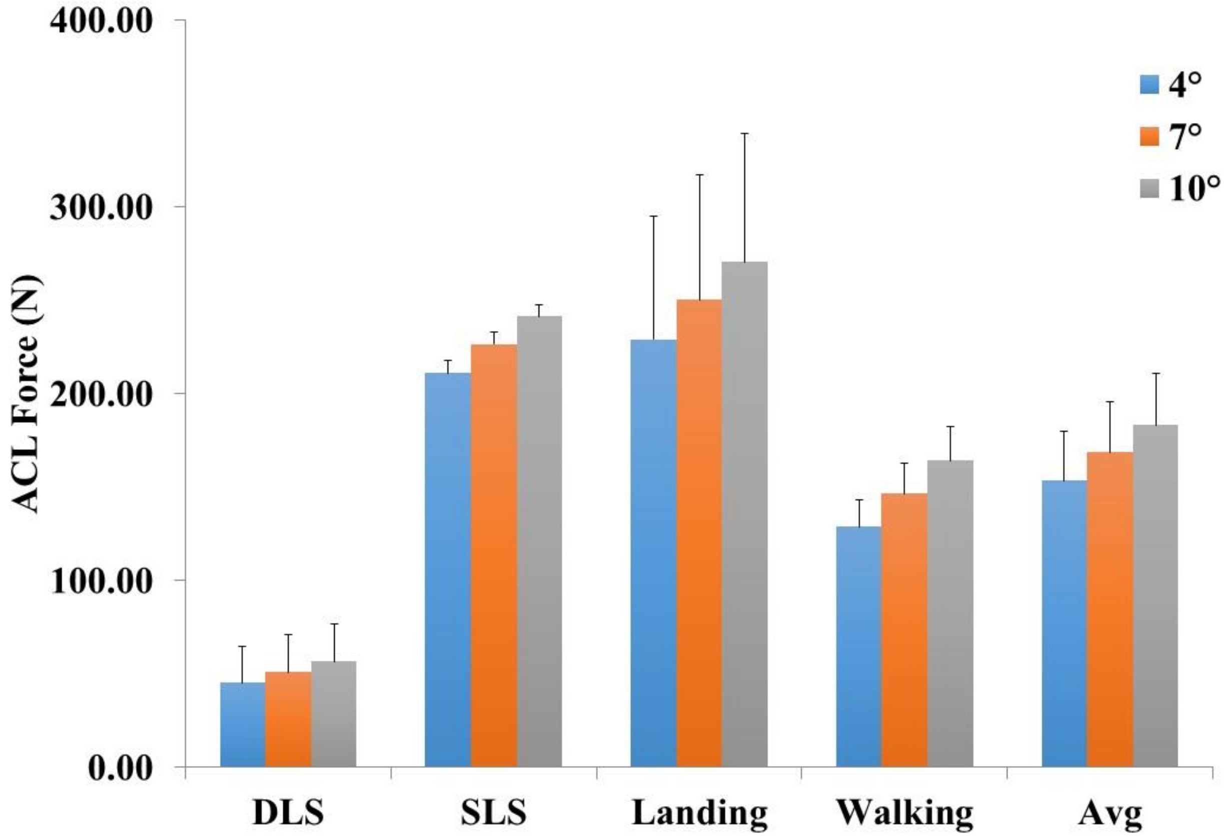

FIGURE 18: ACL forces for one subject across the four tasks when the PTS was decreased to 4° (blue) and increased to 10° (grey) in comparison to the used 7° value (orange).

To depict how significant these changes in PTS were on ACL forces, as well as to see how consistent the changes in PTS were across tasks, the percent changes in ACL forces from the 7° PTS are shown in Figure 19. By decreasing the PTS to 4°, ACL forces decreased by 10.0±2.7%, whereas increasing the PTS to 10° increased ACL forces by 9.8±2.8%. This effect was largest in the double leg squat trials and smallest in the single leg squat trials, but was fairly consistent across all trials of each task type. Because there was not a dependency on task for the effect of the change in PTS, we do not believe that the interpretation of our results would change.

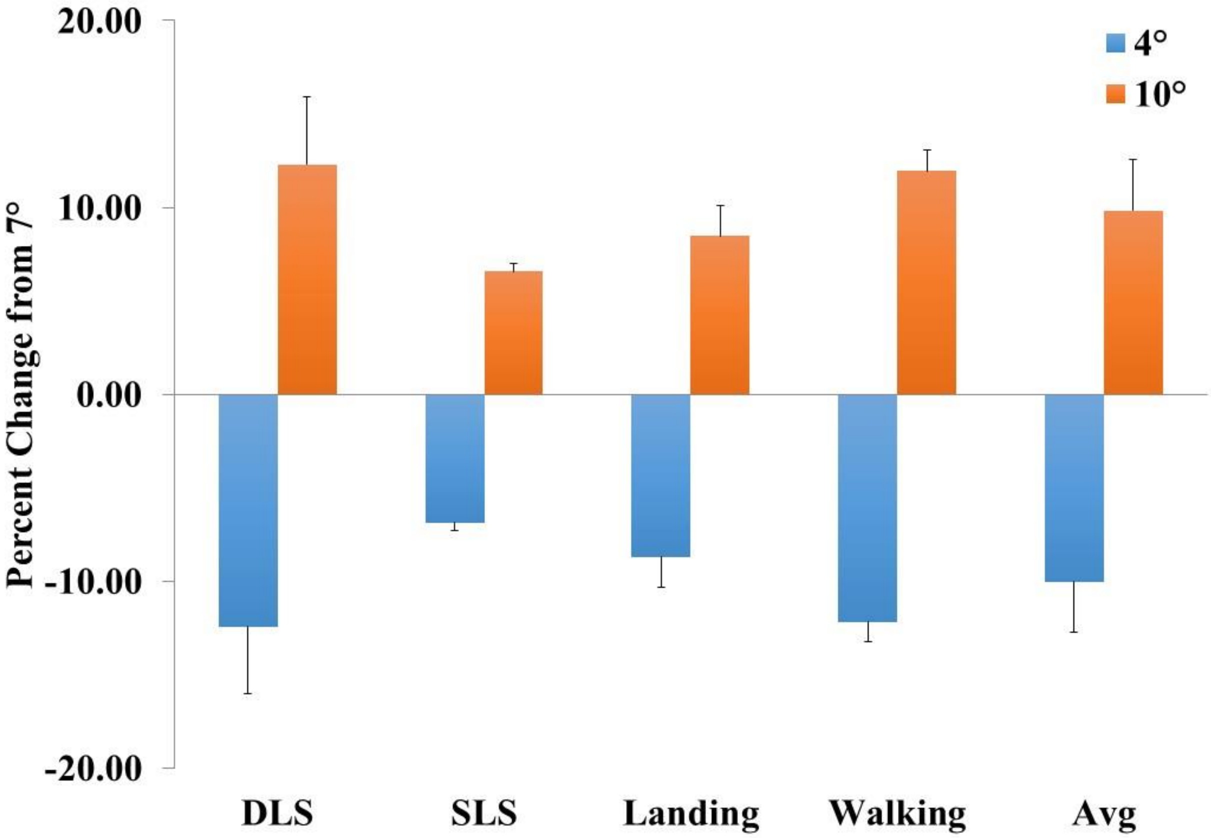

FIGURE 19: The percent changes in ACL forces when the PTS was decreased to 4° (blue) and increased to 10° (orange) from the original 7° value used in one subject.

To ensure that the changes in PTS were similarly affected in the perturbation analyses, a separate but similar sensitivity analysis was done on the perturbation data for the same subject (Figure 20). Again, decreasing the PTS to 4° decreased ACL forces in each of the four perturbations by 10.2±0.34%, and increasing the PTS to 10° increased ACL forces in each perturbation by 10.0±0.34%. The effect was very consistent across each strength perturbation.

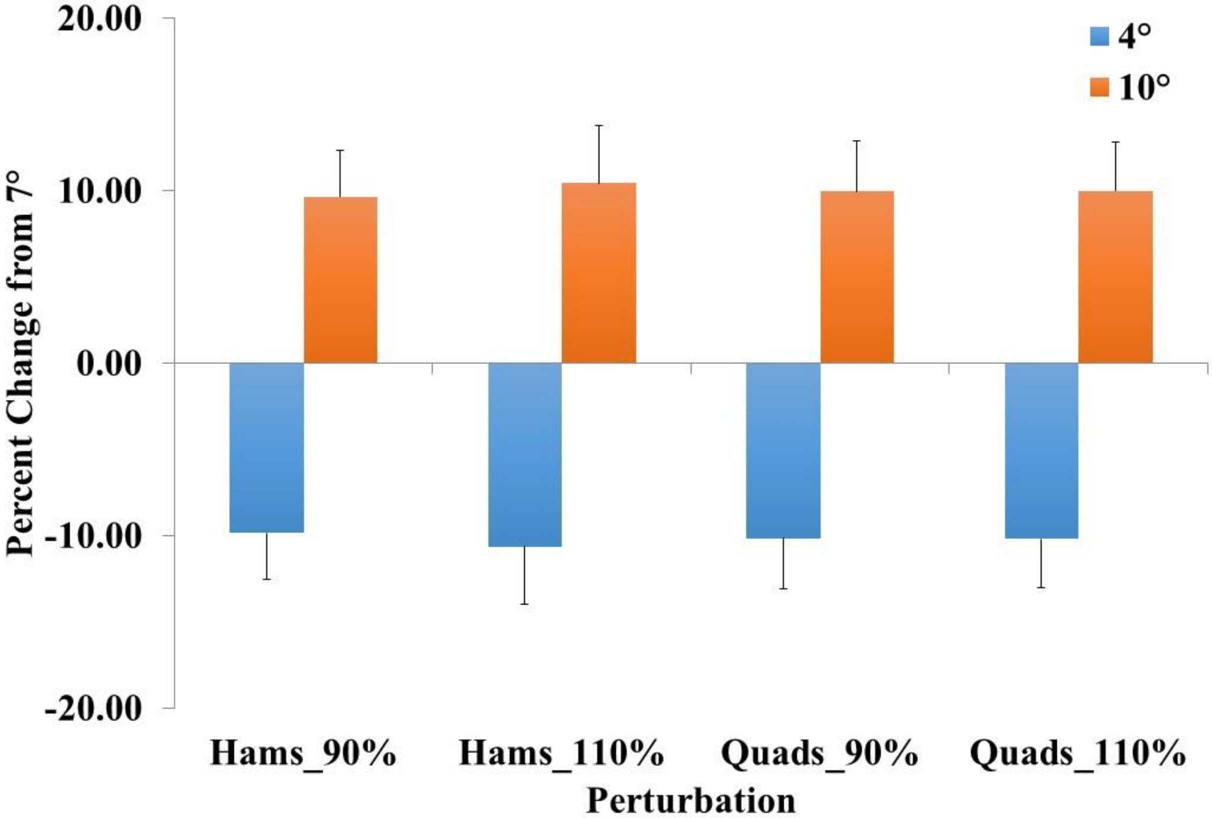

FIGURE 20: The percent changes in ACL forces when PTS was decreased to 4° (blue) and increased to 10° (orange) from the original 7° used across the four different perturbation analyses for one subject.