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This research first examined the validity of net knee joint moment estimations, calculated as the difference between quadriceps and hamstrings torques estimated using either an isometric or angle and action specific sEMG/Torque ratio calculated during calibration actions, during the impact phase of the initial landing of a drop jump maneuver. Second, this research investigated the extent to which the torque impulses of the medial and lateral aspects of the quadriceps and hamstrings, estimated during the impact phase of the initial landing of a drop jump maneuver using an angle and action specific sEMG/Torque ratio predicted knee joint mechanics associated with ACL injury risk, in the three planes of motion. Forty healthy active females, between the ages of 18 and 25, participated in the study. Participants performed maximal calibration actions on an isokinetic dynamometer (eccentric and isometric quadriceps, concentric and isometric hamstrings) while surface electromyographic (sEMG) data were collected from the vastus lateralis, vastus medialis, bicep femoris and semitendinous. Subsequently, a drop jump maneuver was performed while three dimensional biomechanical data as well as sEMG data from the above mentioned muscles were collected. Based on the calibration actions, individualized isometric as well as angle and action specific (eccentric quadriceps, concentric hamstrings) sEMG/Torque ratios (sEMG amplitude divided by half of the torque produced) were computed for each of the four muscles, from full extension to 90 degrees of knee flexion. Using the knee flexion data during the landing maneuver, the sEMG/Torque ratio was then estimated for the impact phase of the drop jump maneuver. It was then divided by the concurrently acquired sEMG to estimate torques for the four afore mentioned muscles during the impact phase of landing. Muscle torques were resolved into a net joint moment as the difference between the sum of the extensors and flexors, and the impulses were then calculated for each of the muscle torques and for the net joint moments. High risk knee joint mechanics, in the three planes of motion, were observed during the impact phase of the initial landing of the drop jump. A RMANOVA tested differences between the net joint moments estimated based on isometric or angle and action specific measurements and inverse dynamics analysis. Regression models assessed the extent to which the muscle torque impulses, estimated using the angle and action specific sEMG/Torque ratio during the impact phase of the initial landing of a drop jump maneuver, predicted each of the seven variables identified as high risk knee joint mechanics. First, the results revealed that net knee joint moment based on the angle and action specific sEMG/Torque ratio provided a closer estimation of the net knee joint moment calculated using an inverse dynamics analysis than the net knee joint moment based on the isometric sEMG/Torque ratio. Second, muscle torque impulses, estimated using the angle and action specific sEMG/Torque ratio, were significantly predictive of only frontal and transverse moments about the knee. Secondary analyses revealed that when including simple ground contact kinematic variables and impact phase duration into the regression models, muscle torques predictivity of high risk knee joint biomechanics often increased. Hence, it was concluded that the angle and action specific sEMG/Torque ratio provides a better estimation of sagittal joint moments than the traditional isometric approach to sEMG normalization. Future studies should investigate the factors influencing ground contact knee joint kinematics and impact phase duration during the initial landing of a drop jump maneuver.

# ESTIMATION OF MUSCLE TORQUE IMPULSES AND ABILITY TO PREDICT HIGH-RISK KNEE JOINT MECHANICS DURING LANDING MANEUVERS

By

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Approved by

______________________________
Committee Chair

To my wife Magali,

For your unconditional love.

To my Parents,

For your support through all these years.

# APPROVAL PAGE

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# CHAPTER I

# INTRODUCTION

Anterior Cruciate Ligament (ACL) injury is described as the "largest single problem in orthopedic sport medicine" (Renstrom, Ljungqvist et al. 2008), not only because of an incidence of approximately 200,000 annually in the US (Griffin, Albohm et al. 2006), but also because of short and long-term health consequences (Noyes, Mooar et al. 1989; Ferretti, Conteduca et al. 1991; Deacon, Bennell et al. 1997; von Porat, Roos et al. 2004). However, such injuries may be preventable to a degree, as the majority (70 to 80%) of the cases are the result of a non-contact mechanism (Boden, Dean et al. 2000). There has been documented success of ACL prevention programs (Hewett, Lindenfeld et al. 1999; Mandelbaum, Silvers et al. 2005) but ACL injuries continue to occur while the specific risk factors are poorly understood. Therefore, a thorough understanding of the mechanisms of ACL injury is necessary for more focused programs to be implemented (Renstrom, Ljungqvist et al. 2008; McLean, Lucey et al. 2010).

ACL injuries are commonly observed during deceleration maneuvers (McNair, Marshall et al. 1990; Ferretti, Papandrea et al. 1992; Boden, Dean et al. 2000; Fauno and Wulff Jakobsen 2006) and therefore are more common in sports like soccer, basketball and volleyball where this type of maneuver is often performed. Non-contact ACL injury is generally estimated to occur shortly (17 to 50 ms) after the foot hits the ground (Boden, Dean et al. 2000; Krosshaug, Nakamae et al. 2007; Koga, Nakamae et al. 2010) and to result from a multiplanar loading mechanism (Shimokochi and Shultz 2008; Quatman, Quatman-Yates et al. 2010). ACL load begins to increase during the flight phase, likely a result of quadriceps activation (Torzilli, Deng et al. 1994; Cerulli, Benoit et al. 2003) and is thought to culminate in actual rupture as a result of various combinations of shallow knee flexion, valgus rotation, tibial internal rotation and anterior tibial translation as evidenced by video observations of actual injury events (McNair, Marshall et al. 1990; Boden, Dean et al. 2000; Olsen, Myklebust et al. 2004; Hewett, Myer et al. 2005).

This description of the ACL injury mechanism is supported by evidence of in-vitro studies demonstrating that ACL loading or even rupture occurs as a result of the afore mentioned knee joint mechanics (Meyer and Haut 2005; Withrow, Huston et al. 2006; Lo, Muller et al. 2008; Meyer and Haut 2008; Withrow, Huston et al. 2008). In a cadaveric setup, adding 50 N of anterior tibial shear force during weightbearing knee flexion increased ACL load (54 N) compared to body weight only (33 N) (Lo, Muller et al. 2008). In another cadaveric model, Meyer et al. (2008) found that isolated internal rotation knee torques (33±13 Nm) resulted in ACL rupture in all the knees tested. Finally, ACL strain was increased by 30 % when an impulsive axial knee loading was applied to create a knee abduction moment rather than a pure flexion moment (Withrow, Huston et al. 2006). Therefore, in agreement with retrospective findings of ACL injury mechanism, valgus and internal tibial rotation excursions, anterior tibial translation as well as their causative forces can be considered high-risk mechanics, particularly when they occur within the first 50 ms after ground contact (Hewett, Myer et al. 2005; Krosshaug, Nakamae et al. 2007).

This body of evidence does not inform the respective contribution of external and internal forces to those high-risk mechanics. In-vivo and in-vitro studies have sought a better understanding of this issue and found that quadriceps torques, through the knee extensor mechanism, are inherently central to the maintenance of joint integrity and center of mass control during deceleration maneuvers (Withrow, Huston et al. 2006). However, they may also play an important role in the injury mechanism (Griffin, Albohm et al. 2006). To produce an extensor moment, quadriceps forces, transmitted to the tibia through the patellar tendon, increase shear and compressive forces in the three planes of motion. Quadriceps forces have been reported to induce anterior tibial translation (Hirokawa, Solomonow et al. 1992; Torzilli, Deng et al. 1994; MacWilliams, Wilson et al. 1999; Kwak, Ahmad et al. 2000; DeMorat, Weinhold et al. 2004), internal tibial rotation (Reuben, Rovick et al. 1989; MacWilliams, Wilson et al. 1999; Kwak, Ahmad et al. 2000; DeMorat, Weinhold et al. 2004) and valgus excursion (Kwak, Ahmad et al. 2000; DeMorat, Weinhold et al. 2004) partly resulting from the interaction of the compressive and shear components of the quadriceps force vector with the posterior tibial slope (Dejour and Bonnin 1994; Li, Rudy et al. 1998) and differences in medial to lateral menisci geometry (Blankevoort and Huiskes 1996; Vedi, Williams et al. 1999; Meyer and Haut 2008; Stijak, Herzog et al. 2008). Similarly, the application of impulsive axial forces results in anterior tibial translation as well as valgus and internal tibial rotations (Blankevoort and Huiskes 1996; Meyer and Haut 2005; Meyer and Haut 2008). With limited muscular contributions, physiological levels of axial forces can strain (Withrow, Huston et al. 2008) or rupture the ACL (Meyer and Haut 2005). The hamstrings, with a dual insertion on the posterior aspect of the lower leg act antagonistically to the motions and forces induced by the quadriceps by controlling anterior tibial translation (MacWilliams, Wilson et al. 1999; Kwak, Ahmad et al. 2000; Mesfar and Shirazi-Adl 2006) and maintaining neutral alignment in the frontal plane (MacWilliams, Wilson et al. 1999; Kwak, Ahmad et al. 2000; Lloyd and Buchanan 2001). Collectively, evidence in-vitro suggests that knee joint mechanics are influenced by thigh muscle forces and external forces (Kernozek and Ragan 2008) and how those interact with the posterior- inferior tibial slope (Dejour and Bonnin 1994; Li, Rudy et al. 1998) and differences in medial and lateral tibial posterior-inferior slopes (Blankevoort and Huiskes 1996; Meyer and Haut 2008; Stijak, Herzog et al. 2008).

Specific to sagittal plane control, in-vitro evidence suggests that the balance of quadriceps and hamstrings forces largely determines mechanical outcomes at the knee (Withrow, Huston et al. 2008). Compressive forces are increased with quadriceps and hamstrings co-contraction (MacWilliams, Wilson et al. 1999), and the net knee moment is a result of the balance of flexor and extensor torques amplitudes (Elftman 1966; Hof, Pronk et al. 1987; Lloyd and Buchanan 2001). Further in-vitro evidence suggests that quadriceps (Torzilli, Deng et al. 1994; Markolf, Burchfield et al. 1995) and ground reaction forces (Meyer and Haut 2005) increase anterior tibial translation and that hamstrings forces decrease it (Markolf, Burchfield et al. 1995).

In-vivo however, the collective contribution of hamstrings and quadriceps to knee joint mechanics is less investigated. When assessing the contribution of the thigh muscles to knee joint mechanics during a deceleration maneuver, Shultz et al. (2009) reported that the hamstrings and quadriceps activity after ground contact together with their isometric strength were not significant predictors of net knee joint extensor moment or of knee flexion excursion. Quadriceps activation explained only 7.3 % of the variance in anterior shear forces determined via an inverse dynamics analysis after accounting for sex, hip and knee excursions as well as knee extension moment (Shultz, Nguyen et al. 2009). These findings (Shultz, Nguyen et al. 2009) are difficult to reconcile with in-vitro findings that thigh muscles forces are a major contributor to knee joint mechanics (Markolf, Burchfield et al. 1995; Withrow, Huston et al. 2008).

This naturally leads to the examination of surface electromyography (sEMG), a method that assesses the electrical activity of the muscles of interest as an alternative to the, almost impossible, direct measurement of muscle forces (Merletti, Rainoldi et al. 2001). For inter-individual or inter-muscle comparisons to be possible the sEMG signal has to be normalized (Dubo, Peat et al. 1976) as raw values are highly variable between subjects and muscles (Wojtys and Huston 1994; Huston and Wojtys 1996; Hewett, Lindenfeld et al. 1999). The isometric method of sEMG normalization, commonly used throughout the literature, expresses the amplitude of the sEMG signal observed during dynamic function as a percent of the peak value of the same sEMG signal observed during a maximal voluntary isometric action (Dubo, Peat et al. 1976). Its use is based on the assumption that the normalized sEMG values are representative of the forces produced by the muscle. In turn, this requires that the relation between sEMG and torque (the isometric sEMG/Torque ratio) is identical across action velocities and joint angles. This appears to be a limitation of using the isometric method and the resultant isometric sEMG/Torque ratio since during dynamic actions the torque generating capability of a muscle, for a constant level of activation, is a function of both angular velocity and joint angle (Lunnen, Yack et al. 1981; Aagaard, Simonsen et al. 1994; Aagaard, Simonsen et al. 2000; Yeadon, King et al. 2006). The explanation for this dependency is that the force that sarcomeres can produce is influenced by their overlap (joint angle), as well as the velocity of their displacement (angular velocity) (Brown, Scott et al. 1996). Using the "isometric method" to normalize sEMG, and making the assumption that the isometric sEMG/Torque ratio can be used to describe the torque generating capability of a muscle during dynamic function, may therefore have limited relevance during dynamic actions (Dubo, Peat et al. 1976). This may explain the afore-mentioned discrepancy between in-vivo and in-vitro findings regarding the contribution of thigh muscles to knee joint mechanics.

An alternative method to normalize the sEMG signal may address the above mentioned limitations. The angle and action specific sEMG/Torque ratio is based on calibration actions during which sEMG and torque data are collected. The ratio between the amplitude of the sEMG observed and the torque produced can be modeled as a function of joint angle and joint angular velocity (Doorenbosch and Harlaar 2003). Doorenbosch and Harlaar (2004) report good to excellent internal validity of estimating quadriceps and hamstrings torques based on the angle and action specific sEMG/Torque ratio, knee joint angle and sEMG collected during dynamic actions. Actual (as measured by the isokinetic dynamometer) and estimated (using the angle and action specific sEMG/Torque ratio, knee joint angle and sEMG) torques were compared with the resultant small differences (11 to 20% of the peak in actual torques) lending evidence to its clinical relevance to estimate muscle torques during simple movements (Doorenbosch and Harlaar 2004). In this study, there was a 13.8% difference between the net knee joint moments calculated based on quadriceps and hamstrings torques estimated using the angle and action specific sEMG/Torque ratio as previously described, and that calculated through an inverse dynamics analysis (Doorenbosch and Harlaar 2003). Furthermore, the same research group also provides evidence that this method is useful to estimate antagonistic contribution of the quadriceps and hamstrings to net knee joint moment during functional activity, more specifically, during the push off phase of a jump (Doorenbosch and Harlaar 2003). Based on this data, the hamstrings to quadriceps co-contraction was higher in ACL deficient compared to healthy individuals (Doorenbosch and Harlaar 2003). The relevance of the previously presented finding is two-fold, first it suggests that using the angle and action specific sEMG/Torque ratio allows determination of the individual hamstrings and quadriceps contributions to net knee joint moments and second it appears to be valid in distinguishing between clinically relevant populations of healthy and ACL deficient individuals.

Despite the apparent relevance of using the angle and action specific sEMG/Torque ratio to estimate quadriceps and hamstrings muscle torques during dynamic function, we have been unable to locate studies using this method to observe the contribution of quadriceps and hamstrings torques to knee joint mechanics during deceleration maneuvers. Together with three dimensional knee joint mechanics, the afore-mentioned angle and action specific sEMG/Torque ratio may allow further insight into the contribution of thigh muscle torques to high-risk knee joint mechanics.

## Statement of the Problem

Non-contact ACL injury likely occurs during the first 50-70 ms of a deceleration maneuver (Boden, Dean et al. 2000; Krosshaug, Nakamae et al. 2007; Koga, Nakamae et al. 2010) as a result of anterior tibial translation, valgus and internal tibial rotation. These high-risk knee joint mechanics are influenced by the interaction of ground reaction, hamstrings and quadriceps forces (Withrow, Huston et al. 2008), and joint geometry (Meyer and Haut 2005; Meyer and Haut 2008). Evidence in-vitro strongly suggests that quadriceps and hamstrings are ACL antagonists and agonists, respectively (Wojtys and Huston 1994; Huston and Wojtys 1996; Hewett, Lindenfeld et al. 1999), and that through appropriate co-contraction can largely limit the occurrence of anterior tibial translation (Kwak, Ahmad et al. 2000), valgus (Lloyd and Buchanan 2001) and internal tibial rotation (MacWilliams, Wilson et al. 1999; Markolf, O'Neill et al. 2004).

However, the contribution of quadriceps and hamstrings to high-risk knee joint mechanics, as assessed by normalizing their sEMG using the ‘isometric method’ and relying on the isometric sEMG/Torque ratio, has not been clearly demonstrated in individuals performing deceleration maneuvers in a laboratory environment. The angle and action specific sEMG/Torque ratio is a clinically oriented method that has been shown to efficiently distinguish the co-contraction patterns of ACL deficient and healthy individuals during the push off phase of a jump (Doorenbosch and Harlaar 2003). This approach would also be useful to estimate thigh muscles torques during dynamic function, based on the individualized angle and action specific sEMG/Torque ratio as observed during isokinetic calibration actions. This method may help overcome limitations of the “isometric method” to better understand the mechanical contributions of quadriceps and hamstrings muscles to high-risk knee joint mechanics.

## Objective and Hypotheses

Our objective was twofold; first assess the validity of calculating the net knee joint moment using the angle and action specific sEMG/Torque ratio to estimate muscle torque impulses during the impact phase of the initial deceleration of a drop jump maneuver, and second, to determine the extent to which the vastus lateralis, vastus medialis, bicep femoris and semitendinous torque impulses, estimated using the angle and action specific sEMG/Torque ratio, predicted variables indicative of high-risk knee joint mechanics during the impact phase of the initial deceleration of a drop jump maneuver.

**Hypothesis 1:** During the impact phase of the initial deceleration of a drop jump maneuver, the knee joint moment, calculated as the difference between the summed vastus lateralis and vastus medialis and the summed bicep femoris and semitendinous torque impulses estimated using the isometric sEMG/Torque ratio will differ from that calculated based on the angle and action specific sEMG/Torque ratio and from the net knee joint moment calculated through an inverse dynamics analysis.

Specifically,

a) The net sagittal plane joint moment impulses calculated (quadriceps minus hamstrings) based on vastus lateralis, vastus medialis, bicep femoris and semitendinous torque impulse estimated with the isometric sEMG/Torque ratio will be lesser than that estimated using an inverse dynamics analysis or the angle and action specific sEMG/Torque ratio during the impact phase of the initial deceleration of a drop jump maneuver. Additionally, the net sagittal plane joint moment impulses, calculated based on muscle torque values estimated using the angle and action specific sEMG/Torque ratio and inverse dynamics will not differ.

b) The isometric sEMG/Torque ratio will render an estimation of quadriceps torque impulses lower than those estimated through the angle and action specific sEMG/Torque ratio during the impact phase of the initial deceleration of a drop jump maneuver. Also, the isometric sEMG/Torque will render an estimation of hamstrings torque impulses greater than those estimated through the angle and action specific sEMG/Torque ratio during the impact phase of the initial deceleration of a drop jump maneuver.

**Hypothesis 2:** During the impact phase of the initial deceleration of a drop jump maneuver, greater vastus lateralis and vastus medialis and lesser bicep femoris and semitendinous torque impulses, estimated using the angle and action specific sEMG/Torque ratio, will predict greater magnitudes of variables indicative of high-risk knee joint mechanics in the three planes of motion.

Specifically:

In the sagittal plane, greater vastus lateralis and vastus medialis and lesser bicep femoris and semitendinous torque impulses estimated using the angle and action specific sEMG/Torque ratio will predict:

a) Lesser knee flexion excursions

b) Greater peak internal knee extensor moment

c) Greater peak anterior shear forces

In the frontal plane, greater vastus lateralis and vastus medialis and lesser bicep femoris and semitendinous torque impulses estimated using the angle and action specific sEMG/Torque ratio will predict:

d) Greater knee valgus excursions

e) Greater peak internal knee valgus moment

In the transverse plane, greater vastus lateralis and vastus medialis and lesser bicep femoris and semitendinous torque impulses estimated using the angle and action specific sEMG/Torque ratio will predict:

f) Greater internal tibial rotation excursions

g) Greater peak internal knee internal rotation moment

## Limitations and Assumptions

1. Results from this dissertation cannot be generalized to populations other than the college aged females studied, or to tasks other than the double leg deceleration in preparation for a maximal jump.
2. All participants will provide a maximum effort during testing.
3. Surface electromyography is a reliable and valid method of measuring muscle activity during dynamic activity.
4. Surface electromyography obtained over the electrode placements for each muscle is adequately representative of the muscle as a whole.
5. Three dimensional kinematics accurately model the true motions of body segments.
6. Inverse dynamics calculations represent the total moments occurring at the joint and as such include the contribution of both passive and dynamic structures of the joint.
7. The proposed angle and action specific sEMG/Torque ratio model does not account for the change in hip angle during the impact phase of landing.
8. The proposed angle and action specific sEMG/Torque ratio model does not account for all muscles crossing the knee joint.
9. The angle and action specific sEMG/Torque ratio model only relies on a singular velocity of calibration action.

10. This work does not account for other factors, such as anatomical and hormonal, potentially associated with high-risk knee joint mechanics.

## Delimitations

1. Only college-aged female participants who are healthy with no musculoskeletal injury to either lower extremity for the past 6 months and have not had surgery on either lower extremity participated.
2. All measurements were only obtained from the dominant stance leg.
3. Data, results and interpretation were obtained during the impact phase of the initial deceleration of a double legged drop jump maneuver.

## Operational Definitions

1. Impact phase: Period between foot contact (GRF>10N) and peak vertical ground reaction force (F2).
2. Impulse (Nm*s): The area under the estimated net joint moment or muscle torque curve over the impact phase of the initial landing of a drop jump maneuver.
3. Dominant stance leg: Defined as the stance limb when kicking a ball.
4. <u>Isometric sEMG/Torque ratio (mV/Nm)</u>: The ratio of sEMG amplitude (non normalized) to torque produced during isometric calibrations actions on an isokinetic dynamometer.
5. <u>Angle and action specific sEMG/Torque ratio (%maxMVICsEMG/Nm</u>**)**: The angle (between 0 and 100° of knee flexion) specific ratio of sEMG amplitude (normalized to the peak observed during a maximal voluntary isometric action) to torque produced. This ratio was calculated during, eccentric quadriceps (-270° $s^{-1}$) and concentric hamstrings (90° $s^{-1}$) calibration actions on an isokinetic dynamometer.

6. Muscle torque estimated using the Isometric sEMG/Torque ratio (Hypothesis 1) (Nm): Estimation of the torque produced by a muscle over the impact phase of the initial landing of a drop jump maneuver and calculated as the ratio of the unique Isometric sEMG/Torque ratio and the amplitude of the sEMG observed.
7. Muscle torque estimated using the angle and action specific sEMG/Torque ratio (Hypotheses 1 and 2) (Nm): estimation of the torque produced by a muscle over the impact phase of the initial landing of a drop jump maneuver and calculated as the ratio of the action and angle specific sEMG/Torque ratio and the amplitude of the sEMG observed.
8. Muscle Torque Impulse (Nm*s): Integration of the muscle torques estimated using the isometric or angle and action specific sEMG/Torque ratios over the impact phase of the initial landing of a drop jump maneuver.

   Using the ***isometric*** sEMG/Torque ratio this rendered the following variables:

   Vastus lateralis torque impulse *($VL_{ISO}$, Nm*s)*.

   Vastus medialis torque impulse *($VM_{ISO}$, Nm*s)*.

   Bicep femoris torque impulse *($BF_{ISO}$, Nm*s)*.

   Semitendinous torque impulse *($ST_{ISO}$, Nm*s)*.

   Using the **angle and action specific** sEMG/Torque ratio this rendered the following variables:

   Vastus lateralis torque impulse ($VL_{DYN}$, Nm*s).

   Vastus medialis torque impulse ($VM_{DYN}$, Nm*s).

   Bicep femoris torque impulse ($BF_{DYN}$, Nm*s).

   Semitendinous torque impulse ($ST_{DYN}$, Nm*s).

9. Net Knee Joint Moment Impulse (Hypothesis 1): Integration of the internal knee extensor moment, over the impact phase of the initial landing of a drop jump maneuver, and

calculated: Through an inverse dynamics analysis, (KEM, Nm*s) and as the difference between the summed vastus lateralis and medialis and the summed bicep femoris and semitendinous torque impulses estimated using the *isometric sEMG/Torque ratio* ($NET_{ISO}$, Nm*s) or the **angle and action specific sEMG/Torque ratio** ($NET_{DYN}$, Nm*s).

# CHAPTER II

# REVIEW OF LITERATURE

In order to best understand the contribution of quadriceps and hamstrings torque impulses to high-risk knee joint mechanics during the impact phase of the initial deceleration of a drop jump maneuver, a review of the pertinent literature is necessitated. The following chapter will review mechanisms of ACL loading, neuromechanics of deceleration activities, and evidence supporting the adoption of a new method to estimate knee torques.

## Mechanisms of ACL Loading

The knee consists of two major articulations, the patello-femoral joint, including the patella and its contact with the femoral trochlea, and the tibio-femoral joint, made of the femoral condyles and their contact with the tibial plateaus (Martini and Timmons 1995). The motion of the knee joint consists of six degrees of freedom. The three main planar rotations are sagittal plane flexion-extension, frontal plane abduction-adduction and transverse plane internal and external rotation (Martini and Timmons 1995). There are also three translations that occur with these rotations including medial-lateral, anterior-posterior and superior-inferior translations (Martini and Timmons 1995). Given the bony anatomy of the knee, it is necessary for passive and active soft tissue restraints to contribute to joint stability.

Passive restraint of the tibiofemoral joint is provided by the lateral and medial menisci, joint capsule, posterior cruciate, anterior cruciate, lateral and medial collateral ligaments as well as the bony arrangements of the knee (Hughes and Watkins 2006). The anterior cruciate ligament (ACL) is located in the intercondylar notch and plays a primary role in countering anterior shear forces (Butler, Noyes et al. 1980; Arnoczky 1983) and internal tibial rotation (Markolf, Burchfield et al. 1995; Hame, Oakes et al. 2002) and a secondary role in countering valgus forces (Markolf, Burchfield et al. 1995). The ACL becomes more important in restraint during extreme joint positions (e.g. hyperextension (Lin, Lai et al. 2009)) and may reach failure if the load imparted upon it is too high. In the following section we will present the experimental and observational findings that have contributed to our understanding of high-risk knee joint mechanics, or those mechanics that increase ACL load and may lead to its rupture.

### The Injury Event

Retrospective investigations continue to provide important information regarding the events that contribute to ACL load and may result in its rupture (McNair, Marshall et al. 1990) by characterizing the positions and excursions at the knee during actual ACL injury events (Olsen, Myklebust et al. 2004; Krosshaug, Nakamae et al. 2007; Koga, Nakamae et al. 2010). Those studies have characterized the injury mechanism to be linked to specific kinematic occurrences such as shallow knee flexion and valgus collapse (Ireland 1999; Krosshaug and Bahr 2005). Using a video based 3 dimensional reconstruction technique (Krosshaug and Bahr 2005) of 10 female handball players injuring their ACL during a deceleration maneuver, Koga et al. (2010) reports similar findings as previous literature in that the knee is in shallow flexion (Mean: 23°; Range: 11 - 30°) and neutral in the frontal plane at ground contact before injury. They also report kinetic and transverse plane information regarding the injury mechanism. Specifically, they found that the estimated instant of ACL rupture was similar in timing to the estimated peak in ground reaction force (40 ms). They also report that between ground contact and the estimated time of injury, sharp kinematic changes in the three planes of motion occurred (Excursions from instant of contact to 40 ms post; 12° Valgus, 8° Internal tibial rotation, 24° Flexion) (Koga, Nakamae et al. 2010). Collectively this lends evidence towards a multi-planar loading mechanism (Shimokochi and Shultz 2008; Quatman, Quatman-Yates et al. 2010).

The results in Koga et al. (2010) were slightly different from that of previous work of actual ACL injury occurrences, which reported that external, not internal, tibial rotation was part of the injury mechanism (Ireland 1999; Olsen, Myklebust et al. 2004). This dissimilarity may be due to differences in methods used to assess knee joint kinematics during the injury event. Olsen et al. (2004) relied solely on expert visual analysis of video footage, where the small amplitude and rapidity of internal rotation may have gone unnoticed. Since the determination of the time of injury was difficult in those studies, the more visible, external tibial rotation occurring post injury (Meyer and Haut 2005; Koga, Nakamae et al. 2010) may have been wrongly identified as part of the injury mechanism (Koga, Nakamae et al. 2010).

Collectively, retrospective and laboratory studies suggest that impulsive axial loading, shallow knee flexion (Cerulli, Benoit et al. 2003), valgus rotation (Koga, Nakamae et al. 2010), anterior shear force (Fleming, Renstrom et al. 2001) and tibial rotation (Meyer and Haut 2005; Koga, Nakamae et al. 2010) during the impact phase of a deceleration maneuver (Cerulli, Benoit et al. 2003; Koga, Nakamae et al. 2010) can be characterized as high-risk knee joint mechanics.

### Internal and External Influences on ACL Load

The suggestion that the previously mentioned knee joint mechanics are implicated in the ACL injury event necessitates an understanding of in-vitro and in-vivo studies that demonstrate increased ACL load with the aforementioned high-risk knee joint mechanics. It is important to understand how these biomechanical studies have provided supporting evidence to observed knee joint mechanics of the ACL injury occurrence by reproducing the demands of deceleration maneuvers and assessing the resulting ACL load.

### *External Loads*

Impulsive Loading

The contribution of impulsive axial loading to knee joint mechanics has been demonstrated in biomechanical studies closely reproducing knee joint positions as well as forces and rates of impulsive loading observed during deceleration maneuvers. Cadaveric testing observed that when the knee is flexed to 30°, an impulsive axial load of 5.4 ±2.0 kN induced ACL rupture in all of the knees tested (Meyer and Haut 2008). Their design included an incremental increase of the loads applied upon the knee, which allowed assessment of the kinematic changes with the loads just prior to that inducing injury. Results showed that impulsive loading (4.8 kN) induced a 12 ± 5.2 mm posterior femoral displacement relative to the tibia, 3.9 ± 4 ° of internal tibial rotation together with a medial 2.1 ± 4.8 mm femoral displacement relative to the tibia. Mechanistically, these occurrences can be explained by the geometry of the tibial plateaus. The posterior slope (around 10°) of the tibial plateau facilitates the posterior translation of the femur relative to the tibia in the presence of anterior shear force and partly explains the contribution of impulsive axial loading to anterior tibial translation (Dejour and Bonnin 1994; Li, Rudy et al. 1998). Another important factor explaining the contribution of axial impulsive loading to knee joint mechanics is the greater lateral, compared to medial, tibial plateau slope that induces tibial internal and valgus rotations as a coupled motion (Blankevoort and Huiskes 1996; Meyer and Haut 2008; Stijak, Herzog et al. 2008)

Internal Rotation Moment

Other work has also examined the contribution of internal rotation moments to high-risk knee joint mechanics with in-vivo and in-vitro studies applying transverse plane torques to understand how such loads affect ACL load and knee joint kinematics (Fleming, Renstrom et al. 2001; Meyer and Haut 2008). Using young and active subjects undergoing arthroscopic surgery, a variety of external loads were applied to the knee, in 20° of flexion, to assess their influence on ACL strain (Fleming, Renstrom et al. 2001). Internal rotation moments of 10 Nm were reported to increase ACL strain independent of the application of a weightbearing load (~3% strain) (Fleming, Renstrom et al. 2001). However, 10 Nm internal rotation torque may not necessarily be representative of physiological values during deceleration maneuvers (Venesky, Docherty et al. 2006). This issue was addressed in a cadaveric setup simulating weightbearing in 30° of knee flexion where internal rotation moments (33 ± 13 Nm) were found to result in ACL rupture in all of the (7) knees tested (Meyer and Haut 2008). Since they incrementally increased the moments applied, they could also observe the effect of internal tibial rotation moments when injury does not occur. Internal rotation torques of 31 ± 9.4 Nm induced a large internal tibial rotation (45 ± 18 °), but also 11 ± 6 ° of valgus rotation and 9 ± 3.3 mm of anterior tibial displacement. These findings show that physiologically relevant internal tibial rotation torques can rupture the ACL and that such loads are also associated with other high-risk knee joint mechanics, thereby potentially contributing to the ACL injury mechanism.

External Valgus Moments

External valgus moments have been reported to prospectively predict ACL injury (Hewett, Myer et al. 2005) and as such it is important to know how they may contribute to ACL loading. In the same setup as previously described in the operating room, Fleming et al. (2001) reported that 15 Nm valgus torques increased ACL load (~2 % strain) when the knee was in 20° of flexion and weightbearing. However, a 15 Nm valgus torque may be less than the loads encountered during deceleration maneuvers (Shin, Chaudhari et al. 2009). To address this issue, a simulation model of ACL strain, re-created physiologically relevant loads matched to represent those actually experienced by individuals during a sub maximal single leg deceleration maneuver (Shin, Chaudhari et al. 2009). The output from the model suggests that ACL strain increases with increased valgus moment (51 Nm of external valgus moment, strain: 7.6 %), but that the highest ACL strain observed was still below failure levels previously described (9 - 15 %) (Butler, Guan et al. 1992; Momersteeg, Blankevoort et al. 1995). Although isolated valgus loading has a moderate influence on ACL loading, its importance in a multiplanar mechanism study should not be disregarded.

***Internal Loads***

The above evidence demonstrates that impulsive axial loading as well as valgus and internal rotations increase ACL load and may lead to its rupture. However, the aforementioned studies only assess the effects of external loadings on ACL. Therefore, they do not inform the changes in knee joint mechanics that occur as a result of the internal forces exerted by the muscles crossing the knee joint. Thus, an understanding of the dynamic restraint system is warranted in a study related to the mechanical influence of thigh muscles to high-risk knee joint mechanics. The dynamic stabilizers of the knee include all the muscles crossing the joint, the tensor fasciae latae, the gastrocnemius, gracillis, hamstrings, semitendinosus, semimembranosus, biceps femoris and quadriceps rectus femoris, vastus lateralis and vastus medialis (Lloyd and Buchanan 2001). The quadriceps and hamstrings have the largest influence on knee joint mechanics, not only because they both are large muscles, with large torque producing capabilities, but also because they have moment arms that can support varus/valgus motions and moments (Lloyd and Buchanan 2001). Thus the next section will be limited to these two main muscle groups.

Quadriceps Forces

In-vitro studies have sought a better understanding of the contribution of quadriceps forces to the injury mechanism, by applying physiological loads on the quadriceps tendon and observing the resultant ACL load and/or knee joint kinematics. The quadriceps forces are transmitted to the tibia via the patellar tendon (Lloyd and Buchanan 2001) and the extent to which those forces result in knee extension torque is influenced by the change of the extension moment arm as a function of knee joint angle with the largest knee extensor moment arm typically in 30 ° of flexion (Grood, Suntay et al. 1984). The infrapatellar tendon force vector (Nunley, Wright et al. 2003) is defined as the angle between the patellar tendon and the longitudinal axis of the tibia and largely determines the contribution of quadriceps forces to compressive and shear forces during dynamic function (Nunley, Wright et al. 2003). When the knee is in shallow flexion, the patellar tendon/tibial shaft angle is largest (Nunley, Wright et al. 2003) and therefore quadriceps forces have the largest effect on anterior tibial translation and subsequent ACL load because of the importance of the horizontal component of their vector of force (Grood, Suntay et al. 1984; Li, Rudy et al. 1998).

Frontal Plane

The influence of the quadriceps on frontal plane knee joint motion is determined by the interaction of patella tendon forces and knee joint geometry as well as the balance of forces created by the different components of the quadriceps. Vastus medialis and lateralis are thought to have internal valgus moment arms when acting about the medial tibiofemoral articulation and internal varus moment arms when acting about the lateral tibiofemoral articulation when the knee is close to full extension (Buchanan, Kim et al. 1996). In vivo evidence suggest that when the knee is in 60 ° of flexion both the medial and lateral components of the quadriceps results in valgus moments at the knee (Zhang, Wang et al. 2003). Furthermore, applied patella tendon forces may affect frontal plane knee joint kinematics to a different extent when transferred to the medial and lateral aspects of the tibiofemoral joint. This is due to known differences in medial and lateral tibial plateaus geometry (Blankevoort and Huiskes 1996; Vedi, Williams et al. 1999; Meyer and Haut 2008; Stijak, Herzog et al. 2008). More specifically, the more concave and deeper medial tibial meniscus is more stable than the lateral one, facilitating valgus as a motion coupled with anterior tibial translation (Vedi, Williams et al. 1999). As such valgus may occur as a result of the anterior tibial translation and the differences in lateral and medial tibial condyles geometry (DeMorat, Weinhold et al. 2004). This thought is supported by reports of quadriceps forces inducing a knee valgus rotation, or a lateral tibial displacement most pronounced near full extension (Li, Rudy et al. 1998; DeMorat, Weinhold et al. 2004). Using a cadaveric setup of the knee, fixed in 20 ° of flexion and submitted to large (4500 N), unopposed, quadriceps forces, the authors report that a 2.3 ° valgus rotation occurred (DeMorat, Weinhold et al. 2004). However, the use of large forces in a non weightbearing, fixed knee joint angle setup limits the conclusions that can be drawn regarding the external validity of the results. Looking at the same issue with an open kinetic chain, in-vitro, set up and much lesser quadriceps forces (200 N) at knee angles between 30 and 90 ° of knee flexion, Li et al. (1998) reported that tibial lateral displacement was most pronounced at 30 ° of knee flexion. Further clarification of the mechanism leading to these observations came from another in-vivo experiment where each of the quadriceps heads were individually stimulated and their contribution to knee joint moments assessed (Zhang, Wang et al. 2003). The authors report that the vasti and rectus femoris respectively created valgus and varus moments about the knee joint (Zhang, Wang et al. 2003). The resulting moments, of the complete quadriceps group, in the frontal planes are therefore differentially influenced by rectus femoris and the vasti (Zhang, Wang et al. 2003). To summarize the findings in the frontal plane, there is initial evidence that the vasti have similar moment arms directed towards neutral alignment in the frontal plane. Their forces can, especially when the knee is close to full extension, induce valgus motion because of the differences in geometry between the lateral and medial menisci (Li, Rudy et al. 1998; DeMorat, Weinhold et al. 2004).

Transverse Plane

The quadriceps also play a role in transverse plane motion of the tibiofemoral joint as studied mostly in-vitro (Hirokawa, Solomonow et al. 1992; MacWilliams, Wilson et al. 1999; Li, DeFrate et al. 2004). Independent of whether the cadaveric model included weightbearing, unopposed quadriceps forces applied to a knee induces internal tibial rotation (Hirokawa, Solomonow et al. 1992; MacWilliams, Wilson et al. 1999; Kwak, Ahmad et al. 2000). The extent to which quadriceps forces induce internal tibial rotation is most pronounced when the knee is close to full extension, at 15 to 30 ° of flexion, 118 N of quadriceps force induced 6 to 7 ° of internal tibial rotation, whereas at 60 ° the same force resulted in only 2 to 3 ° of internal rotation (Hirokawa, Solomonow et al. 1992).

Mechanistically, the fact that quadriceps action induces internal rotation can be explained by in-vivo findings. By separately eliciting the quadriceps heads Zhang et al. (2003) found that all quadriceps components induced internal tibial rotation moments. In the transverse plane, there is collective evidence of the quadriceps (Zhang, Wang et al. 2003) and weightbearing contributing to internal tibial rotation (Meyer and Haut 2005).

Hamstrings Forces

The quadriceps are central to maintenance of postural stability during deceleration maneuvers (Withrow, Huston et al. 2006), however, as described above, they are also thought to play a role in high risk knee joint mechanics (Griffin, Albohm et al. 2006). To fully understand active restraint mechanics acting on the knee joint, hamstrings forces, considered antagonistic to those exerted by the quadriceps (Baratta, Solomonow et al. 1988), must be described. The hamstrings have a dual insertion on the posterior aspect of the lower leg, on the medial surface of the tibia for the semitendinosus and semimembranosus and on the lateral side of the head of the fibula for bicep femoris (Lloyd and Buchanan 2001).

In the sagittal plane, the capability of the hamstrings to create a flexion moment is largely angle dependent, with peak torques occurring near full extension (Anderson, Madigan et al. 2007). Knee angle also affects posterior translation of the tibia relative to the femur that result from hamstrings action (MacWilliams, Wilson et al. 1999; Kwak, Ahmad et al. 2000; Mesfar and Shirazi-Adl 2006) as the hamstrings tendons become more parallel to the tibial plateau with increasing knee flexion (Pandy and Shelburne 1997) and therefore become more efficient at producing posterior shear force (Mesfar and Shirazi-Adl 2006).

In the frontal plane, the bicep femoris has a large valgus moment arm on the medial condyle and a flexion moment arm on the lateral condyle; on the other hand the semitendinosus has a moment arm directed towards flexion on the medial condyle and a varus moment arm on the lateral condyle (Lloyd and Buchanan 2001). This suggests that in the frontal plane the combined action of the hamstrings (semitendinosus and biceps femoris) is geared towards the maintenance of neutral alignment. However, in a cadaveric model investigating the kinematics resulting from applied hamstrings forces, a 267 N hamstrings force induced a 4 to 6 ° varus rotation between 0 and 30° of knee flexion (Kwak, Ahmad et al. 2000). In the same study the hamstrings force induced 8° external rotation of the tibia (Kwak, Ahmad et al. 2000). There is limited evidence of the isolated influence of hamstrings forces on knee joint mechanics because the purpose of most set ups is to assess the extent to which the hamstrings can decrease the ACL load or kinematic changes induced by quadriceps forces. However, reports show that hamstrings forces counter out of plane mechanics that may result from quadriceps forces such as valgus (MacWilliams, Wilson et al. 1999; Kwak, Ahmad et al. 2000) internal tibial rotation (Kwak, Ahmad et al. 2000) and anterior shear forces (MacWilliams, Wilson et al. 1999; Kwak, Ahmad et al. 2000; Mesfar and Shirazi-Adl 2006) and these findings provide strong support to the notion that the hamstrings act as agonists to the ACL (Baratta, Solomonow et al. 1988; More, Karras et al. 1993).

It is also important to understand the effect of hamstrings and quadriceps co-contraction when external forces are applied to the knee. Hamstrings and quadriceps can resist varus/valgus moments concomitantly with flexion moments through the simultaneous generation of flexion and extension moments (Lloyd and Buchanan 2001). In this work, co-contraction provided resistance for 11-14% of the external valgus/varus moments imparted upon the joint. Cadaveric modeling experiments have observed the resultant ACL load when hamstrings to quadriceps co-contraction changed (Withrow, Huston et al. 2008). Using a cadaveric knee set up designed to impart a two to three bodyweights impulsive axial load upon a cadaveric knee joint, the authors varied hamstrings forces and report a hamstrings to quadriceps co-contraction ratio (H force / Q force) of 0.64 practically negates the effect of a 2093 N impulsive axial force (ACL strain: 0.8 %) (Withrow, Huston et al. 2008). However when the ratio was much lower (0.22) ACL strain was much higher (ACL strain: 3 %) despite a lower (1700 N) impulsive loading. This suggests that there is a reciprocal relation between internal and external forces that affects ACL load, so that when hamstrings forces are increased, with quadriceps forces remaining constant, greater ground reaction forces can be applied to the joint without increasing ACL load. This is further confirmed by the reports of an in-vivo model designed to assess the rotational stiffness provided by the maximal voluntary activation of the leg muscles. The purpose was to assess the change in knee rotational stiffness, assessed as the response to an 80 N impulsive internal rotation load applied to the external aspect of the foot with the knee in 90° of flexion. Findings show that when the participants were maximally activating, rotational stiffness was increased by 178 and 218 % compared to the relaxed condition (Wojtys, Ashton-Miller et al. 2002). Taken together this information suggests that the simultaneous and collective function of the hamstrings and quadriceps can contribute to the control of excursions and forces in the three planes of motion.

In summary, when co-contracting, the hamstrings and quadriceps produce forces in opposite directions, generate compressive forces (Elftman 1966), and increase knee stiffness in multiple planes of motion (Solomonow, Baratta et al. 1987; Baratta, Solomonow et al. 1988; Lloyd and Buchanan 2001; Wojtys, Ashton-Miller et al. 2002). This active stabilization of the knee aids passive restraint systems in maintaining joint stability (Solomonow, Baratta et al. 1987; Baratta, Solomonow et al. 1988; More, Karras et al. 1993) and equalizes articular surface pressure distribution (More, Karras et al. 1993). Collectively, these mechanics are thought to allow for less stress to be transmitted to the ACL for a specific level of externally applied force (More, Karras et al. 1993). Thus, combined action of the hamstrings and quadriceps has the ability to control loads in all planes of motion and potentially reduce ACL load.

This review of high-risk knee joint mechanics for ACL injury demonstrated that the ACL is loaded during the impact phase of a deceleration maneuver when the knee is close to full extension and undergoing forces and excursions in multiple planes of motion. It is also demonstrated that quadriceps and hamstrings forces can contribute to both greater and lesser high-risk knee joint mechanics which in turn may influence ACL load. However, we were unable to locate work specifically addressing this issue during a deceleration maneuver. The next section will focus on the literature seeking a better understanding of the contribution of the quadriceps and hamstrings activity to high-risk knee joint mechanics during deceleration maneuvers.

## Deceleration Neuromechanics

The following section will discuss: 1) the theoretical contribution of internal and external forces to knee joint mechanics during a deceleration maneuver, and 2) the findings related to the contribution of quadriceps and hamstrings to high-risk knee joint mechanics as observed in-vivo.

### Theoretical Neuromechanics of Deceleration

During a deceleration maneuver, hamstrings and quadriceps co-contraction is necessary to maintain knee joint stability (Baratta, Solomonow et al. 1988; More, Karras et al. 1993), and to avoid extreme knee joint positions (O'Connor, Monteiro et al. 2009). Prior to ground contact, preparatory quadriceps forces may load the ACL (Cerulli, Benoit et al. 2003) by shifting the tibia anteriorly as demonstrated in-vitro (Torzilli, Deng et al. 1994). This is most apparent when knee flexion is less than 30° due to the importance of the horizontal component of the vector of quadriceps force (Smidt 1973; van Eijden, de Boer et al. 1985; Buff, Jones et al. 1988). Upon foot contact, ground reaction forces experienced result in an external knee flexion moment that must be counteracted by an internal extensor moment so that the knee does not collapse in flexion. This function is provided through eccentric action of the quadriceps, increasing the knee extension moment (Blackburn and Padua 2008; Hanson, Padua et al. 2008) and contributing to anterior translation of the tibia relative to the femur (DeMorat, Weinhold et al. 2004). Thus, isolated eccentric quadriceps action is necessary to the maintenance of knee integrity but is also thought to contribute to high-risk knee joint mechanics.

As antagonists to the quadriceps, the hamstrings also activate prior to ground contact (Palmieri-Smith, Wojtys et al. 2008; Shultz, Nguyen et al. 2009). However, whether they are acting eccentrically, isometrically or concentrically after ground contact is still a topic of debate. The hamstrings must shorten with knee flexion and lengthen with hip flexion since they cross both the hip and the knee (Visser, Hoogkamer et al. 1990). Therefore, the extent to which they change length is a function of the relative changes in hip and knee flexion. Since those hip and knee flexion excursions occur in multiple combinations (Blackburn and Padua 2008) it is difficult to determine the action of the hamstrings precisely. Although the change of hamstrings length has not been reported during the initial deceleration of a drop jump maneuver, there is evidence to suggest that they are either acting concentrically (shortening) or isometrically (remaining at the same length) early in deceleration maneuvers (Robertson, Wilson et al. 2008; Jonhagen, Halvorsen et al. 2009). One study modeled hamstrings length as a function of the combinations of knee and hip angles during a fully loaded squat (Robertson, Wilson et al. 2008). They reported that bicep femoris and semitendinosus shorten during the descending phase of the squat (Figure 1), occurrences coupled with knee and hip motions.

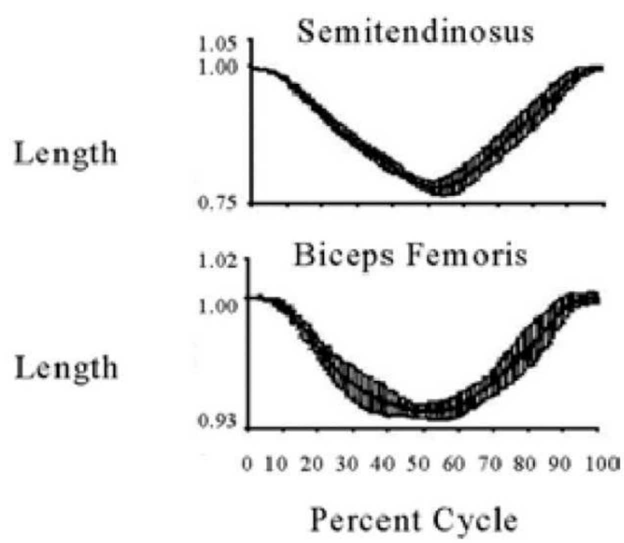



Figure 1: *Semitendinosus and bicep femoris Lengths Showing a Relative Shortening During the Flexion Phase (0-50%) of a Squat.*
*The change in hamstrings length, with reference to their standing length, during a loaded squat maneuver shows that bicep femoris and Semitendinosus lengths decrease throughout the descending phase of a squat maneuver. 0 % of the cycle represents the beginning of the squat, when the individual is standing, 50 % is the deepest knee flexion and 100 % when they return to standing (Robertson, Wilson et al. 2008).*

A slightly different model based on the identification and tracking of the origin and insertion of the hamstrings during a jump lunge, with coupled hip and knee flexion, reported similar findings, with bicep femoris shortening during simultaneous knee and hip flexion (Jonhagen, Halvorsen et al. 2009). The methods also allowed for an estimation of the quadriceps length and findings report that the hamstrings shortening velocity was less than that of the lengthening of the quadriceps (Figure 2) (Jonhagen, Halvorsen et al. 2009).

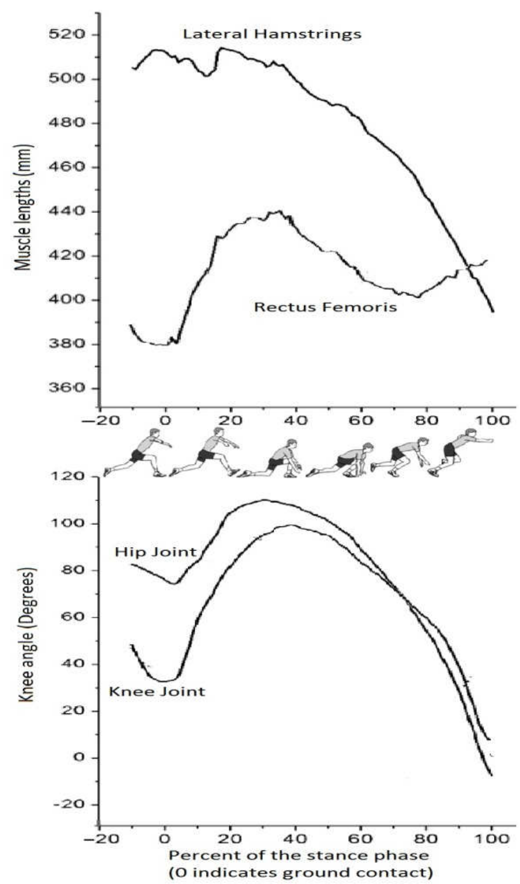



Figure 2: *Joint Angles and Muscle Lengths During a Jump Lunge.*
*The hamstrings shorten between ground contact (0%) and deepest knee flexion (40% of the landing and propulsion phases) (Jonhagen, Halvorsen et al. 2009)*

As the previously mentioned activities have joint motions similar to deceleration maneuvers there appears to be preliminary evidence that the hamstrings may act concentrically during deceleration maneuvers. From this concentric muscle action, the horizontal component of the hamstrings force vector results in a posterior shear force that counteracts the anterior shear forces and ACL load from the ground reaction force and quadriceps action (Yu, Lin et al. 2006). This mechanism is considered to be most efficient when the knee flexion angle is larger than 30° (Imran and O'Connor 1998). Collectively, this evidence suggests that hamstrings and quadriceps are acting concentrically and eccentrically, respectively, and that the hamstrings shortening velocity is less than that of the quadriceps.

***Inverse Dynamics***

When the foot is in contact with the ground, inverse dynamics analysis provide a means to investigate the net joint moment resulting from all linear and rotational forces acting about the joint (Winter 1990). Determination of these loads is based upon: ground reaction forces, kinematic information about body segments and anthropometric data. Just as the linear and rotational forces are transmitted through, and absorbed by, the successive segments up the kinetic chain, inverse dynamics calculations successively estimate the forces occurring at the joints in a distal to proximal direction. Moments and forces are commonly normalized to markers of an individual’s anthropometrics to best allow inter-individual comparisons. It is important to remember that joint forces and moments calculated represent the internal forces produced by both the active and passive structures. As such, inverse dynamics do not inform of the respective flexor and extensor torque exerted upon the joint by the antagonistic muscle groups. Therefore, during knee flexion in a closed kinetic chain, the net knee moment, as estimated through an inverse dynamics analysis, is the sum of the quadriceps and hamstrings forces together with the restraints provided by the passive structures of the knee. Detailed knowledge of the antagonistic muscular contributions to knee joint mechanics may allow a better understanding of the factors contribution to knee joint mechanics during deceleration maneuvers. As mentioned before, ground reaction forces create an external knee flexion moment. If this moment is completely balanced by a strong quadriceps action (internal extension moment) the result will be a rapid deceleration (large decrease in knee flexion velocity, small knee flexion excursion) with high ground reaction forces. This would result in what is described as a stiff landing style (Zhang, Bates et al. 2000) and likely higher stress on the passive structures (Butler, Crowell et al. 2003). Conversely, a well graduated quadriceps and hamstrings co-contraction will result in longer period of deceleration (smaller decrease in knee flexion velocity, larger knee flexion excursion) with decreased ground reaction forces. This would result in what is described as a soft landing style (Zhang, Bates et al. 2000) and likely less stress transmitted to the passive structures (Butler, Crowell et al. 2003). Therefore, the magnitude of ground reaction forces observed is in part modulated by the respective magnitude of quadriceps and hamstrings torques. However, the information provided by kinetic analyses (including inverse dynamics) does not provide an insight into this antagonistic modulation, or the contribution of quadriceps and hamstrings to the net joint moment and only inform of the net knee joint extensor moment.

In summary, studies suggest that the balance of internal extension (eccentric quadriceps) and flexion (isometric/concentric hamstrings) moments with external forces (such as ground reaction) largely determine the kinetic and kinematic occurrence, such as anterior shear force and anterior tibial translation at the knee joint. However, current methods only allow for the assessment of net knee joint moments and largely fall short of providing relevant information regarding the respective contribution of quadriceps and hamstrings torques.

### In-Vivo Neuromechanics of Deceleration

Since muscle forces cannot be directly measured in-vivo, surface electromyography (sEMG) is commonly used as an alternative as it provides a non invasive means to observe the activation of muscles. sEMG data collection is based on the application of electrodes, which are designed to record the voltage potentials propagated across the superficial muscles fibers lying beneath the skin. As muscle force and sEMG amplitude are, in a broad sense, correlated in isometric actions (Enoka 2002), the amplitude of the sEMG signal can be used to understand how muscle activity relates to motions and forces determined by kinematics and kinetics (Dubo, Peat et al. 1976). It is common practice to normalize the sEMG collected during dynamic function to a peak value obtained during a maximal voluntary isometric action. The rationale for this practice is the amplitude and frequency content of the raw sEMG signal between muscles and individuals is influenced by non physiological factors such as electrodes placement or subcutaneous fat (Dubo, Peat et al. 1976). As such, normalized sEMG values represent the percent of maximal activation during an isometric action and give an indication of the level of activation going through the muscle.

The literature is limited with regard to the contribution of quadriceps and hamstrings to high-risk knee joint mechanics during deceleration maneuvers. To our knowledge only two studies have assessed the predictors of high-risk knee joint mechanics, based on kinematic, kinetic and sEMG analyses of deceleration maneuvers (Sell, Ferris et al. 2007; Shultz, Nguyen et al. 2009). There appears to be partial agreement with evidence in-vitro of the contributors to high-risk knee joint mechanics. More specifically, Sell et al. (2007) and Shultz et al. (2009) found that knee joint angle at peak posterior ground reaction forces and knee flexion excursion, respectively, were small predictors of anterior shear force; in agreement with the in-vitro findings. It was also reported that the net knee extensor moment as determined via inverse dynamics was the strongest predictor of anterior shear force. This moment is largely a function of the net force, or the summation of quadriceps and hamstrings forces, transmitted to the patellar tendon (Krevolin, Pandy et al. 2004) and is a large contributor to anterior shear force similar to the cadaveric work presented previously (Nunley, Wright et al. 2003). Thus, estimating these muscle forces and resultant moments may be important in fully understanding their influence on landing mechanics.

However, the contribution of the quadriceps and hamstrings during deceleration maneuvers was not found to be as determinant of landing mechanics in-vivo (Sell, Ferris et al. 2007; Shultz, Nguyen et al. 2009) as evidenced in-vitro (Withrow, Huston et al. 2008). In vitro, quadriceps forces are reported to be large contributors to anterior shear forces and net knee extension moments (Nunley, Wright et al. 2003) and hamstrings forces to flexion and posterior shear forces (Mesfar and Shirazi-Adl 2006). The predictors considered in vivo included the strength and activation (Shultz, Nguyen et al. 2009) or activation only (Sell, Ferris et al. 2007) characteristics of the quadriceps and hamstrings. Shultz et al. (2009) reported that quadriceps activation, even when accounting for individual strength, did not predict peak net knee extensor moment. Both Shultz and Sell also report similar results that the quadriceps and hamstrings activity were small or non significant predictors, respectively, of peak anterior shear forces determined via inverse dynamics. More specifically, the only significant neuromuscular predictor of anterior shear forces was quadriceps activity in the 250 ms post deceleration. It explained 7.3% more of the variance of anterior shear force than that explained by peak knee extensor moment, knee flexion excursion and hip flexion excursion (Shultz, Nguyen et al. 2009). The findings collectively fail to confirm the contribution of quadriceps and hamstrings torques to knee joint mechanics as highlighted in vitro. However, they confirm the importance of the internal net joint moment to predict anterior shear forces. Sell et al. (2007) points out that only knowing the internal net extensor moment observed via inverse dynamics does not allow one to determine whether its change is modulated by greater quadriceps or by lesser hamstrings forces.

To summarize, in agreement with in-vitro findings, shallow knee joint angles, ground reaction forces and greater net knee extensor moments are important contributors to greater anterior shear force during in vivo deceleration maneuvers. However, lack of knowledge of the specific contribution of quadriceps and hamstrings torques to the net internal knee extensor moment during deceleration maneuvers limits our ability to characterize the contribution of thigh muscle torques to high-risk knee joint mechanics. When assessed during deceleration maneuvers, the relevance of using sEMG to assess the contribution of quadriceps and hamstrings to kinetics and kinematics of the knee joint appears to be minimal, in disagreement with in-vitro findings. One possible explanation for this discrepancy is that the sEMG methods commonly used to estimate muscle activity fail to fully represent the relation between the amplitude of the signal and the amplitude of the force produced by the muscle.

## Evidence Supporting the Adoption of a New Method to Estimate Knee Torques

During a deceleration maneuver, the quadriceps and hamstrings muscles function antagonistically to control knee motion and assist in stabilization of the joint. Since muscle force is practically impossible to measure in-vivo, sEMG has been used as an alternative to study the contribution of muscles to movement. Because sEMG represents the electrical activity going through the muscle sEMG as detected at the surface of its belly, the amplitude of the signal is related to the isometric force produced at the muscle (Woods and Bigland-Ritchie 1983; Disselhorst-Klug 2009). However, this quantitative relation is influenced by external factors, such as the length and the rate of change in length that occur in the muscle when the force and sEMG measurement are made (Bigland and Lippold 1954). Based on these differences, we will present the limitations of using the isometric method to infer the torques produced about the joint by the quadriceps and hamstrings during the impact phase of a deceleration maneuver.

## The Isometric Method and its Potential Limitations

### *Technique and Purpose*

The need for the normalization of the sEMG signal has been long recognized due to the inherently high variability of the raw sEMG signal. One of the most common normalization methods in gait studies was introduced by Dubo et al. (1976). In this process, each sEMG data point collected during dynamic function is divided by the peak sEMG recorded during a maximal voluntary isometric action. This technique has allowed researchers to compare antagonistic muscles in the same individual and also has provided a mean to make inter-individuals comparisons.

### *Limitations of Using sEMG to Represent Torque*

Making inter-individual and inter muscle comparisons largely assumes that surface EMG is an accurate representation of torque, independent of the individual, the knee joint angle, the knee joint angular velocity or even the muscle of interest. However, it is important to remember that sEMG may not be an accurate representation of torque during dynamic function since the sEMG/Torque ratio changes across the range of motion (Doorenbosch and Harlaar 2003), joint velocities (Bigland and Lippold 1954) and individuals (Dubo, Peat et al. 1976).

The purpose of this section is to describe the limitations of using the isometric method to infer the torques produced about the joint by the muscles. Early on, the use of the isometric method was pointed out to have limitations in interpreting dynamic muscular function (Dubo, Peat et al. 1976), however this has been somewhat ignored in the literature. These limitations are related to the fact that the total force producing capability of a muscle is affected by the number of cross bridges formed and the force produced by each of those cross bridges (Kandel, Schwartz et al. 1995). The number of cross bridges formed depends on the length of the muscle (Kandel, Schwartz et al. 1995), the force produced is a function of the velocity of cross bridge motion (Kandel, Schwartz et al. 1995). The respective influences of length and velocity on force production are described using two separate relationships, the length-tension relationship and the force-velocity relationship. These relationships have originally been modeled in-vitro, but for the purpose of this document, in-vivo results will be presented. In-vivo the angle torque relationship (length-tension) is similar to in-vitro however it is also affected by the change of muscle moment arm as a function of knee joint angle (Maganaris 2001; Krevolin, Pandy et al. 2004).

The Angle-Torque Relation

Briefly, the angle torque relation demonstrates a relative increase in torque production in the mid range of muscle length/range of motion, followed by a small decrease (Figure 3)(Newman, Jones et al. 2003). In-vivo the angle torque relationship (i.e. length tension relationship) is similar to in-vitro. However, it is also affected by the change of muscle moment that occurs with knee flexion (Maganaris 2001; Krevolin, Pandy et al. 2004).This suggests that, in order to infer torque from activation levels, one has to account for knee joint angle since, for a specific level of activation, 30 to 40 % more torque can be produced between 50 and 70 ° of knee flexion compared to other knee joint angles (Figure 3).

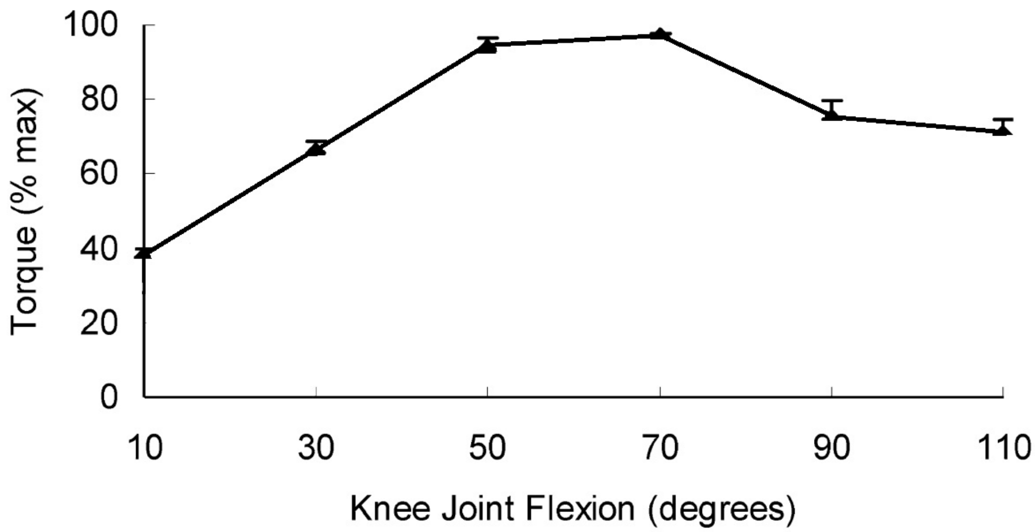



Figure 3: *Quadriceps angle-torque relationship.*
*The dependency of torque production on knee joint angle during knee extension is demonstrated as the peak angle torque for quadriceps torque production is between 50 and 70 ° of knee flexion (adapted from (Newman, Jones et al. 2003)).*

The Velocity-Torque Relation

The velocity-torque relation reveals a differential response of concentric and eccentric actions across the muscle velocity spectrum, even when accounting for differences in activation patterns along the velocity spectrum (Figure 4). This suggests that when a muscle is acting eccentrically, its torque producing capability for a constant level of activation is larger than the same muscle contracting at the same, concentric velocity, and also that this difference increases with increasing eccentric and concentric velocities (Yeadon, King et al. 2006).

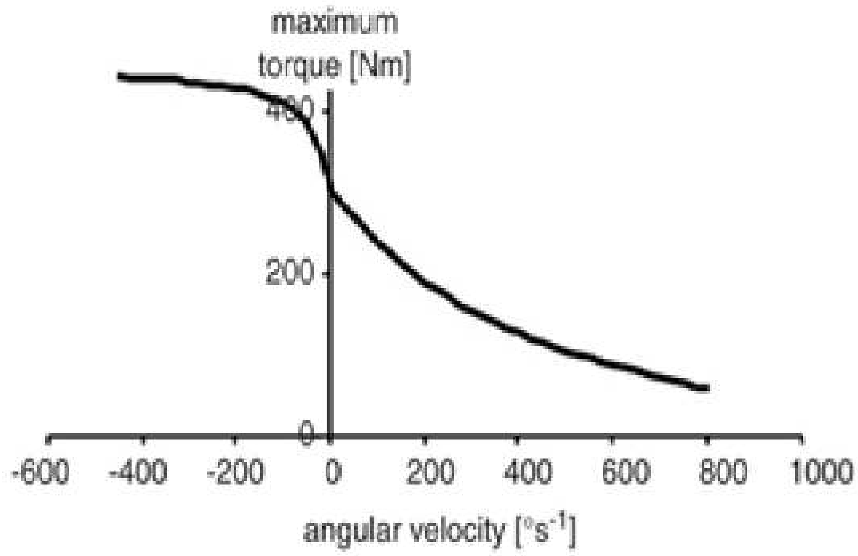



Figure 4: *The relation between velocity and torque production (Yeadon, King et al. 2006). The dependency of knee extension torque production on angular velocity for a maximal level of activation is demonstrated as the largest difference in torque production capability is observed between high eccentric velocities and high concentric velocities.*

Even though the two relationships presented above are physiologically independent, they combine to affect how much torque a muscle can produce as depicted in Figure 5. Synthesizing both the angle-torque and velocity-torque relationships renders a complex but useful three dimensional graph (Figure 5) that models the force generating capability of a muscle for a specific level of activation. Muscle fiber forces are greatest during high velocities lengthening actions when the muscle is in a lengthened position; conversely, the lowest force production occurs when the muscle fiber is shortening at high velocity and at shorter lengths (Brown, Scott et al. 1996). As stated previously, using the isometric method logically implies that sEMG amplitude during dynamic function is representative of the torque produced about the joint by a muscle. In turn this implies that independent of knee joint angle or knee angular velocity, a muscle can produce the same torque, for a constant level of sEMG. Given the above review of the angle-torque and torque velocity relationships, this appears to not be the case.

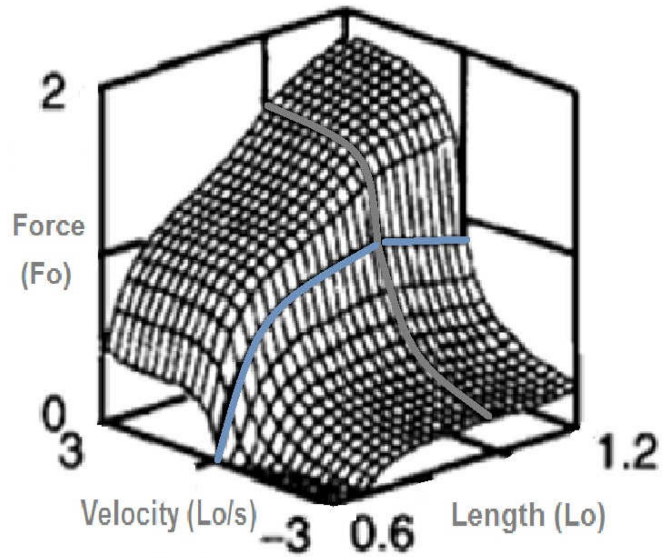



Figure 5: *Force-length-velocity relationships of a muscle fiber (Brown, Scott et al. 1996). The dependency of force production capability is modeled as a function of both the velocity of movement and the length of the muscle for a constant level of activation. 0 velocity represents an isometric action, 1.0 length is the optimal muscle length and force is expressed as a function of the force produced at the peak force angle during an isometric action.*

Furthermore, studies using maximal voluntary isometric actions use only one angle to test for peak sEMG and that angle varies largely amongst studies [e.g. 25° in Shultz et al. (2009) and 60° in Sell et al. (2007)]. This means that any sEMG observed during dynamic function will be interpreted as having the same sEMG/Torque ratio as at the specific knee joint angle chosen for the isometric action. However, the previous discussion of the angle-torque and torque-velocity relationships demonstrated that this is not a valid approach since, for a specific level of activation, largely different torques may be produced depending on the knee joint angle (Figure 3) and knee joint angular velocity (Figure 4).

Another aspect of this issue appears when one aims to compare antagonistic muscles at a specific knee joint angle. That is, one muscle is theoretically contracting eccentrically and the other ones concentrically, with largely different sEMG/Torque ratio due to the differences in velocity of the action undertaken (Westing, Cresswell et al. 1991). This complicates quadriceps and hamstrings comparisons during deceleration maneuvers as they have been found to act eccentrically and isometrically/concentrically (Robertson, Wilson et al. 2008), respectively, and those actions produce notably different magnitudes of torques (Westing, Cresswell et al. 1991). The isometric method does not account for this issue due to the nature of its calculations (Dubo, Peat et al. 1976). The isometric method does not account for knee joint angle or angular velocity specificities in the sEMG/Torque relation; or for differences in torque generation capabilities amongst quadriceps and hamstrings. Therefore, normalizing sEMG using the isometric method may lead to large errors of interpretation of the mechanical influences of the quadriceps and hamstrings about the knee joint during deceleration maneuvers.

To understand the extent to which this error may impair our interpretation of the mechanical contribution of the quadriceps and hamstrings to knee joint mechanics, the respective functions of quadriceps and hamstrings muscles need to be detailed during deceleration maneuvers. The quadriceps start from a relatively shortened position with the knee in about 20° of flexion and act mostly at lengths shorter than that of their peak torque production during eccentric actions (60 to 69° of flexion) (Aagaard, Simonsen et al. 1995; Brughelli, Cronin et al. 2010). Since the quadriceps are the main contributor to deceleration at the knee, they must lengthening in a rapid manner with knee flexion velocity suggested to be 150 $°.s^{-1}$ - 230 $°.s^{-1}$ at touchdown during the deceleration of a stop jump task (Yu, Lin et al. 2006). Based on angle-torque and velocity-torque relationships we can observe that the sEMG/Torque ratio in an isometric action at relatively short lengths is much different, in this case higher, than that which can be produced at the same length during a lengthening action (Kellis and Baltzopoulos 1998). The difference becomes even greater when the knee moves into more flexion. Therefore, using the isometric method will assume a higher ratio than should be and as such will likely lead to an underestimation of the torque produced about the joint by the quadriceps during a deceleration maneuver as surmised from EMG data.

During landing, the hamstrings start from a relatively lengthened position very close to the peak torque angle for concentric hamstrings action (peak torque angle around 35° at 60 °s-1) (Aagaard, Simonsen et al. 1995; Onishi, Yagi et al. 2002) and 50° of knee flexion during isometric actions in a fully prone position (Kilgallon, Donnelly et al. 2007). The velocity at which the hamstrings are acting during deceleration maneuvers is difficult to determine (as we have been unable to locate this specific information for deceleration maneuvers) as it depends on the combined flexion of the hip and the knee (Blackburn and Padua 2008). Evidence presented in an earlier section of this literature review shows that the hamstrings are possibly remaining isometric or shortening at slow velocity (Robertson, Wilson et al. 2008; Jonhagen, Halvorsen et al. 2009), however this remains specific to each individual and is influenced by the combined motions of the knees and hips (Jonhagen, Halvorsen et al. 2009). Overall, this concentric action is likely much slower than that observed in the quadriceps due to the single joint nature of the majority the quadriceps musculature.

Therefore the assumption made by the isometric method that the sEMG/Torque ratio remains similar to that observed in an isometric action at a specific knee joint angle of flexion may not be as misleading for the hamstrings as it is for the quadriceps and therefore using the isometric method may lead to only a minor misestimation. More specifically, during concentric actions the sEMG/Torque ratio is only somewhat higher than for isometric actions (Brown, Scott et al. 1996), which suggests that the estimation of the concentric torque produced about the joint by the hamstrings, as provided by the isometric method, is likely to only be a slight overestimation of its actual value. However, this is based on the assumption that the hamstrings are shortening or remaining isometric during the impact phase of a landing. Given the information that we have presented in an earlier section regarding the high inter-individual variability in hamstrings changes in length, this may be problematic. In fact if the hamstrings are actually lengthening during the impact phase, the sEMG/Torque relation would be largely altered and the isometric method would provide a large underestimation of the actual torque produced by the hamstrings.

When comparing the normalized sEMG of antagonistic muscles, other issues arise since the isometric method does not account for differences in torque generating capability between the quadriceps and hamstrings. The quadriceps can produce much more knee extension torque eccentrically than the hamstrings can produce flexion torque concentrically or isometrically. Aagaard et al. (1995) provide data to make the comparison between the maximal torque generating capability of the quadriceps acting eccentrically and the hamstrings acting concentrically. The quadriceps can produce around 300 Nm at 120 $°s^{-1}$ whereas the hamstrings can only produce around 100 Nm at 120 $°s^{-1}$ (Aagaard, Simonsen et al. 1995). When comparing the normalized sEMG of the quadriceps and hamstrings, differences in the maximal, or angle specific torque generating capability of those muscles, are not accounted for by the isometric method. Since antagonistic muscles' normalized sEMGs are compared as such (Kellis, Arabatzi et al. 2003), it means that the mechanical influence of the quadriceps upon the knee joint is largely underestimated and that the mechanical influence of the hamstrings upon the joint is slightly over-estimated.

Thus, the use of the isometric method to infer the torques produced about the joint by antagonistic muscles may lead to large errors of estimation, with the quadriceps and hamstrings torques being largely under and over-estimated, respectively. Studies have compared the normalized sEMG of antagonistic muscles (Burden, Trew et al. 2003), and the above mentioned errors may accumulate and lead to erroneous conclusions. In turn this may explain the lack of significance of the in-vivo findings regarding the contribution of thigh muscles to high-risk knee joint mechanics (Sell, Ferris et al. 2007; Shultz, Nguyen et al. 2009). Collectively, the evidence presented here provides strong support to the notion that it is necessary to account for differences in sEMG/Torque ratio across muscles, knee joint angles and knee angular velocities for valid comparisons of the mechanical influences of quadriceps and hamstrings about the knee joint during deceleration maneuvers to be possible.

To summarize, we have presented findings that demonstrate large differences in the sEMG/Torque relation across concentric and eccentric actions suggesting that inferring the mechanical influence of hamstrings and quadriceps about the knee joint based on sEMG normalized using the isometric method may lead to a poor estimation of their relative contribution. The use of a calibration scheme addressing those issues appears to be justified. Models (to be discussed below) have addressed these issues and gathered valid information regarding the mechanical influence of quadriceps and hamstrings during lower extremity dynamic function.

### The sEMG/Torque Ratio Method

#### *Purpose and Technique*

The purpose behind the use of sEMG/Torque ratio modeling has been primarily driven by a clinical need to understand the contribution of thigh muscles torques to knee joint mechanics (Doorenbosch and Harlaar 2003; Doorenbosch and Harlaar 2004). This section will therefore focus on models of muscle torque estimation that are less computationally intensive than forward dynamics so that they can be used as clinical tools based on direct observation rather than on post-collection data treatment and optimization.

As in the isometric method, the initial step to sEMG/Torque processing is based on discrete quadriceps and hamstrings “calibration” actions, where torque, knee joint angle, knee angular velocity and sEMG are collected synchronously (Doorenbosch and Harlaar 2003; Doorenbosch and Harlaar 2004). For each action, the sEMG/Torque ratio is calculated across the range of motion and then modeled as a second order polynomial as a function of the knee joint angle (Doorenbosch and Harlaar 2003). This results in a muscle specific equation where the input variable is knee joint angle and the output variable, the sEMG/Torque ratio. In some cases, the dependency of the sEMG/Torque ratio on velocity has also been described (Doorenbosch and Harlaar 2003). Based on kinematic data gathered during dynamic function, the observed knee joint angle is used as an input in the previously described equations to interpolate the sEMG/Torque for the quadriceps and hamstrings, separately. Finally, the sEMG collected during dynamic function and the simultaneously interpolated sEMG/Torque ratio can be used to estimate muscle torque. Not only does this provide an estimation of isolated muscle torque, but it can also be used to solve for the net knee joint moment (Doorenbosch and Harlaar 2003).

***Validity***

Both Internal (Doorenbosch and Harlaar 2004; Doorenbosch, Joosten et al. 2005) and external validity (Doorenbosch and Harlaar 2003) of the sEMG/Torque method have been previously assessed. Internal validity is defined as the ability of the method to estimate muscle torques as observed through isokinetic assessment. External validity assessment focuses on the quantitative comparison of the resolved net joint moment as previously described and the net knee joint moment determined via inverse dynamics analysis.

In Doorenbosch and Harlaar (2004), healthy participants were asked to perform maximal concentric actions of the knee flexors and extensors at seven different velocities (30, 60, 90,120, 150,180 and 210 $°s^{-1}$). For each individual separately, data from five of those velocities (30, 60, 90,180 and 210 $°s^{-1}$) were used to create a second order polynomial function expressing the relation between sEMG and torque as a function of muscle, knee joint angle and of knee velocity. The internal validity was verified by inputting actual sEMG, velocity and knee joint angle from one of the two remaining velocities (150 and 180 $°s^{-1}$) and comparing the estimated torque values with the actual torque values recorded by the dynamometer. Those comparisons included the root mean square as well as the absolute difference. The absolute estimation error of the sEMG/Torque model at 150 $°s^{-1}$ was larger in extension (Absolute: Mean: 11.8 ± 6.2 Nm, RMS: 19 ± 9%) than flexion (Absolute Mean: 8.78 ±3.69 Nm, RMS: 20 ± 7%). This error was considered acceptable for the purpose of estimating isokinetic, mono-joint movement torque based on sEMG as more complex models to estimate torque based on sEMG, have found similar errors (20 to 29 % error) (Hof, Pronk et al. 1987).

Additionally, sub maximal (50% and 75 % of MVC) actions have been used to calibrate the sEMG/Torque model (Doorenbosch, Joosten et al. 2005). When this model was used to estimate torque at 100 % effort, reports suggest similar validity to that observed in the previously presented study (Doorenbosch, Joosten et al. 2005). The relative errors ranged from 6 to 21% for extension and 12 to 23% for flexion (Doorenbosch, Joosten et al. 2005).

Limited work has investigated the validity of using the sEMG/Torque model to estimate the net knee joint moment during dynamic function. Doorenbosch and Harlaar (2003) used concentric calibration actions at $90°s^{-1}$ to model the dependency of the sEMG/Torque ratio on knee joint angle using a second order polynomial. Participants were then asked to perform a single legged jump with the sEMG electrodes still attached, while 2D kinematics and ground reaction forces were also collected during the push off phase. Subsequently, sEMG/Torque ratios were interpolated, for the hamstrings and quadriceps separately, based on the knee joint angle observed during the push off. Quadriceps and hamstrings torques were estimated separately by dividing the sEMG data by the sEMG/Torque ratio for the same angle. Finally, subtracting hamstrings from the quadriceps estimated torques rendered an estimation of the net knee joint moment which was further compared to the data provided by an inverse dynamics analysis. To estimate the external validity of the sEMG/Torque method, the absolute (in Nm) and relative (as a percent of the moment provided by the inverse dynamics analysis) differences between the net extensor moment found via inverse dynamics and the net joint moment found via sEMG/Torque were calculated (RMS error: $15.3 \pm 3.7$ Nm, or $13.3 \pm 3.8$ %). It is difficult to comment on the quality of the fit of the sEMG/Torque method to inverse dynamics (due to lack of comparable data), aside from the rather small difference between the two methods. However, it is important to note that the sEMG/Torque and inverse dynamics address different components of the dynamic stability systems. The biggest difference is that inverse dynamics estimate net joint moment as created by all knee dynamic and passive components, whereas sEMG/Torque only studies the contribution of the quadriceps and hamstrings to the net joint moment. As such it can be expected that the two methods will provide slightly different estimates of the net joint moment.

Estimating muscle or joint torque using the sEMG/Torque ratio is a clinically relevant method, as it relies on the performance of only a few, not necessarily maximal, calibration actions. It allows estimation of net joint or muscle moment with similar levels of accuracy as in other, more complex, models e.g.: 22% (Hof, Pronk et al. 1987). Despite its demonstrated relevance, this method has never been used to analyze the contribution of thigh muscles torques to high-risk knee joint mechanics.

***Limitations & Shortcomings***

Using sEMG/Torque modeling relies on three main assumptions that may affect its validity and/or the relevance of the findings. First, it is assumed that the muscles act at the same velocity as that observed during the calibration actions. This may be problematic for muscles, such as the hamstrings, that cross two joints and therefore may lengthen or shorten depending on the respective and simultaneous motion of the hip and knee joints. Since this combination of motions at the hip and knee joints are known to occur in different ways between individuals during lower extremity motion, the validity of the findings may be affected, even if the evidence found in the literature lends evidence to the contrary (Doorenbosch and Harlaar 2003; Doorenbosch and Harlaar 2004; Doorenbosch, Joosten et al. 2005).

Second there is an inherent assumption that the angle specific relation between sEMG and torque is linear across intensities (Doorenbosch, Joosten et al. 2005). There is some evidence that this relation might be slightly curvilinear in some limited cases, for example when the rectus femoris and vastus medialis are contracting at intensities between 20 and 40% Alkner and Tesch (2000). However, it appears that it does not affect the validity of this method and that maximal or sub maximal calibration actions may be used without altering the validity of the resultant model (Doorenbosch and Harlaar 2004; Doorenbosch, Joosten et al. 2005).

Finally, it is assumed that the relation between sEMG and torque remains the same between open kinetic chain, as in calibration actions, and closed kinetic chain, as in dynamic function. Addressing this issue Alkner et al. (2000) found no difference in the isometric sEMG/Torque relation of the quadriceps muscle group between leg press and knee extension.

Collectively, it appears that sEMG/Torque modeling may be a viable approach to study quadriceps and hamstrings mechanical influence upon the knee joint during deceleration maneuvers. When purporting to infer mechanical contributions of thigh muscles to knee joint motion and moments, the processing procedure of the SEMG recordings should consider differences in the SEMG/Torque relation between muscles, knee joint angles and knee angular velocities.

The goal of this work was to provide a rationale for the use of a sEMG/Torque calibration procedure in studying the mechanical influence of the quadriceps and hamstrings upon the knee joint during deceleration maneuvers. This method accounts for known differences in the sEMG/Torque relation across muscles, knee joint angles and knee joint angular velocities and provides an estimation of quadriceps and hamstrings torques during deceleration maneuvers. This will contribute to a better understanding of the torques imparted about the knee joint by the quadriceps and hamstrings and in turn of their potential contribution to high-risk knee joint mechanics.

Summary

With knee kinetics and kinematics being central to high-risk mechanics and ACL load, the first part of the review focused on the description of the influence of internal and external forces on ACL load during the impact phase of a deceleration maneuver, providing a background of the kinematic, kinetic and neuromuscular factors that contribute to ACL load both in controlled and more realistic environments.

Next, studies reproducing the demands of a deceleration maneuver in a laboratory environment were presented, specifically as they relate to the contribution of thigh muscles activity to anterior shear force and knee extensor moment. It was demonstrated that the findings of those studies are somewhat in disagreement with the findings of in-vitro studies, as they generally fail to identify associations of thigh muscle activity to high-risk knee joint mechanics.

Lastly, we presented the limitations of using sEMG normalized with the isometric method to represent the torque produced about the joint by a muscle. We further introduced potential reasons for the lack of findings in the relation between thigh muscle activity and knee joint mechanics, which are likely linked to the fact that the isometric method does not account for fundamental differences in sEMG/Torque across muscles, knee joint angles and knee angular velocities. We finally introduced the rationale for a relatively new method, accounting for specificities in the sEMG/Torque ratio, to investigate the mechanical contribution of quadriceps and hamstrings to high-risk knee joint mechanics during deceleration maneuvers.

Although it is clear from evidence in-vitro and common understanding of neuromechanics that thigh muscle forces largely contribute to high-risk mechanics during deceleration maneuvers, there is no clear evidence of this in the in-vivo literature. This may be due to long recognized limitations of using the isometric method to infer the torques created about the knee joint by thigh muscles. Based on the current research it appears that the use of an action specific sEMG/Torque relation may contribute to a better understanding of the mechanical contribution of thigh muscles to high-risk knee joint mechanics.

# CHAPTER III

# METHODS

The overall objective of this research was to determine the mechanical contribution of thigh muscles torques impulses, determined through individualized, muscle head and action specific modeling of the sEMG/Torque relation as a function of knee joint angle, to high-risk knee joint mechanics during the impact phase of the initial deceleration of a drop jump maneuver. The central hypothesis was that greater quadriceps torques and lesser hamstrings torques impulses would predict greater magnitudes of high-risk knee joint mechanics.

## Participants

Forty-three healthy female, college students, between the ages of 18 and 25 were recruited from the University to participate in the study. Exclusion criterion for the study were: body mass index >30 (BMI = $wt/ht^2$); a history of knee injury involving the osteochondral surface, ligament, tendon, capsule, or menisci; any medical conditions affecting the connective tissue; a vestibular or balance disorder; or physical activity levels less than 2 or more than 10 h/week. Prior to participation, participants read and signed a consent form approved by the University's Institutional Review Board for the protection of human participants. Only females were studied not only because they have a greater incidence of ACL compared to the male population (Arendt, Agel et al. 1999) but also because this higher incidence is thought to be largely due to specific, modifiable, neuromuscular factors (Hewett, Myer et al. 2005; Hewett, Myer et al. 2007).

## Instrumentation

A calibrated Biodex System 3 isokinetic dynamometer (Biodex Medical Systems Inc.; Shirley, NY) was used to record torque, position and velocity during the strength measurements (calibration actions) used to calculate the sEMG/Torque ratios. A 16 channel Myopac telemetric system (Run Technologies, Mission Viejo, CA) recorded surface electromyography (sEMG) activity of the vastus lateralis (VL), vastus medialis (VM), semitendinosus (ST) and bicep femoris (BF) during the calibration actions and during the drop jump maneuvers. The Myopac unit has an amplification of 1mV/V with a frequency bandwidth of 10 to 1000Hz, a common mode rejection ratio of 90dB min at 60Hz, an input resistance of 1 MΩ, and an internal sampling rate of 8 KHz. The sEMG signal was detected with 10 mm bipolar Ag-AgCl surface electrodes (Blue Sensor N-00-S; Ambu Products, Ølstykke, Denmark) with a center-to-center distance of 20 mm. sEMG data was acquired, stored and exported using DataPac 2K2 lab application software (Version 30.13, Run Technologies, Mission Viejo, CA).

During the drop jump, kinematic data for the pelvis, left thigh, left shank and left foot were sampled at 240 Hz using a PhaseSpace Motion capture Optical system with LED sensors (PhaseSpace; San Leandro, CA) and Motion Monitor software (Innovative Sports Training; Chicago, IL). Kinetic data (1000Hz) was collected using a force plate (Type 4060-nonconducting; Bertec Corporation, Columbus, OH). Those data were automatically collected 500 ms prior to ground contact and 2500 ms after ground contact.

## Procedures

All participants completed a familiarization session followed by a data collection session 2 to 7 days later.

### Familiarization Session

Informed consent was obtained from the participant and the session began by measuring the standing height (cm), body mass (kg), hip, waist and chest circumferences using a standard tape measure, and recording demographic information of the participant, including age and dominant stance limb (determined as the stance leg used to kick a ball). The participant also filled in physical activity and menstrual history questionnaires (See Appendices B and C). The participant was then equipped with sEMG. All skin areas were first thoroughly cleaned with isopropyl alcohol and shaved if necessary. 10mm bipolar Ag-AgCl sEMG electrodes (Blue Sensor N-00-S, Ambu Products, Ølstykke, Denmark) were placed midway between the motor point and the distal tendon of the vastus lateralis and biceps femoris of the dominant limb (left), perpendicular to the length of the muscle fibers (Konrad 2005). A reference electrode was placed over the bony portion of the proximal anteromedial shaft of the tibia.

The participant was then positioned on the dynamometer with the seatback tilted at 25° hip flexion from anatomical zero. The rationale for using 25° of hip flexion was that it approximated the average hip position in the 100 ms post ground contact during landing (Decker, Torry et al. 2003). Seat length was adjusted so that the participants legs were hanging freely with the posterior knee ~1 cm away from the seat edge. The axis of rotation of the knee was aligned with that of the dynamometer using the lateral epicondyle as an anatomical landmark. Straps were secured around the hip and thigh to minimize the contribution of accessory muscles to the torque measured. The dynamometer attachment length was adjusted so that the shin cuff rested comfortably on the tibia, approximately 3 cm proximal to the medial malleolus, to allow full dorsi and plantar flexion. The prescribed range of motion (90°) was defined between 0° (full active extension) and 90° of knee flexion. To assess full active extension and account for seat pad compression, the tester performed successive adjustments to the dynamometer arm so that when the participant was contracting their quadriceps, the leg was at 0 ° of flexion. Then the limb was moved to 90° of flexion by the tester to finalize the definition of the range of motion.

Once set up was complete, the participant was first trained to perform ramping isometric actions (five seconds) of the quadriceps and hamstrings at 25° of knee flexion. The participant was instructed to gradually increase their effort level in producing torque in the desired direction and aim to reach maximal torque by the second or third second. The participant performed 5 to 10 sub maximal actions followed by 2 to 3 maximal actions with 30 seconds of rest between.

The isometric sEMG data for the vastus lateralis (VL) and the bicep femoris (BF) during the maximal ramping isometric repetition with the highest torque observed were saved and band pass filtered (4$^{th}$ order Butterworth, Zero-Lag, 10-300 Hz) and the RMS smoothing (100 ms constant) was calculated (Shultz, Nguyen et al. 2009). The resultant respective peaks in VL and BF isometric sEMG were retained and used in determination of a 20% MVIC threshold of preactivation in other data collection.

The participant then performed 3-5 isokinetic familiarization actions for the eccentric quadriceps (270° $s^{-1}$) and concentric hamstrings (90° $s^{-1}$) calibration actions. Those velocities were chosen as representative of knee flexion velocity observed during the deceleration phase, as previous reports suggest that not only are hamstrings acting concentrically (Robertson, Wilson et al. 2008; Jonhagen, Halvorsen et al. 2009) but also that they shorten slower than the quadriceps lengthen, mostly because their change in length is simultaneously influenced by hip and knee kinematics (Robertson, Wilson et al. 2008; Jonhagen, Halvorsen et al. 2009). Another rationale for choosing these velocities is that there would be similar movement artefacts in the sEMG signal during calibration actions and actual drop jump maneuver which should improve the validity of the model. After these, the VL and BF peaks in isometric sEMG obtained previously were entered separately into DataPac to set up a visual pre-activation threshold and elicit a 20% preactivation prior to the isokinetic muscle action. The 20 % value was chosen as it has been reported as a pre-activation amplitude during a drop jump maneuver (Shultz, Nguyen et al. 2009) and as a mean to control for large differences in individuals pre-activation levels observed during pilot testing.

Performance of isokinetic actions was instructed in three phases, preactivation, maintenance and maximal activation. For preactivation the participant was asked to gradually increase muscle activity until they heard the command ‘hold’, which was given when they reached the desired level (15 to 25 % of max sEMG during MVIC) and marked the end of pre-activation and the beginning of maintenance. During the maintenance phase, lasting around one second, the participant had to keep their muscle activation constant; when this was completed another ‘hold’ command was given. This last command signaled the participant that the isokinetic action was about to start. Within the next one or two seconds, the maximal activation started with the experimenter actually released the dynamometer head manually, and the participant was instructed to start a maximal action as rapidly and as forcefully as possible. For the experimenter most of the testing relied on the use of DataPac processing capabilities which allow for real time filtering of sEMG data, with the ability to flash an LED on a graphic interface upon attaining a specific threshold. Pilot work revealed that programming of 15 % of peak isometric sEMG optimized the ability to ensure pre-activation of 20 ± 5% MVIC at the beginning of the isokinetic action (see pilot work below).

Immediately following each action, torque and position data were low pass filtered ($4^{th}$ order Butterworth, Zero-Lag, 11 Hz) based on findings of power spectrum residual analyses and full wave rectified and low pass filtered for sEMG ($4^{th}$ order Butterworth, Zero-Lag, 3 Hz)(Winter 1990). The experimenter verified the 20 ± 5 % activation level in the 100 ms prior to the beginning of the isokinetic action. In order to train/familiarize the participant, she completed 5 to 10 successful actions, including pre-activation, release and maximal isokinetic action. A successful action was defined as starting with the average sEMG in the 100 ms prior to release between 15 to 25 % of the maximal isometric sEMG for the muscle of interest and followed by a maximal effort as described by the participant. Once the individual had performed three quadriceps and hamstrings actions successfully, drop jump testing was begun.

For the drop jump maneuver training, the participant stood on a 45 cm box, in a starting position with feet shoulder width apart, hips and knees extended, toes facing forward, equal weight on both feet and hands at ear level. The participant was then instructed on how to perform the task, and considered as trained upon completion of 5 consecutive successful trials. A trial was deemed successful if the participant:1) Slid off the box; 2) Landed with each foot on each force plate both prior to and following the maximal jump; 3) Produced a maximal effort during the propulsion phase; and 4) Kept their hands at ear level.

**Data Collection Session**

The participant was first equipped with sEMG as described on the familiarization day with the vastus medialis and Semitendinosus also being instrumented, and wore running shoes (Uraha 2, Adidas, USA). The participant started with a 5 minute warm up on a cycle ergometer at 50 W and completed the warm up as described above. Upon completion of this step the participant performed three, five seconds ramping MVICs of the quadriceps, first, and hamstrings, second, as described above. The peak sEMG values of the summed VL and VM during isometric knee extension and the summed BF and ST during isometric knee flexion efforts were recorded and averaged over the three repetitions. 20 % of this value was used as activation threshold for the strength measurements. After 1 minute of rest the participants were asked to perform three successful, as previously defined and verified each time, eccentric quadriceps ($270\ ° s^{-1}$) and concentric hamstrings actions ($90\ ° s^{-1}$) (counterbalanced order) in the passive mode of the Biodex 3. A rest period of 20 seconds was observed between each maximal action to allow for adequate recovery and minimize effects of fatigue or surface tissue warming to avoid changes in the frequency and amplitude characteristics of the sEMG signal (Konrad 2005). The specificity of the passive mode available on Biodex 3 system is that it acts upon the joint regardless of the effort provided by the participant and as such is easier for participants to understand, especially in the eccentric mode (Deighan, Croix et al. 2003). Dynamometer voltages representative of unprocessed knee joint angle, knee velocity, torque and vastus lateralis, vastus medialis, Biceps Femoris and Semitendinosus sEMGs were then exported in .csv format for further analysis.

<u>Development of Strength Protocol.</u> The strength testing procedure described above was chosen as a result of prior experiments that revealed the importance of controlling pre-activation levels during strength testing for the calculation of the sEMG/Torque relation. Specifically, we assessed the between day reliability of the strength testing procedures by having 11 individuals perform three eccentric quadriceps ($-270°s^{-1}$) and three concentric hamstrings ($90°s^{-1}$) actions on an isokinetic dynamometer (Biodex System 3) acting in passive mode. The tests were performed on two separate occasions 5-7 days apart. In accordance with the previously described methods, participants performed 3 maximal actions after a 5 minute warm up. Data were collected through DataPac and further exported as .csv files. The data were then imported into Matlab (R2008b, The MathWorks, Natick, MA) for further analysis, including filtering and gravity correction as recommended by the manufacturer. From these torque/time curves we extracted both peak torques and body weight normalized peak torques for the quadriceps and hamstrings, the average

of the three trials was used for each day. Over the two days those values of the absolute peak torque were tested for consistency (ICC 2,1), and precision (standard error of measurement). The consistency ranged between 0.89 and 0.95 and the precision between 4.4 and 12.3 Nm. (Table 1)

**Table 1. Interday Reproducibility and Precision of Peak Torque**

| Peak Torque | Day 1 Mean±SD | Day 2 Mean±SD | ICC | SEM |
|---|---|---|---|---|
| Quadriceps (Nm) | 180±54 | 165±50 | 0.94 | 12.3 |
| Hamstrings (Nm) | 53±19 | 55±20 | 0.95 | 4.4 |
| Normalized quadriceps (Nm/kg) | 20.5±0.5 | 20.3±0.5 | 0.89 | 0.2 |
| Normalized hamstrings (Nm/kg) | 0.7±0.2 | 0.8±0.2 | 0.92 | 0.1 |

Despite the satisfactory consistency and precision of torque data, we found that pre-activation levels, i.e. the activation occurring prior to the release of the dynamometer level arm, varied widely, both amongst participants and between occasions. Therefore, we subsequently had 10 participants visit the laboratory on two separate occasions in a pilot study attempting to control preactivation. During the first (familiarization) visit they were introduced to the laboratory environment, signed informed consent forms and were first trained to perform isometric knee flexion and extension, hamstrings concentric and quadriceps eccentric actions. They were then instructed to pre-activate as described previously to 20 % of the peak sEMG observed during their MVIC by gradually increasing pre-activation in the muscle group of interest, which was visually monitored on screen via visual display. When the participant could perform the ramping satisfactorily, she performed 3-5 repetitions of the complete action including the maximal isokinetic action (initiated manually by the experimenter when the LED flashed upon reaching 15 % of MVIC sEMG amplitude). On the second visit the participant was fully equipped with sEMG electrodes as described above including VL and VM representing the quadriceps and BF and ST representing the hamstrings. The participant was then positioned on the Biodex, as in the first session, and performed isometric and isokinetic actions as previously described. Three successful trials were saved and stored for further analysis. The average filtered and normalized sEMGs over the 100 ms prior to the release were then identified in Matlab (R2008b, The MathWorks, Natick, MA). Those values, for the three repetitions, were tested for within-day consistency (ICC 3,1) and precision (SEM). The results presented below (Table 2) suggest that the set pre-activation values obtained from our protocol could be precisely implemented. ICC values showed moderate to low values, which are in large part likely due to the fact that the measures studied were inherently inducing very low between subject variability due to the fixed outcome. Overall, the relatively similar means and the small standard deviations across repetitions provides evidence of a consistent measure.

**Table 2. Preactivation Values (ICC, SEM)**

| | Rep 1 %MVIC Mean±SD | Rep 2 %MVIC Mean±SD | Rep 3 %MVIC Mean±SD | ICC | SEM %MVIC |
|---|---|---|---|---|---|
| | | | **Quadriceps** | | |
| 90 $°s^{-1}$ | 21±5 | 23±8 | 20±4 | 0.67 | 30.7 |
| 180 $°s^{-1}$ | 21±4 | 19±5 | 21±3 | 0.56 | 20.8 |
| 270 $°s^{-1}$ | 19±5 | 19±4 | 19±4 | 0.38 | 30.4 |
| | | | **Hamstrings** | | |
| 90 $°s^{-1}$ | 21±5 | 18±4 | 20±3 | -0.28 | 50.2 |
| 180 $°s^{-1}$ | 20±4 | 22±3 | 20±3 | 0.07 | 30.2 |
| 270 $°s^{-1}$ | 21±5 | 20±4 | 21±4 | 0.55 | 30.3 |

This was further tested by calculating the limits of agreement as described by (Bland and Altman 1986). Figure 6 reveals that there a small bias towards greater values than 20 %, overall most preactivation values fall within the ± 5% window.

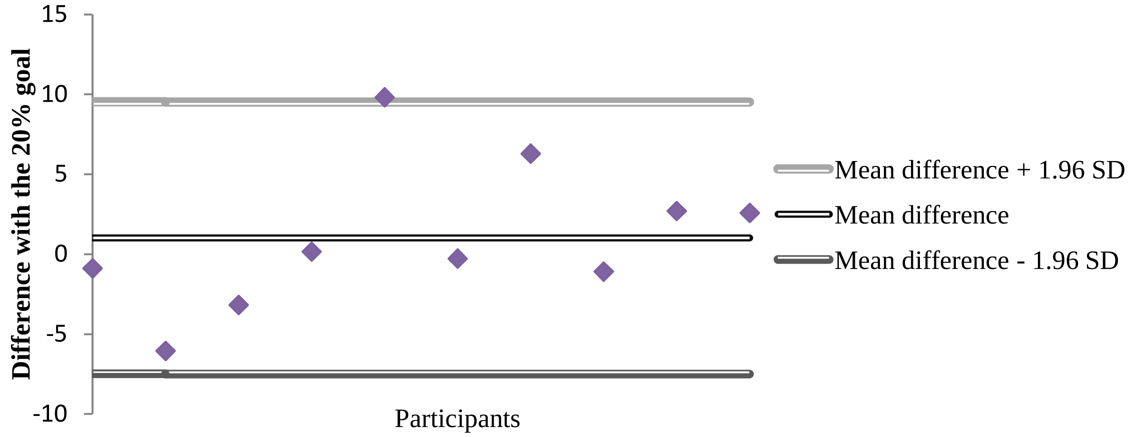



Figure 6: *Preactivation values, Limits of agreement.*
*The agreement between the actual pre-activation values and the sought value (20 %). The limits of agreement are calculated as the average difference over three repetitions with reference to the 20 % goal.*

## Drop Jump Maneuver

Upon completion of the strength testing, the participant performed drop jump testing. With sEMG electrodes attached, twenty LED sensors (Impulse, PhaseSpace, San Leandro, CA) (four per segment) were secured to the foot, tibial shaft, the lateral thigh, and sacrum to obtain 3D positions and orientation of each rigid segment. A segmental reference system defined body segments with the positive X-axis defined as the posterior to anterior axis; positive Y-axis defined as the distal to proximal longitudinal axis; and positive Z-axis defined as the left right axis. The ankle joint center was determined by the midpoint between the medial and lateral malleoli, the knee joint center by the midpoint between the medial and lateral joint line, and the hip joint center was determined by the rotation method (Leardini, Cappozzo et al. 1999). Vertical ground reaction force data was collected at 1000 Hz with a Bertec force plate (model 4060-NC; Bertec Corporation, Columbus, OH). Following the methods described on the familiarization day, the participant completed five successful trials during which complete biomechanical data was collected. The time synchronized kinematic, kinetic, and sEMG data were stored and further exported as .exp files.

## Data Reduction

All further data reduction and analysis were performed in Matlab (R2008b, The MathWorks, Natick, MA) using proprietary algorithms. The .csv and .exp files were imported into a database created using Matlab (R2008b, The MathWorks, Natick, MA) and further processed within that database.

### The sEMG/Torque Ratio

#### *Isometric Calibration Actions*

Torque (Figure 7) from isometric actions was low pass filtered ($4^{th}$ order Butterworth, Zero-Lag, 11 Hz).

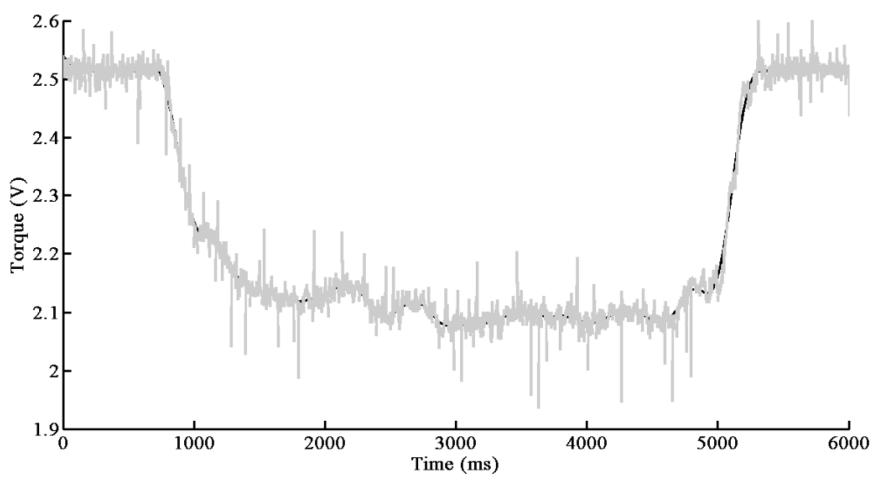



Figure 7: *Raw and Filtered Dynamometer Voltage Indicative of Isometric Quadriceps Torque. Processing of the voltage signal representative of the torque produced. Prior to (Dark grey line) and following (Light grey line) low pass filtering ($4^{th}$ order Butterworth, Zero-Lag, 11 Hz).*

Filtered torque data (V) was then converted to Nm according to manufacturer's calibrations and gravity corrected. For each isometric action the peak torque produced was recorded. The sEMG data was also processed (Figure 8) using full wave rectification, band pass filtering (4$^{th}$ order Butterworth, Zero-Lag, 10-300Hz) and RMS smoothing (100 ms constant) (Shultz, Nguyen et al. 2009). Peak sEMG values for VM and VL were recorded from isometric extension actions and peak BF and ST recorded from isometric flexion actions. The isometric sEMG/Torque ratio was calculated for each isometric knee extension action by dividing the peak VL and VM sEMG amplitudes by half of the peak extension torque produced, and, for isometric knee flexion efforts, by dividing BF and ST peak sEMG amplitudes by half of the peak flexion torque produced. The decision to assume equal contribution from the two heads of the quadriceps to the knee extensor torque was based on previous findings that demonstrate this occurrence in vitro (Lieb and Perry 1968). The four isometric sEMG/Torque values were then averaged over the three repetitions and retained for further estimation of the muscle torques during the impact phase of the initial landing of a drop jump maneuver using the isometric sEMG/Torque ratio.

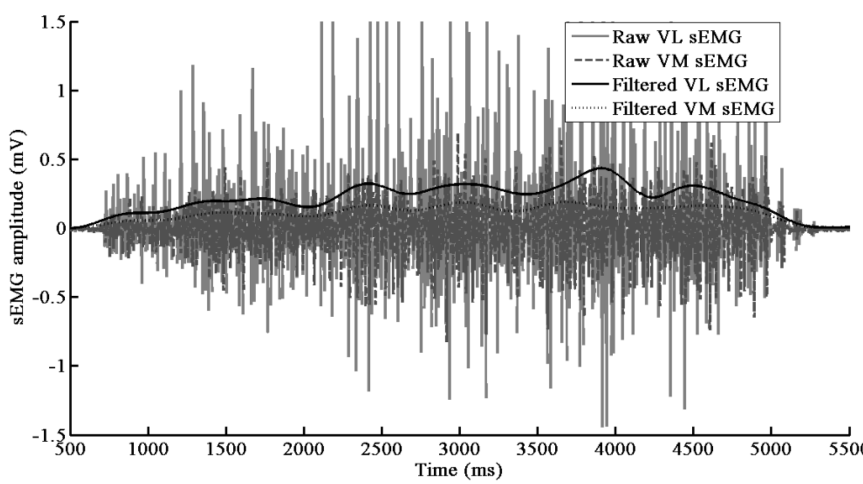



Figure 8: *Raw and Filtered Quadriceps (VL And VM) sEMG During an Isometric Action.*
*Processing of the sEMG, raw (Grey Lines) and processed (Black lines), bandpass fitler (4$^{th}$ order Butterworth, Zero-lag, 10-300 Hz), and RMS smoothing (100 ms constant).*

***Isokinetic Calibration Actions***

Torque (Figure 9) and position data from the isokinetic calibration actions were low pass filtered ($4^{th}$ order Butterworth, Zero-Lag, 11 Hz). After filtering, and accounting for AC baseline, the torque data was converted to Nm according to manufacturer's specifications.

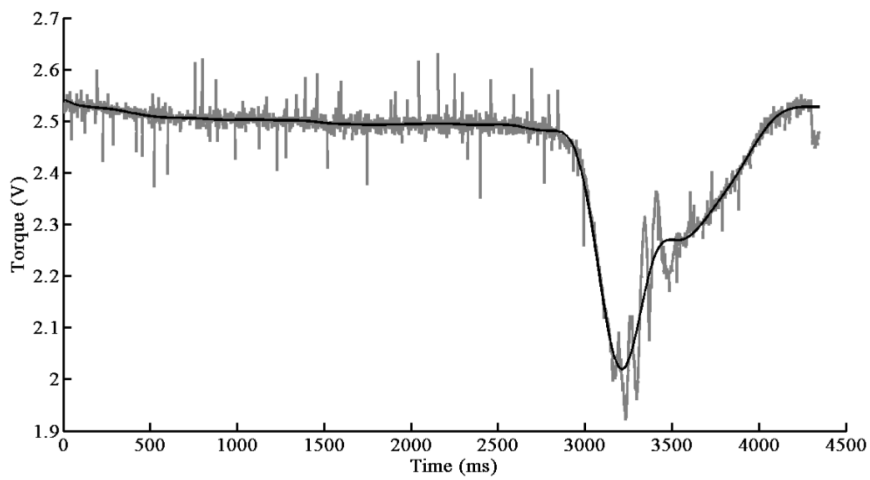



Figure 9: *Raw and Filtered Dynamometer Voltage Indicative of Eccentric Quadriceps Torque ($270°s^{-1}$).*
*Processing of the voltage signal representative of the torque produced. Prior to (Light grey line) and following (Black line) low pass filtered ($4^{th}$ order Butterworth, Zero-Lag, 11 Hz).*

Raw voltage indicative of position was low pass filtered ($4^{th}$ order Butterworth, Zero-Lag, 11 Hz), converted to degrees and used to calculate the gravity correction according to manufacturer's specifications. Subsequently, the position data was used to truncate isokinetic actions defined as the knee reaching ten (beginning) and 80 degrees of knee flexion (end). The torque data were then gravity corrected, and truncated as illustrated in Figure 10.

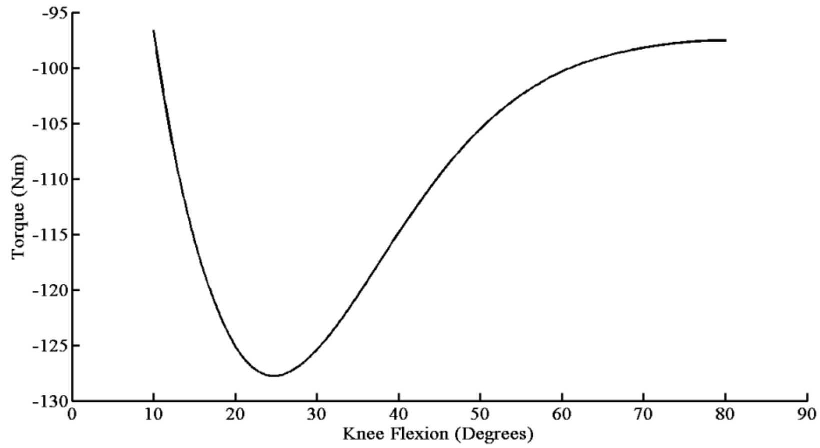


Figure 10: *Gravity Corrected and Truncated Eccentric Quadriceps Torque (Nm)*

The sEMG data from the dynamic actions were processed using full wave rectification and low pass filtering ($4^{th}$ order Butterworth, Zero-Lag, 4Hz)(Winter 1990). The sEMG data was truncated in the same way as the torque data (based on position data), and finally normalized to the peak MVIC sEMG used in the calculation of the isometric sEMG/Torque ratio.

Then, separately for each of the four muscles studied, the angle and action specific sEMG/Torque ratio was calculated by dividing, for each data point, the processed sEMG by half of the torque produced (Figure 12). This data was then linearly extrapolated to render values of the angle and action specific sEMG/Torque ratio between 0 and 100 degrees of knee flexion.

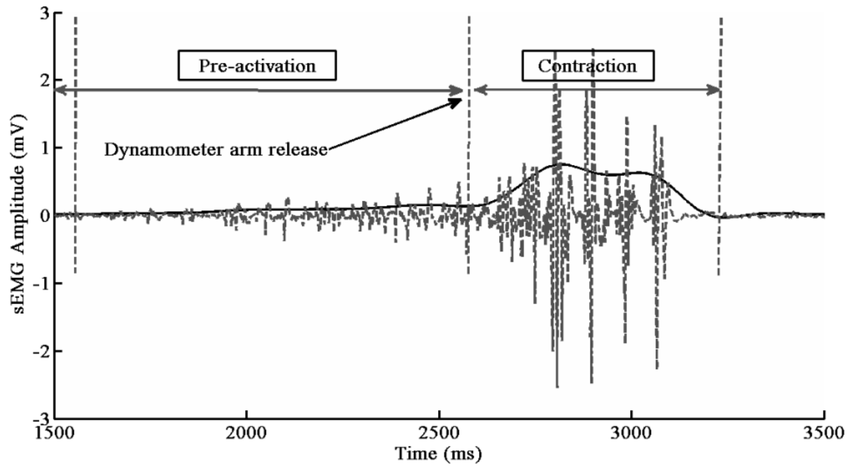



Figure 11: *Representative sEMG Processing.*
*Raw (Light Grey line) and rectified filtered (Dark Grey line) sEMG as a function of time during an isokinetic calibration action.*

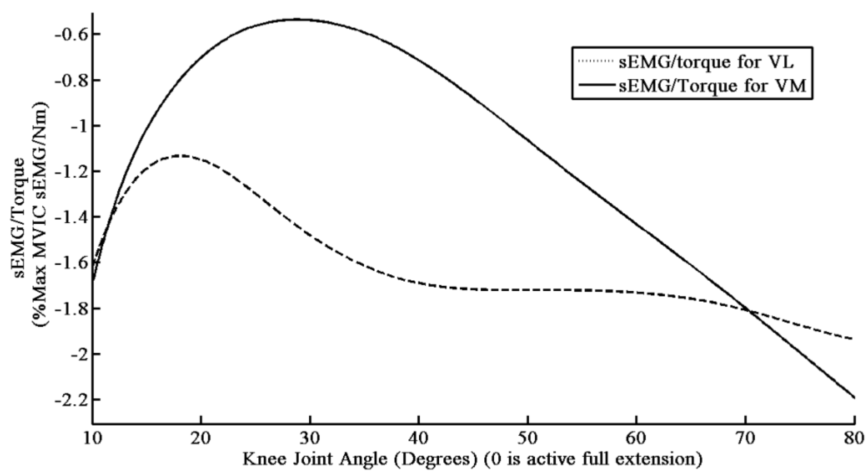



Figure 12: *sEMG/Torque Ratios for vastus lateralis and vastus medialis.*

***Internal validation of the angle specific sEMG/Torque ratio***

Internal validity of the angle and action specific sEMG/Torque ratio (ability of using the above described angle and action specific sEMG/Torque ratio to estimate muscle torque and predict the actual dynamometer measured torque) was previously assessed. After processing the sEMG and torque as previously described, we modeled the resulting curve as a function of the knee joint angle using a 2nd order polynomial (Doorenbosch and Harlaar 2003). This function represents, for each knee joint angle, the percentage of the peak MVIC sEMG necessary to create one Nm (Figure 12). Then, for each repetition, we interpolated the sEMG/Torque ratio based on knee joint angle data collected during the isokinetic calibration action (Eccentric quadriceps (-270°$s^{-1}$) and concentric hamstrings (90°$s^{-1}$). Then we divided the processed sEMG by the angle and action specific sEMG/Torque ratio to estimate knee extensor (using VL and VM), or flexor (using BF and ST), torque. These torque curves estimated based on the angle and action specific sEMG/Torque ratio were then compared to the measured (output from Biodex) torque curves as described in Figure 13.

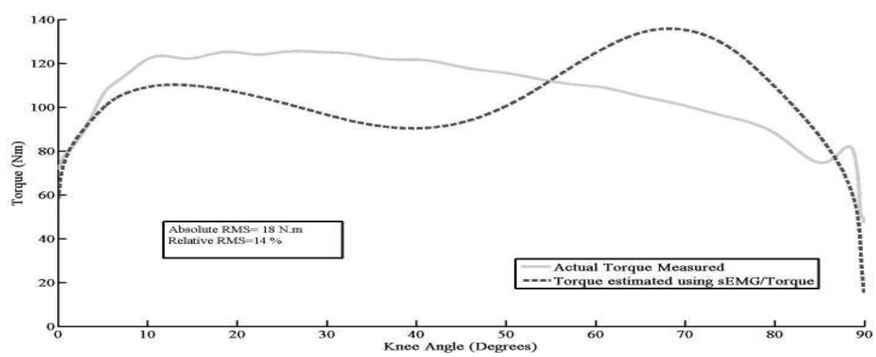



Figure 13: *Representative Data Curve for Actual and Estimated Torques (Hamstrings Concentric).*
*Measured (grey line) and estimated (dotted line) torques during dynamometer testing.*

The root mean square values (relative error) are presented in Table 3 and fall within the range reported previously in the literature for concentric actions included in Table 3 for comparison.

**Table 3. sEMG/Torque Relation Polynomial and Root Mean Square Error**

| Torques (Nm) | Absolute Error (Nm) | Relative Error (%) |
|---|---|---|
| **Quadriceps** | | |
| Ecc 90 $°s^{-1}$ | 32±13 | 12±4 |
| **Con 120*°s^-1^* | *70.6±20.5* | *11±3* |
| **Con 150*°s^-1^* | *110.8±60.2* | *19±9* |
| Ecc 180$°s^{-1}$ | 32±17 | 13±4 |
| Ecc 270$°s^{-1}$ | 32±13 | 14±3 |
| **Hamstrings** | | |
| Con 90$°s^{-1}$ | 12±3 | 14±3 |
| **Con 120*°s^-1^* | *50.8±10.7* | *14±4* |
| **Con 150*°s^-1^* | *80.9±30.7* | *20±7* |
| Con 180$°s^{-1}$ | 12±6 | 15±4 |
| Con 270$°s^{-1}$ | 11±3 | 17±4 |

*Numbers in *Italic* are from Doorenbosch et al. (2004).

### Reduction of the Biomechanical Data

Three dimensional knee joint angles were calculated using Euler angle definitions with a rotational sequence of Z Y' X" (Kadaba, Ramakrishnan et al. 1989). Raw kinematic data was linearly interpolated to force-plate data and subsequently low-pass filtered (4$^{th}$ order Butterworth, Zero-Lag, 12 Hz). Knee intersegmental forces and moments were calculated using an inverse dynamics analysis within the MotionMonitor software (Innovative Sports Training, Chicago, IL). Successful trials were exported (.exp format) including knee kinematics and kinetics in the coronal, transverse, and sagittal planes, vertical, medial/lateral and anterior/posterior ground reaction forces and unprocessed VL, VM, BF and ST sEMGs.

Further data processing took place in Matlab (R2008b, The MathWorks, Natick, MA). The impact phase of the initial deceleration was defined from the point where the vertical ground reaction force exceeded 10N, to peak in vertical ground reaction force (F2) as exemplified in Figure 14.

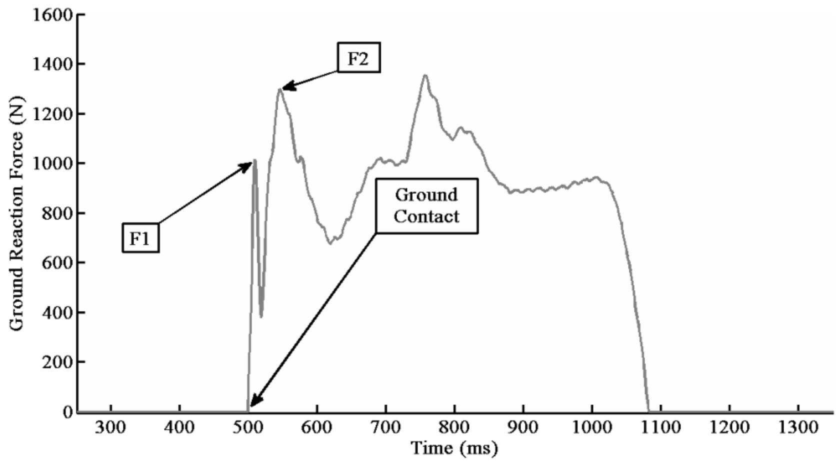



Figure 14: *Ground Reaction Force During Landing.*
*Identification of the beginning (ground contact) and end (2 peak in ground reaction force:F2) of the impact phase from the vertical ground reaction force data.*

Over this impact phase, 3D knee biomechanics including moment impulses and kinematic excursions in 3 planes, as well as peak anterior knee shear force were calculated and averaged over 5 trials. (Table 4) This rendered seven variables indicative of high risk knee joint mechanics for each participant; which were retained for further statistical analysis and are described thereafter.

**Table 4. Descriptors of calculated independent variables.**

| Variable | Calculation | Name (Unit) | Direction |
|---|---|---|---|
| **Flexion excursion** | Difference between knee flexion angle at landing and at the end of the impact phase of the initial landing of a drop jump maneuver | KFE (°) | Flexion (-) Extension (+) |
| **Peak extension moment** | Maximal value of the net internal knee extensor moment as calculated through an inverse dynamics analysis during the impact phase of the initial landing of a drop jump maneuver | PKEM (Nm) | |
| **Peak anterior shear force** | Maximal value of the anterior shear force during the impact phase of the initial landing of a drop jump maneuver as calculated through an inverse dynamics analysis | PASF (N) | Anterior (+) Posterior (-) |
| **Frontal plane excursion** | Difference between knee adduction/abduction angle at landing and at the end of the impact phase of the initial landing of a drop jump maneuver | KVE (°) | Valgus rot. (+) Varus rot. (-) |
| **Peak frontal plane moment** | Identified as the maximal value of the net abduction moment as calculated through an inverse dynamics analysis during the impact phase of the initial landing of a drop jump maneuver | PKVM (Nm) | |
| **Transverse plane excursion** | Calculated as the difference between knee internal/external rotation angle at landing and at the end of the impact phase of the initial landing of a drop jump maneuver | KRE (°) | External rot. (+) External rot. (-) |
| **Peak transverse plane moment** | Maximal value of the net internal/external rotation moment as calculated through an inverse dynamics analysis during the impact phase of the initial landing of a drop jump maneuver | PKRM (Nm) | |

## Drop Jump Data and Thigh Muscles Torques Estimation

### *Muscle Torque Estimation Based on the Isometric sEMG/Torque Ratio*

For the muscle torque estimation based on the isometric sEMG/Torque ratio the non normalized sEMG amplitudes of the sEMGs collected during the drop jump were used. The sEMG was band pass filtered (4$^{th}$ order Butterworth, Zero-Lag, 10-300 Hz) full wave rectified, and then RMS smoothed (25 ms constant) was calculated (Shultz, Nguyen et al. 2009). The isometric sEMG/Torque ratio used was unique over the whole impact phase. To estimate the muscle torques for each of the four heads the isometric sEMG/Torque ratio was divided by the amplitude of the processed sEMG for each data point during the impact phase. The impulse of those muscle torques estimated using the isometric sEMG/Torque ratio ($\underline{VL_{ISO}}$ :Vastus Lateralis torque impulse, $\underline{VM_{ISO}}$: vastus medialis torque impulse, $\underline{BF_{ISO}}$: bicep femoris torque impulse, $\underline{ST_{ISO}}$ :semitendinous torque impulse) were used to calculate the net knee joint moment impulse ($\underline{NET_{ISO}}$).

To address Hypothesis 1 a, the net knee joint moment impulse, calculated as the difference between the summed estimated quadriceps torques impulses ($\underline{VL_{ISO}}$ and $\underline{VM_{ISO}}$) and the summed estimated hamstrings torques impulses ($\underline{BF_{ISO}}$ and $\underline{ST_{ISO}}$), was averaged over five trials and retained as a raw value ($\underline{NET_{ISO}}$).

To address Hypothesis 1 b the four estimated muscle torque impulses were averaged over 5 trials. These four values ($\underline{VL_{ISO}}$, $\underline{VM_{ISO}}$, $\underline{BF_{ISO}}$, $\underline{ST_{ISO}}$) were retained for further statistical tests.

***Muscle Torque Estimation Based on the Angle and Action Specific sEMG/Torque Ratio***

For the purpose of thigh muscles torque estimation during the impact phase of landing using the angle and action specific sEMG/Torque ratio, sEMG data were full wave rectified, low pass filtered (4$^{th}$ order Butterworth, Zero-Lag, 4 Hz) (Figure 15) and normalized to the max sEMG during MVIC actions as previously recorded. The determination of the angle specific sEMG/Torque ratio during the impact phase of the initial landing of a drop jump maneuver was based on the knee joint angle observed during the impact phase of the initial landing of a drop jump maneuver. To achieve this, we relied on a direct look-up technique in which Matlab extracts the sEMG/Torque ratio from the actual values calculated, as presented above (c.f. Isokinetic calibrations actions), during eccentric actions for the quadriceps and concentric actions for the hamstrings. For each data point (i.e. Knee flexion angle) during the impact phase, Matlab refers to the reference array (sEMG/Torque ratio for each knee flexion angle between 0 and 90° of flexion) and extract the “y” (sEMG/Torque ratio) value for the specified “x” (Knee joint angle during the impact phase of the initial landing of a drop jump maneuver). This allows for a direct estimation of the sEMG/Torque coefficient rather an interpolation based on a polynomial (Figure 16).

Then for each data point during the impact phase of the initial landing of a drop jump maneuver, VL, VM, BF and ST torques were estimated by dividing the processed sEMG value by the simultaneous sEMG/Torque ratio.

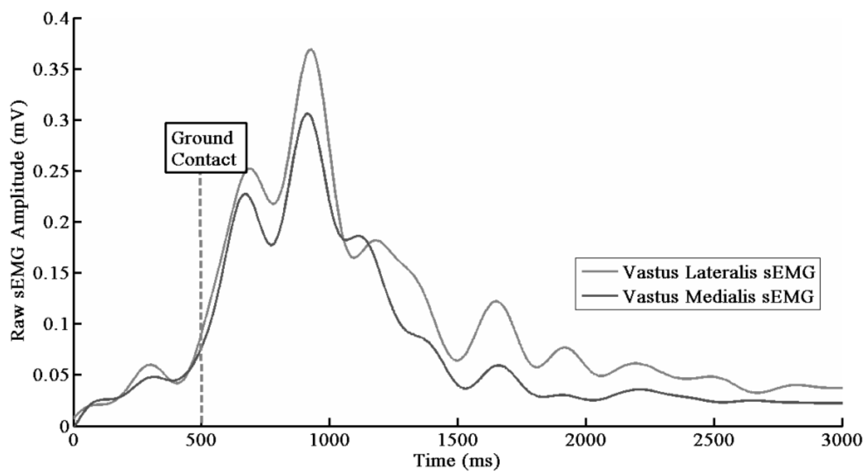



Figure 15: *Quadriceps sEMG During the Drop Jump Maneuver.*
*Representative non normalized filtered sEMG for the vastus medialis (Light Grey Line), vastus lateralis (Dark Grey Line) during a drop jump maneuver.*

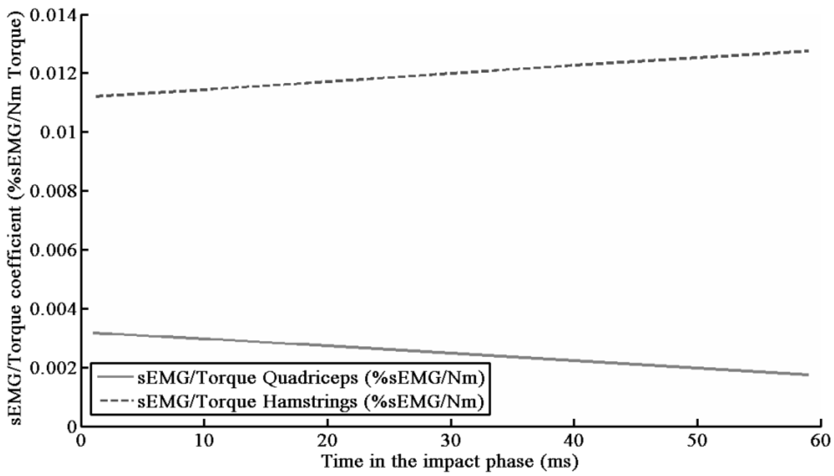



Figure 16: *Quadriceps and Hamstrings sEMG/Torque Coefficients During the Impact Phase. sEMG/Torque coefficients for the quadriceps (Light Grey Line) and the hamstrings (Dark Grey Line) as determined based on knee flexion position during the impact phase of the initial deceleration of the drop jump maneuver and sEMG/Torque relation calculated during calibration actions.*

The impulse of those muscle torques estimated using the angle and action specific sEMG/Torque ratio ($VL_{DYN}$: vastus lateralis torque impulse, $VM_{DYN}$: vastus medialis torque impulse, $BF_{DYN}$: bicep femoris torque impulse, $ST_{DYN}$ :semitendinous torque impulse) were used to calculate the net knee joint moment impulse ($NET_{DYN}$).

For Hypothesis 1 a The Net Knee Joint Moment Impulse ($NET_{DYN}$) calculated based on thigh muscle torques estimated using the angle and action specific sEMG/Torque ratio was averaged over five trials and retained as a raw value for further statistical tests.

For Hypotheses 1 b, 2 b, 2 c, 2 e and 2 g: For the four muscles studied, the torque impulses estimated using the angle and action specific sEMG/Torque ratio (**$VL_{DYN}$, $VM_{DYN}$, $BF_{DYN}$, $ST_{DYN}$**, all Nm*s) were used in the analysis.

For Hypotheses 2a, 2d and 2f: For the four muscles studied, the body weight and height normalized torque impulses estimated using the angle and action specific sEMG/Torque ratio (**$VL_{DYNBWH}$, $VM_{DYNBWH}$, $BF_{DYNBWH}$, $ST_{DYNBWH}$**, all Nm*s/$BW^{-1}$/$Ht^{-1}$) were used in the statistical analysis.

## Statistical Analyses

### Power Calculations

Based on the correlation between independent and dependent variables collected during pilot testing of the protocol (N=5), we found that the average correlations between Impact phase VM torque impulse, Impact phase VL torque impulse, Impact phase hamstrings ST torque impulse, Impact phase hamstrings BF torque impulse and the dependent variables was 0.3. Using those values for the correlations in G*Power version 30.10.2 (Faul, Universität Kiel, Germany) rendered an effect size of 1.17. With six predictors, G power estimated that 25 participants will render 95 % power at $\alpha = 0.05$.

Since the preliminary correlation data was calculated based on 5 subjects only, we explored the statistical power that would be gained from increasing the number of participants. To that effect, and for exploratory purposes only, we increased the number of participants to 50 in G*Power and observed the resultant effect on the power of the analysis. Figure 17 shows that 30 participants will render approximately 98% of power.

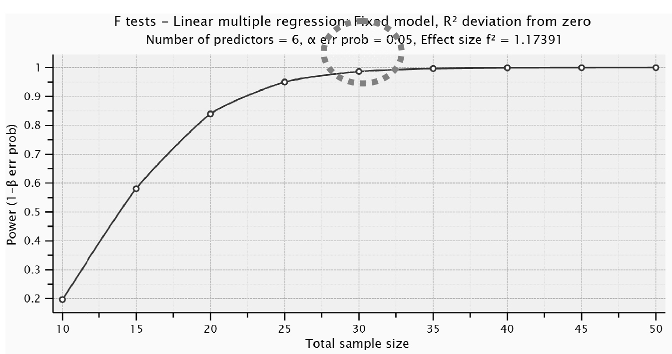



Figure 17: *Estimated Power as a Function of Sample Size.*
*Predicted power as a function of the sample size, with an n of 30 showing 98 % power.*

**Hypothesis Testing**

To address our first hypothesis, that the net sagittal plane joint moment impulses calculated (quadriceps minus hamstrings) based on vastus lateralis, vastus medialis, bicep femoris and semitendinous torque impulses estimated with the isometric sEMG/Torque ratio, will be lesser than that estimated using an inverse dynamics analysis or the angle and action specific sEMG/Torque ratio during the impact phase of the initial deceleration of a drop jump maneuver, we first performed a 1 (Net Knee Joint Moment Impulse) by 3 (Mean of estimation: Inverse dynamics (KEM), Isometric sEMG/Torque ratio ($NET_{ISO}$), angle and action specific sEMG/Torque ratio ($NET_{DYN}$)) Repeated Measures ANOVA. Then, we performed a 4 (Muscles: VL,VM,BF,ST) by 2 (Means of estimation: Isometric sEMG/Torque ratio ($VL_{ISO}$, $VM_{ISO}$, $BF_{ISO}$, $ST_{ISO}$) and angle and action specific sEMG/Torque ratio (**$VL_{DYN}$, $VM_{DYN}$, $BF_{DYN}$, $ST_{DYN}$**) RMANOVA. Significance value was set a priori at $P \leq 0.05$. To further investigate main effects and interactions we performed pair sampled t-test comparisons using Bonferoni correction.

To address our second hypothesis, that during the impact phase of the initial deceleration of a drop jump maneuver, greater vastus lateralis and vastus medialis and lesser bicep femoris and semitendinous torque impulses, estimated using the angle and action specific sEMG/Torque ratio, will predict greater magnitudes of variables indicative of high-risk knee joint mechanics in the three planes of motion, we fitted separate linear regression models to examine the extent to which vastus lateralis (**$VL_{DYN}$**), vastus medialis (**$VM_{DYN}$**), Bicep femoris (**$BF_{DYN}$**) and semitendinous (**$ST_{DYN}$**) torque impulses predicted high-risk knee joint mechanics in the three planes of motion.

In the sagittal plane we studied:

a) The extent to which raw vastus lateralis (**$VL_{DYN}$**), vastus medialis (**$VM_{DYN}$**), Bicep femoris (**$BF_{DYN}$**) and semitendinous (**$ST_{DYN}$**) torque impulses estimated using the angle and action specific sEMG/Torque ratio, predicted knee flexion excursion (KFE) after accounting for body weight and height.

b) The extent to which raw vastus lateralis (**VL**$_{DYN}$), vastus medialis (**VM**$_{DYN}$), Bicep femoris (**BF**$_{DYN}$) and semitendinous (**ST**$_{DYN}$) torque impulses, estimated using the angle and action specific sEMG/Torque ratio, predicted the peak internal knee extension moment (PKEM) after accounting for body weight and height.

c) The extent to which raw vastus lateralis (**VL**$_{DYN}$), vastus medialis (**VM**$_{DYN}$), Bicep femoris (**BF**$_{DYN}$) and semitendinous (**ST**$_{DYN}$) torque impulses, estimated using the angle and action specific sEMG/Torque ratio, predicted peak anterior shear force (PASF) after accounting for body weight.

In the frontal plane we studied:

d) The extent to which body weight and height normalized vastus lateralis (**VL**$_{DYNBWH}$), vastus medialis (**VM**$_{DYNBWH}$), Bicep femoris (**BF**$_{DYNBWH}$) and semitendinous (**ST**$_{DYNBWH}$) torque impulses, estimated using the angle and action specific sEMG/Torque ratio, predicted knee abduction excursion (KVE).

e) The extent to which raw vastus lateralis (**VL**$_{DYN}$), vastus medialis (**VM**$_{DYN}$), Bicep femoris (**BF**$_{DYN}$) and semitendinous (**ST**$_{DYN}$) torque impulses, estimated using the angle and action specific sEMG/Torque ratio, predicted the peak internal knee abduction moment (PKVM) after accounting for body weight and height.

In the transverse plane we studied:

f) The extent to which raw vastus lateralis (**VL**$_{DYN}$), vastus medialis (**VM**$_{DYN}$), Bicep femoris (**BF**$_{DYN}$) and semitendinous (**ST**$_{DYN}$) torque impulses, estimated using the angle and action specific sEMG/Torque ratio, predicted knee internal rotation excursion (KRE) after accounting for body weight and height.

g) The extent to which raw vastus lateralis (**<u>VL</u>**$_{DYN}$), vastus medialis (**<u>VM</u>**$_{DYN}$), Bicep femoris (**<u>BF</u>**$_{DYN}$) and semitendinous (**<u>ST</u>**$_{DYN}$) torque impulses, estimated using the angle and action specific sEMG/Torque ratio, predicted the peak internal knee internal rotation moment (PKRM) after accounting for body weight and height.

# CHAPTER IV

# RESULTS

Forty-three females successfully completed data collection. However, data on three participants was eliminated for technical issues in data collection. Therefore data from forty participants (Age=21.2±1.5 yrs, Height=164.2±7.3 cm, Mass=60±8 kg, BMI= 22±2 kg/m²) were used for analyses. Mean±SD and range (minimum to maximum) for variables of net muscle torques, estimated using both the isometric and angle and action specific sEMG/Torque ratios, and joint moments impulses using the isometric sEMG/Torque, angle and action specific sEMG/Torque and inverse dynamics methods are presented in Table 5 while dependent and independent descriptives during the impact phase of the initial landing of a drop jump maneuver are listed in Table 6 and a correlation table is provided is Table 7.

## Hypothesis 1: Differences in Net Joint Estimation Using Isometric or Action Specific sEMG/Torque

Hypothesis 1 a)

A 1 (Net Knee Joint Moment Impulse) by 3 (Mean of estimation: Inverse dynamics (KEM), Isometric sEMG/Torque ratio ($NET_{\underline{ISO}}$), angle and action specific sEMG/Torque ratio ($NET_{\underline{DYN}}$) Repeated Measures ANOVA revealed a significant effect of means of estimation ($P<0.001$). Post-hoc tests showed significantly lower ($P<0.001$) estimation of net knee joint moment using the isometric sEMG/Torque method ($NET_{ISO}$) (Mean 3.30 ± 1.87 Nm*s) compared to the angle and action specific sEMG/Torque method ($NET_{DYN}$) (Mean 4.29 ± 2.18 Nm*s) or inverse dynamics analysis (KEM) (Mean 4.46 ± 1.91 Nm*s). However, there was no difference between $NET_{DYN}$ and KEM (P = 0.93).

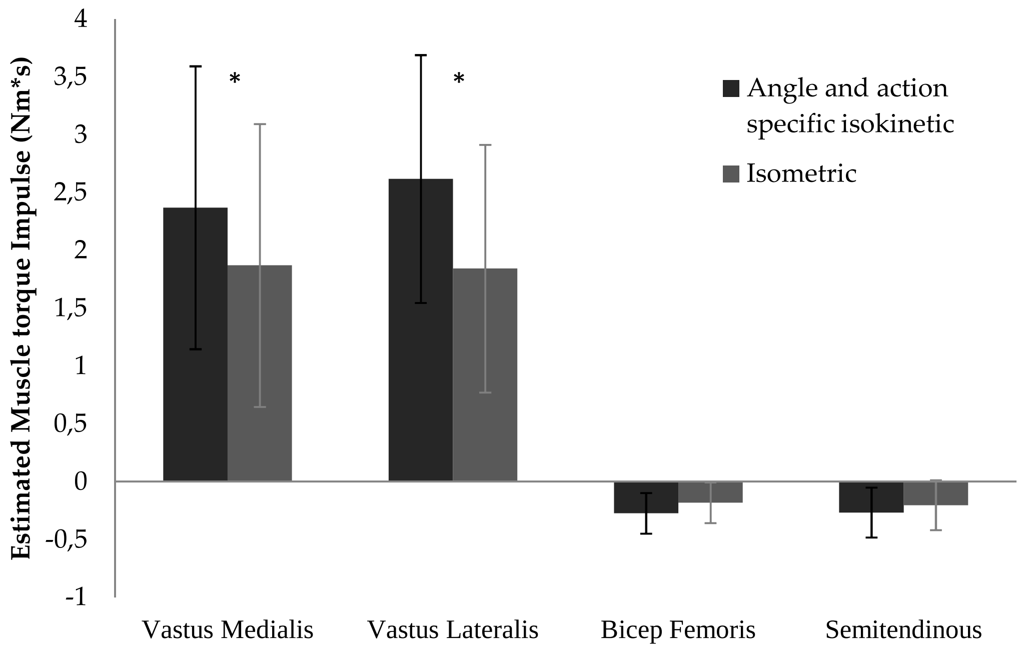
Hypothesis 1 b)

A 4 (Muscles: VL,VM,BF,ST) by 2 (Means of estimation: Isometric sEMG/Torque ratio ($VL_{ISO}$, $VM_{ISO}$, $BF_{ISO}$, $ST_{ISO}$); angle and action specific sEMG/Torque ratio ($VL_{DYN}$, $VM_{DYN}$, $BF_{DYN}$, $ST_{DYN}$) RMANOVA demonstrated a significant interaction between muscle and mean of estimation (P<0.01). Posthoc testing showed that this interaction, as shown in Figure 18, was due to $VL_{ISO}$ (Mean 1.84 ± 1.22 Nm*s) and $VM_{ISO}$ (Mean 1.85 ± 1.08 Nm*s) being lower than $VL_{DYN}$ (Mean 2.23 ± 1.26 Nm*s) and $VM_{DYN}$ (Mean 2.60 ± 1.48 Nm*s), with no difference between $BF_{ISO}$ (Mean -0.18 ± 0.18 Nm*s), $ST_{ISO}$ (Mean -0.20 ± 0.22 Nm*s), $BF_{DYN}$ (Mean -0.28 ± 0.29 Nm*s) and $ST_{DYN}$ (Mean 0.26 ± 0.21 Nm*s).

**Table 5. Net Joint Moment and Muscle Torque Impulses**

| **Estimated Net Moment Impulses (Nm*s)** | **Mean ± SD** | **Range** |
|---|---|---|
| Based on isometric sEMG/Torque ($NET_{ISO}$)* | 3.30 ± 1.87 | 0.98 - 8.29 |
| Based on dynamic sEMG/Torque ($NET_{DYN}$) | 4.29 ± 2.18 | 1.46 - 9.64 |
| Based on inverse Dynamics (KEM) | 4.46 ± 1.91 | 1.16 - 9.37 |
| **Muscle torque impulses (Nm*s)** | | |
| *Estimated using the angle and action specific sEMG/Torque ratio* | | |
| Vastus lateralis ($VL_{DYN}$) | 2.23 ± 1.26 | 0.69 - 7.14 |
| Vastus medialis ($VM_{DYN}$) | 2.60 ± 1.48 | 0.37 - 7.96 |
| Bicep femoris ($BF_{DYN}$) | - 0.28 ± 0.29 | -1.26 - -0.02 |
| Semitendinous ($ST_{DYN}$) | -0.26 ± 0.21 | -1.07 - -0.07 |
| *Estimated using the isometric sEMG/Torque ratio* | | |
| Vastus lateralis ($VL_{ISO}$) | 1.84 ± 1.22 | 0.46 - 5.11 |
| Vastus medialis ($VL_{ISO}$) | 1.85 ± 1.08 | 0.19 - 4.87 |
| Bicep femoris ($BF_{ISO}$) | -0.18 ± 0.18 | -0.72 - -0.01 |
| Semitendinous ($ST_{ISO}$) | -0.20 ± 0.22 | -0.98 - -0.02 |

* $NET_{DYN}$ & KEM >$NET_{ISO}$ (P<0.05)




* Indicates Isometric sEMG/Torque ratio significantly (P<0.05) lower than angle and action specific sEMG/Torque ratio

Figure 18: *The Interaction Between Muscle Studied and Means of Torque Estimation.*
*Angle and contraction specific sEMG/Torque ratio estimates of torque impulses are greater than those obtained using the isometric sEMG/Torque ratio*

## Hypothesis 2: Contribution of Muscle Torque Impulses to High Risk Mechanics

To allow the reader an overview of the data, descriptive statistics and a correlation matrix are respectively provided in tables 6 and 7.

**Table 6. Means ± SD and Range for Dependent and Independent Variables**

All variables are observed over the impact phase of the initial landing of a drop jump maneuver.

| **Variables** | **Mean ± SD** | **Range** |
|---|---|---|
| Muscle Torque Impulses estimated using the angle and action specific sEMG/Torque ratio (Nm*s) | | |
| Vastus lateralis ($VL_{DYN}$) | 2.10 ± 1.18 | 0.67 - 6.56 |
| Vastus medialis ($VM_{DYN}$) | 2.49 ± 1.41 | 0.38 - 7.23 |
| Bicep femoris ($BF_{DYN}$) | -0.33 ± 0.39 | -1.88 - -0.02 |
| Semitendinous ($ST_{DYN}$) | -0.23 ± 0.24 | -1.21 - -0.04 |
| Joint Excursions (Degrees) | | |
| Knee flexion excursion (KFE) | - 44.39 ± 6.86 | - 26.32 - -60.83 |
| Knee frontal plane excursion (KVE) | - 3.99 ± 5.12 | - 16.87 - 6.49 |
| Knee transverse plane excursion (KRE) | -7.35 ± 4.77 | -17.82 - 3.05 |
| Peak Net Knee Joint Moment (Nm) | | |
| Peak internal knee extension moment (PKEM) | 112.22 ± 31.09 | 32.20 - 213.47 |
| Peak internal frontal plane moment (PKVM) | -10.03 ± 15.21 | -42.72 - 35.99 |
| Peak internal transverse plane moment (PKRM) | -8.35 ± 4.99 | -21.54 - 0.47 |
| Force (N) | | |
| Peak anterior shear force (PASF) | 398 ± 78 | 205.56 - 607.91 |

**Table 7. Correlation Matrix of Dependent and Independent Variables (Hypotheses 2a-g)**

| | | **1** | **2** | **3** | **4** | **5** | **6** | **7** | **8** | **9** | **10** | **11** | **12** | **13** |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| **1** | Vastus Lateralis torque impulse (a) | **1** | | | | | | | | | | | | |
| **2** | Vastus Medialis torque impulse (a) | 0.48* | **1** | | | | | | | | | | | |
| **3** | Bicep Femoris torque impulse (a) | -0.33* | -0.07 | **1** | | | | | | | | | | |
| **4** | Semitendinous torque impulse (a) | -0.19 | -0.32* | 0.72* | **1** | | | | | | | | | |
| **5** | Knee flexion excursion (b) | 0.01 | -0.04 | 0.17 | 0.37* | **1** | | | | | | | | |
| **6** | Frontal plane excursion (b) | 0.11 | 0.23 | -0.09 | -0.16 | -0.01 | **1** | | | | | | | |
| **7** | Knee transverse plane excursion b) | -0.10 | -0.02 | 0.09 | -0.09 | 0.27 | 0.37* | **1** | | | | | | |
| **8** | Peak knee extension moment (c) | 0.36* | 0.47* | -0.09 | -0.25 | -0.21 | 0.39* | -0.14 | **1** | | | | | |
| **9** | Peak frontal plane moment (c) | 0.26 | 0.32* | -0.30 | -0.21 | -0.24 | 0.39* | -0.05 | 0.14 | **1** | | | | |
| **10** | Peak transverse plane moment (c) | -0.14 | 0.18 | 0.01 | -0.01 | 0.21 | 0.08 | 0.23 | -0.21 | 0.54* | **1** | | | |
| **11** | Peak anterior shear force (c) | 0.36* | 0.33* | -0.22 | -0.27 | -0.23 | 0.18 | -0.19 | 0.76* | -0.09 | -0.38* | **1** | | |
| **12** | Weight | 0.31 | 0.49* | -0.16 | -0.12 | 0.13 | 0.21 | -.061 | 0.60* | 0.02 | -0.00 | 0.63 | **1** | |
| **13** | Height | 0.17 | 0.46* | 0.09 | -0.03 | -0.04 | 0.38* | 0.10 | 0.42* | 0.15 | 0.18 | 0.25 | 0.65* | **1** |

N=40, * Significant correlations (P<0.05)

a) Integration of the muscle torques estimated using the angle and action specific sEMG/Torque ratios over the impact phase of the initial landing of a drop jump maneuver.
b) Calculated as the difference between knee flexion angle at landing and at the end of the impact phase of the initial landing of a drop jump maneuver.
c) Identified as the maximal value of the anterior shear force as calculated through an inverse dynamics analysis during the impact phase of the initial landing of a drop jump maneuver.

Summary of the Regression Models for Hypothesis 2, Noting R Squared Values and Final Regression Equations

| **Dependent variable** | **Step** | **Predictor variables** | **$R^2$ Change (P value)** | **Final Regression Equation** |
|---|---|---|---|---|
| Knee Flexion Excursion | 1 | BW, HEIGHT | 4.3% (0.440) | $KFE=-0.726+0.283_{BW}-0.240_{HT}-0.027_{VLDYN}+0.102_{VMDYN}-0.144_{BFDYN}+0.524_{STDYN}$ |
| | 2 | $VL_{DYN}$, $VM_{DYN}$, $BF_{DYN}$, $ST_{DYN}$ | 17% (0.166) | |
| Peak Extensor Moment | 1 | BW, HEIGHT | 35.7% (<0.001) | $PKEM=-0.226+0.540_{BW}*-0.009_{HT}+0.248_{VLDYN}-0.016_{VMDYN}+0.358_{BFDYN}-0.397_{STDYN}$ |
| | 2 | $VL_{DYN}$, $VM_{DYN}$, $BF_{DYN}$, $ST_{DYN}$ | 11% (0.172) | |
| Peak Anterior Shear Force | 1 | BW | 39.2% (<0.001) | $PASF=-0.009+0.636_{BW}*+0.260_{VLDYN}-0.209_{VMDYN}+0.187_{BFDYN}-0.342_{STDYN}$ |
| | 2 | $VL_{DYN}$, $VM_{DYN}$, $BF_{DYN}$, $ST_{DYN}$ | 8.1% (0.285) | |
| Frontal Plane Excursion | 1 | BW, HEIGHT | 14.9% (0.050) | $KVE=-2.428-0.127_{BW}+0.444_{HT}+0.010_{VLDYN}+0.048_{VMDYN}-0.063_{BFDYN}-0.104_{STDYN}$ |
| | 2 | $VL_{DYN}$, $VM_{DYN}$, $BF_{DYN}$, $ST_{DYN}$ | 2.9% (0.883) | |
| Peak Frontal Plane Moment | 1 | BW, HEIGHT | 4.6% (0.416) | $PKVM=-1.361-0.540_{BW}*+0.329_{HT}-0.021_{VLDYN}+0.514_{VMDYN}*-0.636_{BFDYN}*+0.349_{STDYN}$ |
| | 2 | $VL_{DYN}$, $VM_{DYN}$, $BF_{DYN}$, $ST_{DYN}$ | 29.4% (0.025) | |
| Transverse Plane Excursion | 1 | BW, HEIGHT | 3.7% (0.494) | $KRE=-1.156-0.213_{BW}+0.238_{HT}-0.123_{VLDYN}-0.079_{VMDYN}-0.110_{BFDYN}-0.158_{STDYN}$ |
| | 2 | $VL_{DYN}$, $VM_{DYN}$, $BF_{DYN}$, $ST_{DYN}$ | 1.4% (0.966) | |
| Peak Transverse. Plane Moment | 1 | BW, HEIGHT | 5.9% (0.323) | $PKRM=-1.474-0.367_{BW}+0.297_{HT}-0.442_{VLDYN}*+0.578_{VMDYN}*-0.588_{BFDYN}*+0.570_{STDYN}$ |
| | 2 | $VL_{DYN}$, $VM_{DYN}$, $BF_{DYN}$, $ST_{DYN}$ | 23.2% (0.077) | |

* indicates significance of the predictor at the 0.05 level.

**Sagittal Plane**

***Knee Flexion Excursion***

When predicting knee flexion excursion, it was found that once accounting for individual differences in height and weight ($R^2$= 4.3%, P =0.440) entering of all the torque impulses into the regression did not significantly increase the predicted variance in knee flexion excursion ($R^2$ change = 17%, P = 0.166) (Table 8). The overall model predicted 21 % of the variance in the knee flexion excursion (P=0.223). Table 9 presents the parameter estimates for the full regression model when examining the contribution of thigh muscles torque impulses, estimated using the angle and action specific sEMG/Torque ratios, over the impact phase of the initial landing of a drop jump maneuver to the knee flexion excursion during the impact phase of the initial landing of a drop jump maneuver.

**Table 8. Regression Coefficients for the Final Regression Model Looking at the Individual Neuromuscular Characteristics Predicting Knee Flexion Excursion**

| Variable | Unstandardized Coefficients | Standard Error | t | Sig. |
|---|---|---|---|---|
| Constant | -20.236 | 27.874 | -0.726 | 0.473 |
| Weight | 0.244 | 0.194 | 1.257 | 0.217 |
| Height | -0.226 | 0.203 | -1.113 | 0.274 |
| Estimated Vastus Lateralis Torque Impulse | -0.160 | 1.155 | -0.138 | 0.891 |
| Estimated Vastus Medialis Torque Impulse | 0.496 | 1.127 | 0.440 | 0.662 |
| Estimated Bicep Femoris Torque Impulse | -2.507 | 4.645 | -0.540 | 0.593 |
| Estimated Semitendinous Torque Impulse | 14.830 | 7.335 | 2.022 | 0.051 |

* Significant Regression Coefficient, P < 0.05

### *Peak Internal Extensor Moment*

When predicting peak internal knee extension moment, it was found that once accounting for individual differences in height and weight ($R^2$= 35.7%, P <0.01) entering of all the torque impulses into the regression did not significantly increase the amount of variance in the peak in internal extensor moment ($R^2$ change = 11%, P = 0.172) (Table 8).

The final regression model predicted 46.7 % of the variance in the peak in internal extensor moment (P=0.001). Table 10 presents the parameter estimates for the full regression model when examining the contribution of thigh muscle torques impulses, estimated using the angle and action specific sEMG/Torque ratios over the impact phase of the initial landing of a drop jump maneuver, to the peak internal knee extensor moment as determined by inverse dynamics.

**Table 9. Regression Coefficients for the Regression Model Looking at the Individual Anthropometrics and Muscle Torque Impulses Predicting Peak Internal Extensor Moment**

| Variable | Unstandardized Coefficients | Standard Error | t | Sig. |
|---|---|---|---|---|
| Constant | -23.446 | 104.763 | -0.226 | 0.823 |
| Weight | 2.111* | 0.722 | 2.923 | 0.006 |
| Height | -0.039 | 0.756 | -0.052 | 0.959 |
| Estimated Vastus Lateralis Torque Impulse | 6.538 | 4.301 | 1.520 | 0.138 |
| Estimated Vastus Medialis Torque Impulse | -0.344 | 4.194 | -0.082 | 0.935 |
| Estimated Bicep Femoris Torque Impulse | 28.189 | 17.291 | 1.630 | 0.113 |
| Estimated Semitendinous Torque Impulse | -50.883 | 27.305 | -1.864 | 0.071 |

* Significant Regression Coefficient, P < 0.05

***Peak Anterior Shear Force***

When predicting peak anterior shear force, it was found that once accounting for individual differences in weight ($R^2$= 39.2%, P <0.001), entering of the muscle torque impulses into the regression did not significantly increase the amount of variance explained ($R^2$ change = 8.1%, P =0.285).

The final regression model predicted 47.7 % of the variance in peak anterior shear force (P<0.001). Table 11 presents the parameter estimates for the full regression model when examining the contribution of thigh muscles torques impulses, estimated using the angle and action specific sEMG/Torque ratios over the impact phase of the initial landing of a drop jump maneuver, to the resultant peak anterior shear force as determined by inverse dynamics.

**Table 10. Regression Coefficients for the Final Regression Model Looking at the Individual Anthropometrics and Muscle Torque Impulses Predicting Peak Anterior Shear Force**

| Variable | Unstandardized Coefficients | Standard Error | t | Sig. |
|---|---|---|---|---|
| Constant | -0.728 | 80.091 | -0.009 | 0.141 |
| Weight | 6.306* | 1.468 | 4.296 | 0.000 |
| Estimated Vastus Lateralis Torque Impulse | 17.403 | 10.669 | 1.631 | 0.112 |
| Estimated Vastus Medialis Torque Impulse | -11.621 | 10.268 | -1.132 | 0.266 |
| Estimated Bicep Femoris Torque Impulse | 37.284 | 42.151 | 0.885 | 0.383 |
| Estimated Semitendinous Torque Impulse | -111.162 | 67.649 | -1.643 | 0.110 |

* Significant Regression Coefficient, P < 0.05

### Frontal Plane

#### *Frontal Plane Excursion*

When predicting frontal plane excursion, it was found that once accounting for individual differences in height and weight ($R^2$= 14.9%, P =0.050) entering of all the torque impulses into the regression did not significantly increase the amount of variance in the frontal plane excursion ($R^2$ change = 2.9%, P = 0.883).

The final regression model was not significant in explaining the variance in the peak knee adduction/abduction excursion based on anthropometrics and muscle torque impulses as investigated in the current study ($R^2$= 17.8%, P= 0.336). Table 12 presents the parameter estimates for the full regression model when examining the contribution of thigh muscles torques impulses, estimated using the angle and action specific sEMG/Torque ratios over the impact phase of the initial landing of a drop jump maneuver to the frontal plane excursion during the impact phase of the initial landing of a drop jump maneuver.

**Table 11. Regression Coefficients for the Final Regression Model Looking at the Individual Anthropometrics and Muscle Torque Impulses Predicting Frontal Plane Excursion**

| Variable | Unstandardized Coefficients | Standard Error | t | Sig. |
|---|---|---|---|---|
| Constant | -51.546 | 21.227 | -2.428 | 0.021 |
| Weight | -0.082 | 0.148 | -0.554 | 0.583 |
| Height | 0.312 | 0.155 | 2.014 | 0.068 |
| Estimated Vastus Lateralis Torque Impulse | 0.044 | 0.880 | 0.050 | 0.961 |
| Estimated Vastus Medialis Torque Impulse | 0.172 | 0.858 | 0.200 | 0.842 |
| Estimated Bicep Femoris Torque Impulse | -0.814 | 3.537 | -0.230 | 0.819 |
| Estimated Semitendinous Torque Impulse | -2.207 | 5.586 | -0.395 | 0.695 |

* Significant Regression Coefficient, P < 0.05

***Peak Internal Frontal Plane Moment***

When predicting peak internal frontal plane moment it was found that once accounting for individual differences in height and weight ($R^2$= 4.6%, P=0.416), adding all the muscle torque impulses significantly increased the variance in peak internal frontal plane moment explained by the model ($R^2$ change = 29.4%, P=0.014).

The final regression model predicted 34 % of the variance in peak internal frontal plane moment (P=0.025). Table 13 presents the parameter estimates for the full regression model when examining the contribution of thigh muscles torques impulses, estimated using the angle and action specific sEMG/Torque ratios over the impact phase of the initial landing of a drop jump maneuver, to the resultant peak internal frontal plane moment as determined by inverse dynamics during the impact phase of the initial landing of a drop jump maneuver. Specifically, greater magnitudes of vastus medialis torque impulse and greater magnitudes of bicep femoris torque impulse were predictive of greater internal valgus moment magnitudes about the knee. Interpretation of the regression equation suggests that for every increase of 1Nm.s in the Vastus Medialis and Bicep Femoris torque impulses, increases of 5.5 and a 24.5 Nm in the peak internal abduction moment would be observed.

**Table 12. Regression Coefficients for the Final Regression Model Looking at the Individual Anthropometrics and Muscle Torque Impulses Peak Internal Frontal Plane Moment**

| Variable | Unstandardized Coefficients | Standard Error | t | Sig. |
|---|---|---|---|---|
| Constant | -76.914 | 56.498 | -1.361 | 0.183 |
| Weight | -1.033* | 0.393 | -2.626 | 0.013 |
| Height | 0.687 | 0.412 | 1.667 | 0.105 |
| Estimated Vastus Lateralis Torque Impulse | -0.274 | 2.342 | -0.117 | 0.908 |
| Estimated Vastus Medialis Torque Impulse | 5.527* | 2.284 | 2.420 | 0.021 |
| Estimated Bicep Femoris Torque Impulse | -24.484* | 9.415 | -2.601 | 0.014 |
| Estimated Semitendinous Torque Impulse | 21.880 | 14.867 | 1.472 | 0.151 |

* Significant Regression Coefficient, P < 0.05

### Transverse Plane

#### *Transverse Plane Excursion*

When predicting transverse plane excursion it was found that once accounting for individual differences in height and weight ($R^2$= 3.7%, P=0.494), adding all the muscle torque impulse did not increase the variance in transverse plane excursion explained by the model ($R^2$ change = 1.4%, P=0.966).

The final regression model was not significant in explaining the variance in the transverse plane excursion moment based on anthropometrics and muscle torque impulses as investigated in the current study ($R^2$= 5.3%, P= 0.927). Table 14 presents the parameter estimates for the full regression model when examining the contribution of thigh muscles torques impulses, estimated using the angle and action specific sEMG/Torque ratios over the impact phase of the initial landing of a drop jump maneuver to the transverse plane excursion during the impact phase of the initial landing of a drop jump maneuver.

**Table 13. Regression Coefficients for the Final Regression Model Looking at the Individual Anthropometrics and Muscle Torque Impulses Predicting Transverse Plane Excursion**

| Variable | Unstandardized Coefficients | Standard Error | t | Sig. |
|---|---|---|---|---|
| Constant | -24.542 | 21.221 | -1.156 | 0.309 |
| Weight | -0.128 | 0.148 | -0.865 | 0.388 |
| Height | 0.156 | 0.155 | 1.005 | 0.389 |
| Estimated Vastus Lateralis Torque Impulse | -0.498 | 0.880 | -0.566 | 0.755 |
| Estimated Vastus Medialis Torque Impulse | 0.265 | 0.858 | 0.309 | 0.969 |
| Estimated Bicep Femoris Torque Impulse | -1.326 | 3.536 | -0.375 | 0.785 |
| Estimated Semitendinous Torque Impulse | 3.120 | 5.584 | 0.559 | 0.823 |

* Significant Regression Coefficient, P < 0.05

***Peak Internal Transverse Plane Moment***

When studying peak internal transverse plane moment it was found that once accounting for individual differences in height and weight ($R^2$= 5.9%, P= 0.323), including all the muscle torque impulses in the model did not significantly increase the amount of variance explained ($R^2$ change = 23.2%, P = 0.077).

The final regression model was not significant in explaining the variance in the peak transverse plane moment based on anthropometrics and muscle torque impulses as investigated in the current study ($R^2$= 26.6 %, P= 0.095). Although the $R^2$ was not significant, the parameter estimates for the final model reveal that increased vastus lateralis, decreased vastus medialis, decreased bicep femoris, and increased semitendinous torque impulses were all significant predictors of increased internal rotation moments (P = 0.014-0.029).

Table 15 presents the parameter estimates for the full regression model when examining the contribution of thigh muscles torques impulses, estimated using the angle and action specific sEMG/Torque ratios over the impact phase of the initial landing of a drop jump maneuver to the peak transverse plane moment during the impact phase of the initial landing of a drop jump maneuver Interpretation of the regression equation suggests that for every increase of 1Nm.s in the Vastus Lateralis and Bicep Femoris torque impulses and for every decrease of 1 Nm.s in the Vastus Medialis and Semitendinous, respective increases of 1.8, 7.42, 2.04 and 11.74 Nm in the peak internal rotation moment would be observed.

**Table 14. Regression Coefficients for the Final Regression Model Looking at the Individual Anthropometrics and Muscle Torque Impulses Predicting Peak Transverse Plane Moment**

| Variable | Standardized Coefficients | Standard Error | t | Sig. |
|---|---|---|---|---|
| Constant | -28.813 | 19.551 | -1.474 | 0.150 |
| Weight | -0.230 | 0.136 | -1.692 | 0.100 |
| Height | 0.203 | 0.143 | 1.427 | 0.163 |
| Estimated Vastus Lateralis Torque Impulse | -1.871* | 0.810 | -2.308 | 0.027 |
| Estimated Vastus Medialis Torque Impulse | 2.041* | 0.790 | 2.582 | 0.014 |
| Estimated Bicep Femoris Torque Impulse | -7.425* | 3.258 | -2.279 | 0.029 |
| Estimated Semitendinous Torque Impulse | 11.747* | 5.145 | 2.283 | 0.029 |

* Significant Regression Coefficient, $P < 0.05$

# CHAPTER V

# DISCUSSION

The first purpose of this study was to examine the differences between estimations of knee joint extensor moment impulse during the impact phase of the initial landing of a drop jump maneuver using three different methods: isometric sEMG/Torque ratio, angle and action specific sEMG/Torque ratio and inverse dynamics analysis. The two first methods calculated the net knee joint moment as the difference between quadriceps and hamstrings torque impulses estimated during the impact phase of the initial landing of a drop jump maneuver using the sEMG/Torque ratio calculated during calibration actions performed on an isokinetic dynamometer. The extensor moment impulse calculated through an inverse dynamics analysis was used as a reference against which the results of the other methods were compared. The second purpose was to assess the extent to which the torque impulses of quadriceps and hamstrings medial and lateral components, estimated using the angle and action specific sEMG/Torque ratio, predicted high risk knee joint mechanics in the three planes of motion during the impact phase of the initial deceleration of a drop jump maneuver.

The primary finding was that, during the impact phase of the initial landing of a drop jump maneuver; the angle and action specific sEMG/Torque ratio provided a better estimation of the net knee joint moment calculated using an inverse dynamics analysis than the isometric sEMG/Torque ratio. We also found that, during the impact phase of the initial landing of a drop jump maneuver, torque impulses of quadriceps and hamstrings medial and lateral components estimated using the angle and action specific sEMG/Torque ratio were moderate predictors of high-risk knee joint mechanics in the frontal and transverse planes of motion.

The following discussion will first focus on the specific differences between net knee joint moments estimated using an inverse dynamics analysis, the isometric sEMG/Torque ratio and the angle and action specific sEMG/Torque ratio. Second, we will discuss the factors that may have affected our ability to predict high risk knee joint mechanics in the three planes of motion based on the torque impulses of quadriceps and hamstrings medial and lateral components estimated using the angle and action specific sEMG/Torque ratio. This will be followed by a discussion of clinical/research implications of the findings and directions for future research.

## Differences in Net Joint Estimation

The net joint moment calculated in an inverse dynamics analysis (PKEM) represents the sum of all moments, occurring at the joint, and created by passive and active structures, but does not allow us to differentiate their respective contributions (Winter 1990). In the current study, the net knee joint moment was also calculated as the difference between the active extensor (Quadriceps represented by the sum of vastus lateralis and vastus medialis) and flexor muscle torques (Hamstrings represented by the sum of bicep femoris and semitendinous). Muscle torques were estimated using two methods, first the isometric sEMG/Torque ratio and second the angle the action specific sEMG/torque ratio. Impulses were used for the statistical analyses.

Our findings are in agreement with our hypotheses and show that the net knee joint moment impulse, based on muscle torque impulses estimated using the angle and action specific sEMG/Torque ratio (Mean: 4.29 ± 2.18 Nm*s), was not significantly different (Mean difference: +3 %) from the net knee joint moment calculated using an inverse dynamics analysis (Mean: 4.46 ± 1.91 Nm*s) while the net knee joint moment calculated based on muscle torques estimated using the isometric sEMG/Torque ratio (Mean: 3.30 ± 1.87 Nm*s) was significantly lower (~25%) than the previous two methods.

To our knowledge, limited studies have compared the net knee joint moment calculated using muscle torques estimation, to the net knee joint moment calculated via an inverse dynamics analysis during a landing maneuver. Doorenbosch et al. (2003), compared the net knee joint moment estimated using an angle and action specific sEMG/Torque ratio and inverse dynamics during the propulsion of a jump. Based on the findings, but without further justification, the authors stated that using the angle and action specific sEMG/Torque ratio to estimate muscle torques during a dynamic maneuver was a clinically relevant method. Arguably, the reason why the authors pronounced this model to be relevant for clinical use is that in their study, it allowed for differentiation of the co-contraction patterns of ACL deficient and healthy participants. In comparison to the current investigation; Doorenbosch et al. (2003) used the propulsion of a squat jump, concentric in nature, whereas we used the impact phase of a landing which is mostly eccentric. Additionally, only five healthy and five ACL deficient individuals were studied in Doorenbosch et al. (2003) whereas the current investigation studied 40 healthy females. Their model resulted in a greater error between the two methods of net knee joint moment estimation (13.3% for healthy individuals) (Average root mean square error (RMS) between inverse dynamics analysis moment and sEMG/Torque ratio estimated net knee joint moment; expressed as a percent of the peak extensor moment as calculated through inverse dynamics) than in our study (average difference between net knee joint moment impulses based on the angle and action specific sEMG/Torque ratio ($NET_{DYN}$) and inverse dynamics (KEM) ≈ 3%). Thus, the current approach appears to be more appropriate than that used by Doorenbosch et al. (2003) to estimate net knee joint moment based on an inverse dynamic analysis. One of the reasons for the difference in the findings may be that Doorenbosch et al. (2003) used a 2$^{nd}$ order polynomial to model the sEMG/Torque ratio as a function of knee angle during the calibration actions. Using a polynomial likely induced some error due to the fitting of the curves. We used a direct interpolation technique that does not carry such error since the algorithm used to interpolate the sEMG/Torque ratio during the impact phase of the initial landing of a drop jump maneuver is set to find the exact, angle specific, sEMG/Torque ratio value gathered from the calibration actions.

Using the more traditional approach of the isometric sEMG/Torque ratio to estimate quadriceps and hamstrings torque impulses and further calculate the net knee joint moment during the impact phase of landing resulted in an ~25% underestimation of the net knee joint moment. It is important to note that such results are difficult to compare with the literature. The isometric method, as is used, ignores the torque production capability of muscles, but uses normalized sEMG as an indicator of the mechanical influence of muscles upon the joint. As such it makes the assumption of the sEMG/Torque ratio without however making the calculations of the resultant muscle torques estimates. To summarize, estimating net knee joint moment using the angle and action specific sEMG/Torque ratio appears to be more valid than other methods used in the literature to estimate net sagittal moment impulse about the knee.

Regarding the second part of hypothesis 1, we hypothesized that quadriceps estimated torque impulses would be greater when using the angle and action specific sEMG/Torque ratio than with the isometric sEMG/Torque ratio and that the hamstrings estimated torque impulses would be higher using the isometric sEMG/Torque ratio. This was based on prior evidence of differences in sEMG/Torque relation across action velocities as previously presented in Figure 4 (Yeadon, King et al. 2006). For the quadriceps, our results confirm our hypothesis; estimated quadriceps torque impulses were greater with the angle and action specific sEMG/Torque ratio ($VL_{DYN} > VL_{ISO}$ by 21%, P=0.004 and $VM_{DYN} > VM_{ISO}$ by 30%, P< 0.001). For the hamstrings the results were opposite to the hypothesis, with no significant differences between estimated torque impulses using the angle and action specific sEMG/Torque ratio and using the isometric sEMG/Torque ratio ($BF_{DYN} > BF_{ISO}$ by 50%, P=0.08 and $ST_{DYN} > ST_{ISO}$ by 31 %, P=0.20).

Quadriceps torque impulses were 21-30% greater when estimated with the angle and action specific sEMG/Torque ratio compared to the isometric sEMG/Torque ratio. This may be explained by lower sEMG/Torque ratio observed during eccentric compared to isometric actions (Yeadon, King et al. 2006). This means that in eccentric, compared to isometric actions, greater torques can be produced for a specific level of activation. Similar to our findings, Yeadon et al. (2006) reported that the maximal torque generating capability of the knee extensors of two international standard athletes was 30 to 40 % higher during eccentric than isometric actions. Thus, current findings for the quadriceps agree well with the limited literature we were able to locate on sEMG/Torque ratios.

Contrary to our hypothesis, hamstrings torque impulses were not different when using the angle and action specific sEMG/Torque ratio or when using the isometric sEMG/Torque ratio. Our hypothesis was based on in-vitro evidence that, independent of muscle length, the force generating capability of a muscle is less during concentric than isometric muscle actions (Brown, Scott et al. 1996; Yeadon, King et al. 2006). In the current study, it appears that this was not the case, and that the isometric sEMG/Torque ratio was similar to the angle and action specific sEMG/Torque ratio. Further insight into this issue is provided by plotting the isometric sEMG/Torque ratio and the angle and action specific sEMG/Torque ratio as calculated during isometric and concentric calibration actions (Figure 19).

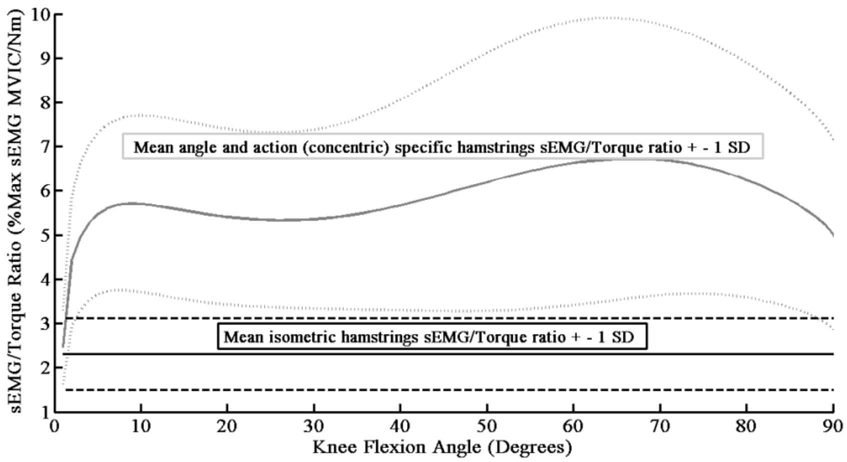



Figure 19: *Isometric And Concentric sEMG/Torque Ratio During Calibration Actions.*
*Greater sEMG/Torque ratios observed during concentric (gray line) than isometric (black line) actions.*

Figure 19 shows that overall the sEMG/Torque ratio is actually greater during concentric than isometric hamstrings actions. Consequently, the similarity of the estimated torque impulses for the angle and action specific sEMG/Torque ratio cannot be explained by similar isometric and angle and action specific sEMG/Torque ratios. In the calculations leading to the estimation of muscle torques, the other variable that may influence torque is the sEMG amplitude. The RMS, used for to filter the sEMG for isometric sEMG/Torque ratio, and low pass filter, used to filter the sEMG for the angle and action specific sEMG/Torque ratio, both result in a linear envelope of the filtered sEMG. However, the low pass filtering of the sEMG during isokinetic actions appears to result in lower absolute and relative values (%max sEMG during MVIC) than RMS smoothing of the isometric actions, which in turn would result in a greater sEMG/Torque ratio simply because of a lower value of the numerator in sEMG/Torque = sEMG divided by torque. Our choice to filter the sEMG differently for the isometric and angle and action specific sEMG/torque ratios, was based on the wide recognition/acceptance of these methods in the literature (Burden, Trew et al. 2003). This difference in processing complicates the comparisons of the estimations of hamstrings torque impulses between methods. However further studies should investigate the extent to which this difference in filtering affects the findings of the association between sEMG and joint mechanics. With regards to our purpose in the current study, using the isometric sEMG/Torque ratio to estimate mechanical outcomes of muscles resulted in large underestimations of the quadriceps and net knee joint moments.

Our ability to closely replicate the impulse of the net knee joint moment during the impact phase of the initial landing of a drop jump maneuver is determinant in the validation of a method designed to assess the contribution of quadriceps and hamstrings to knee joint mechanics in the three planes of motion. Looking at the limits of agreement between the angle and action specific sEMG/Torque and inverse dynamics methods (Figure 20) may bring further insight to the validity of using the angular and action specific sEMG/Torque ratio to estimate the contribution of thigh muscle to net knee joint moment. As the original intent of the limits of agreement was to compare an alternative method to a reference method (Bland and Altman 1986) it is well suited to this analysis.

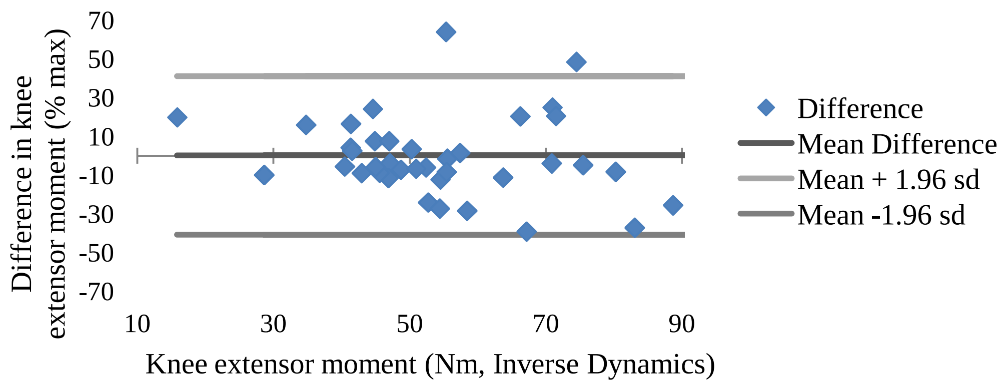



Figure 20: *Limits of Agreement: Inverse Dynamics and Angle and Action Specific sEMG/Torque Estimations Of Net Knee Joint Moment.*
*The majority of net joint moments estimated using the angle and action specific sEMG/Torque ratio were ± 20 % of the net joint moment estimated using an inverse dynamics analysis.*

In order to compare the current errors to the limited literature, we expressed the difference between the net knee joint moments estimated using the angle and action specific sEMG/Torque and inverse dynamics as a function of the peak in inverse dynamics net knee joint moment (Doorenbosch and Harlaar 2003). The limits of agreement (Figure 20) reveal that using the angle and action specific sEMG/Torque ratio over estimates, on average, the net joint moment by less than 1% and that all aside from three data points fall within the range of the mean plus or minus 1.96 standard deviations. This lends support to the validity of the model (Bland and Altman 1986). Additionally, there was almost an equal number of participants for which the angle and action specific sEMG/Torque ratio based estimation of net knee joint moment over or under-estimated the net knee joint moment impulse as determined by inverse dynamics (KEM), which drove the small average difference between the two methods. A description of the amplitudes of the actual errors is warranted here to better understand the difference between the two methods of net knee joint moment estimation. Averages of absolute differences between the two methods of net knee joint moment estimation, during the impact phase of the initial landing of a drop jump maneuver, are visually reported in Figure 21.

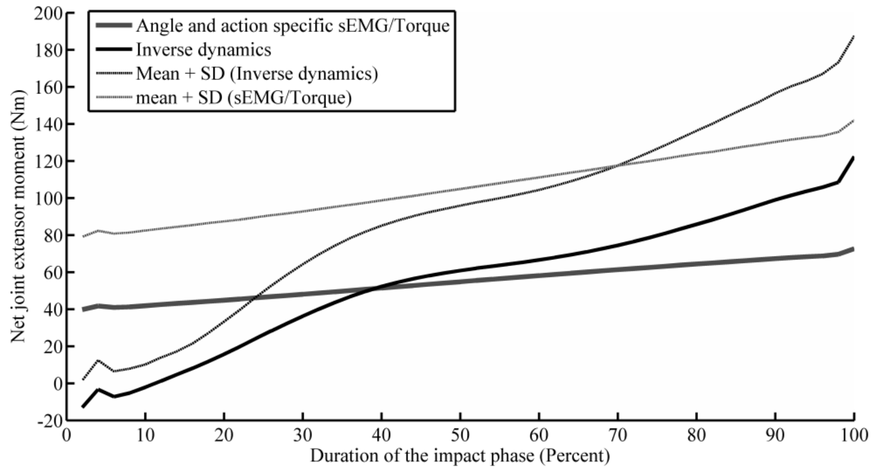



Figure 21: *Net Knee Joint Moments Estimated Using an inverse Dynamics or the Angle and Action Specific sEMG/Torque Ratio During the Impact Phase.*
*The estimation of the net joint moment using angle and action specific sEMG/Torque ratio is most accurate between ~30 - 60 % of the impact phase.*

Figure 21 demonstrates that despite overall errors of estimation of the angle and action specific sEMG/Torque as described in the limits of agreement (Figure 20), the inverse dynamics and sEMG/Torque based net knee joint moment curves during the impact phase of the initial landing of a drop jump maneuver are similar in nature, with the biggest difference between the two occurring between ground contact and 30 % of the impact phase. Overall, the model appears to be most valid in the middle of the impact phase and this may be due to the fact that the muscle actions undertaken in the extremes of the range of motion are most different from those used to calibrate the sEMG/Torque ratio.

Assessing the validity of the angl**e** and action specific sEMG/Torque ratio against inverse dynamics is somewhat complicated by the fact that information represented by the two methods greatly differs. Inverse dynamics analysis estimates the net knee joint moment including the mechanical contributions provided by all active and passive structures (Winter 1990), whereas we only accounted for the torques produced by the quadriceps and hamstrings. Also, inverse dynamics does not allow for specifying the contributions of flexor and extensor torques and therefore does not allow us to directly assess the validity of our muscle specific findings. These issues make it difficult for us to ascertain the origin of the difference between the current method and inverse dynamics. It is likely that various factors, such as the misestimation of extensor and flexor torques or not accounting for the moments created by other active and passive structures, contributed to the differences observed.

Muscle torque estimation during the impact phase of the initial landing of a drop jump maneuver is based on the angle and action specific sEMG/Torque ratio. This ratio is designed to best represent physiological characteristics or the “actual” sEMG/Torque ratio. However, this “actual” ratio remains unknown and if calibration actions are not fully representative of those performed during landing, the sEMG/Torque ratio calculated will not be ecologically valid which in turn may result in misestimation of muscle torques during landing. Differences in body position and/or knee joint flexion velocity between calibration actions and landing maneuvers are two aspects that may hinder the ecological validity of the angle and action specific sEMG/Torque ratio. Despite our best efforts to use hip and knee positions representative of the positions experienced during landing maneuvers, we did not modify hip position to account for increasing hip flexion during landing. This may have affected our calculation of the sEMG/Torque ratio and our estimation of quadriceps and hamstrings torques. Previous evidence reports 33° of hip flexion excursion (14 to 47°) between the instants of ground contact and peak loading (Blackburn and Padua 2008). Our participants performed calibration actions in 25 ° of hip flexion as it represents a value that approximates the middle of the hip range of motion during the impact phase of the initial landing of a drop jump maneuver. However, as hip flexion increases, the length of the Rectus femoris decreases and based on the length/force relation, this is likely to decrease its contribution to the net knee joint moment (Salzman, Torburn et al. 1993; Kong and van Haselen 2010). During calibration actions with the hip at 25 ° of hip flexion, the rectus femoris remained in a relatively lengthened position, contributing to knee extensor torque to a large extent (Salzman, Torburn et al. 1993; Zhang, Wang et al. 2003). This contribution was accounted for in the sEMG/Torque ratio because the torque produced by the rectus femoris was included in the sEMG/Torque ratio; however, the change in rectus femoris contribution to the net knee joint moment was not. If the actual contribution of the rectus femoris decreases during the landing and our method still assumes equal contribution, it may be prone to overestimating the net knee joint moment compared to what is physiologically occurring.

The change in hip flexion angle may also have affected hamstrings torque estimations. As mentioned previously, when the hip flexes hamstrings length increases (Visser, Hoogkamer et al. 1990). The hamstrings were in a relatively shortened position during calibration actions used to calculate the sEMG/Torque ratio. If hip flexion was greater during landing, the hamstrings would have been placed in a better position to produce torque, which may in turn have resulted in an underestimation of the torque produced by the hamstrings (Lunnen, Yack et al. 1981). Isometric hamstrings torque has been shown to increase by 27% between 0 and 45° of hip flexion with a concurrent decrease of 8% in sEMG amplitude. Failure to account for this increased torque producing capability together with decreased sEMG amplitude may result in both an underestimation of the hamstrings torques and an overestimation of the net knee joint moment during the impact phase of the initial landing of a drop jump maneuver.

Knee flexion velocity during the landing may also have affected the ecological validity of the sEMG/Torque method. Specifically, for the quadriceps, we used a 270 $° s^{-1}$ eccentric action, which is less than the average knee flexion velocity during the impact phase of the initial landing of a drop jump maneuver (Current study=- 442± 68 $°s^{-1}$). Further analysis of current data revealed that individuals for which the angle and action specific sEMG/Torque ratio over estimated the net knee joint moment also had significantly greater knee flexion velocities during landing than those individuals for which the angle and action specific sEMG/Torque ratio under estimated the net knee joint moment (-463 ± 75 $°s^{-1}$ Vs -413 ± 44 $°s^{-1}$; P = 0.019). This may indicate that the sEMG/Torque ratio estimated during calibration actions may not be an accurate depiction of what actually happens at the muscle during the impact phase of the landing. The overestimation observed would be linked to a decrease in torque generating capability at greater eccentric velocity. This change would not be accounted for by the current method as we only calibrated the sEMG/Torque relation using one velocity. This hypothesis, that eccentric torque generating capability decrease per unit of sEMG differs from previous reports that torque producing capability of the quadriceps increases slightly with increasing eccentric velocities (Yeadon, King et al. 2006). The fact that we used active females and not international level athletes may explain this difference, as higher level athletes are more likely to be well trained for high velocity eccentric actions. To summarize, the overestimation of net knee joint moment may have been driven by their lessened efficiency during higher eccentric velocities (potentially higher sEMG/Torque ratio) compared to the velocities used during the calibration actions. This in turn would result in an overestimation of the quadriceps torques and of the net joint moment. However, the extent to which using faster calibration actions may have affected the model is difficult to predict as no studies could be found that have published information related to quadriceps angle specific sEMG/Torque ratio at high eccentric velocities in females. Part of the difficulty would necessitate healthy females performing maximal eccentric quadriceps actions at very high velocities (>400 deg/sec), which, from experience, are quite difficult for participants to perform. Possibly, those individuals could be trained to perform such high velocity actions. But, such an extensive eccentric training protocol may result in neuromuscular changes that affect the ability or the way in which individuals perform the landing maneuvers and as such could be detrimental to an observational study.

For the hamstrings, 90°$s^{-1}$ concentric calibration actions were selected because of evidence that hamstrings are acting concentrically during the initial descent phase of lunges (Jonhagen, Halvorsen et al. 2009) and squats (Robertson, Wilson et al. 2008). The exact nature of the hamstrings action during the impact phase of the initial landing of a drop jump maneuver, and the extent to which it varies between individuals, still remains unclear. Despite the previous evidence showing that the hamstrings are likely acting concentrically (Robertson, Wilson et al. 2008; Jonhagen, Halvorsen et al. 2009) there is potential for isometric or eccentric action depending upon the coupled hip and knee sagittal plane kinematics. It is expected that using eccentric calibrations actions would have resulted in a lower sEMG/Torque ratio because of the well accepted greater torque generating capability and lower sEMG amplitude of eccentric Vs. concentric actions (Bigland and Lippold 1954; Kellis and Baltzopoulos 1998). A lower sEMG/Torque ratio would result in higher estimated hamstrings torques and, subsequently, lower net knee joint moment estimation. This may have improved the fit of the sEMG/Torque based estimation of the net knee joint moment to that calculated using an inverse dynamics analysis. To further investigate the type of action undertaken by the hamstrings during the impact phase of the initial landing of a drop jump maneuver, we relied on a simple model proposed by Vandekojt et al. (2008) to estimate hamstrings (Bicep femoris and semitendinous) length based on simultaneous inputs of hip and knee sagittal plane angles. Using the hip and knee sagittal plane data collected in the current study, this model resulted in estimated increases in bicep femoris (11±1.8%) and semitendinous (12±2.0%) lengths during the impact phase of the initial landing of a drop jump maneuver. This is counter to the previously hypothesized shortening actions used for the current model. Given the uncertainty of hamstring action during the impact phase, further work is needed to clarify the calibration action velocity that would best represent their actual length changes during the impact phase of the initial landing of a drop jump maneuver.

Finally, considering other structures potentially creating a knee flexor moment and not accounted for by the current method may be important to explain the discrepancy between the findings of the angle and action specific sEMG/Torque ratio and inverse dynamics. Amongst those, the gastrocnemius may be specifically important to explain the difference in knee flexor moment between the current method and inverse dynamics analysis early in the impact phase of the initial landing of a drop jump maneuver. Previous in vivo work demonstrated that the gastrocnemius contributes to the flexion moment to the largest extent (18 Nm) when the knee is flexed around 30° and the ankle dorsiflexed (Gallucci and Challis 2002). During the period right after ground contact, this may lead to an overestimation of the knee net joint extensor moment (Figures 20 and 21) since the angle and action specific sEMG/Torque ratio does not account for the contribution of the gastrocnemius to the knee flexor moment.

Passive structures and their resultant contributions were also not accounted for in the sEMG/torque ratio method. Amongst those, the ACL may have contributed to the net knee extensor moment calculated through inverse dynamics, in part because it contributes to the control of anterior shear force. The ACL is known to restrain around 87 % of anterior shear forces (Butler, Noyes et al. 1980), when doing so it is going to contribute to a knee flexor moment because of its flexion moment arm (1cm average between 0 and 70 ° of knee flexion)(Herzog and Read 1993). Using this information with the current data suggests that the ACL may be contributing to a 3.5 Nm instantaneous flexor moment (Average ± SD Peak Anterior Shear Force, Current data = 398 N ± 78 * 87 % * 0.01 m = 3.46 ± 0.68 Nm). This may be a significant contribution as it represents 10 – 15 % of the average error of estimation observed between inverse dynamics and angle and action specific sEMG/Torque ratio.

*Summary*

To summarize, the net knee joint moment calculated based on quadriceps and hamstrings torques estimated using the angle and action sEMG/Torque ratio was similar to the net knee joint extensor torque extracted from inverse dynamics. We described how a lack of ecological validity of the calibration actions used to calculate the sEMG/Torque ratio as well the differences in estimated mechanical outcomes between the two methods and the contribution of other structures to the net knee joint moment may explain the observed differences.

## Contribution of Muscle Torque Impulses to High Risk Mechanics

The following section will discuss the findings related to hypotheses 2 a-g in a plane by plane manner.

### Sagittal Plane

***Knee Flexion Excursion***

Reports based on 3D video reconstruction of actual injury events suggest that the small knee flexion excursions (24°) observed between the instant of ground contact and 40 ms post ground contact are part of the ACL injury mechanism (Koga, Nakamae et al. 2010). Decreased knee flexion excursion is thought to be due to the increased quadriceps eccentric torque producted to counteract the external flexor moment created by ground reaction forces (Lloyd and Buchanan 1996). Since females who have lesser knee flexion excursions also have greater quadriceps activation (Malinzak, Colby et al. 2001; Griffin, Albohm et al. 2006), we hypothesized that greater magnitudes of quadriceps torque impulses and lesser magnitudes of hamstrings torques would predict smaller knee excursions. Contrary to our hypothesis, anthropometrics and muscle torque impulses did not predict knee flexion excursion. Similar findings have been previously reported in studies investigating the contribution of neuromechanical factors to knee flexion excursion (Shultz, Nguyen et al. 2009). After accounting for peak isometric strength, neither quadriceps nor hamstrings pre-or post-ground contact normalized activations were found to be significant predictors of knee flexion excursion during the deceleration phase of the initial landing of a drop jump maneuver (Shultz, Nguyen et al. 2009). The current method did not improve the ability to understand neuromuscular contributions to knee flexion excursion despite accounting for the angle and action specificities of the sEMG/Torque ratio.

Our observed knee flexion excursions (23±3° at 40 ms) were similar to those reported in a study of ACL injury mechanism (24° at 40ms) (Koga, Nakamae et al. 2010). This lends evidence to the validity of studying the impact phase of the initial landing maneuver to replicate sagittal plane kinematics representative of the knee joint mechanics leading up to actual ACL injury.

The discrepancy between our hypothesis and the current findings may be due to mechanical factors that have not been accounted for in the analysis we undertook. Based on the assumption that stiffer landings are characterized by smaller flexion excursions, lower deceleration velocities and result from large eccentric quadriceps torques (Devita and Skelly 1992), we split the group by the median in knee flexion excursion and assessed differences between the two groups created (small excursion: 20° Vs. large excursion: 24°, Difference: 22 %). We found that the small excursion group experienced slightly shorter impact phase lengths (76ms Vs 80 ms, Difference: 5 %), greater vastus lateralis (+7%), greater vastus medialis (+17%), and lesser bicep femoris (-76%) and semitendinous (-51%). along with lesser average deceleration velocities (-530 $°\cdot s^{-1}$ Vs. -605$°\cdot s^{-1}$, Difference: -15%). Despite the differences in kinematics, no significant differences in estimated muscle torques could be highlighted between those two groups. This may be due the greater instantaneous torques of the small excursion group not being apparent, in comparison with those undergoing greater excursion. This is due to the fact that those individuals with greater excursion, who actually use lesser instantaneous quadriceps torques, produce them over a 5 % longer period which results in a larger impulse. Together with other findings that females that go on to getting injured have a 16% shorter deceleration phase than those who do not (Hewett, Myer et al. 2005) this demonstrates the potential importance of accounting for the duration of the impact phase.

For an exploratory analysis we reran the regression, including anthropometrics and impact phase duration in a first step, followed by muscle torques impulses. The results showed that after accounting for anthropometrics and length of the impact phase ($R^2$ = 71%, $P<0.001$) including the average torques increased the variance explained by 9.6 % ($P=0.01$). In the final model, lesser impact phase duration (beta: -1.011) and greater vastus lateralis impulse (beta: 0.295) were significant predictors of lesser knee flexion excursion. This suggests that increasing vastus lateralis torque by 1.18 Nm*s is linked to around a three degree decrease in knee flexion excursion. To summarize, accounting for the duration of the impact phase is determinant in understanding the contribution of the thigh muscles to knee joint excursions in the sagittal plane.

***Peak Extension Moment***

Laboratory based investigations report that the peak knee extensor moment is the most direct contributor to anterior shear force (Sell, Ferris et al. 2007; Shultz, Nguyen et al. 2009), which in turn is known to stress the ACL (Butler, Noyes et al. 1980). As a component of the inverse dynamics analysis, it expresses the net internal knee joint moment provided by all structures and is calculated based on kinematics, ground reaction forces and anthropometric data (Winter 1990). Since the quadriceps and hamstrings have large extensor and flexor moment producing capability, respectively, we hypothesized that greater quadriceps torque impulses and lesser hamstrings torque impulses would predict greater peak knee extension moment during the impact phase of landing.

As expected, greater weight was predictive of greater peak knee extension moment ($\beta$ = 0.540, P = 0.006). But contrary to our hypothesis, none of the muscle torque impulses were predictive of the peak net internal knee extensor moment. Similarly, Shultz et al. (2009), the only other study that investigates the contribution of neuromechanical factors to net knee joint moment findings, reported that after accounting for peak isometric strength, neither quadriceps nor hamstrings normalized sEMG (averaged in the 150ms pre-or post-ground contact) were significant predictors of the peak in the knee extension moment during the initial landing of a drop jump maneuver (Shultz, Nguyen et al. 2009). Therefore the current method does not improve our understanding of the neuromuscular contributions to the peak net internal knee extension moment despite accounting for the angle and action specificities of the sEMG/Torque ratio. Potential reasons will be discussed below.

Our findings for the magnitude of peak net knee joint extensor moment are smaller than those observed in the literature; Shultz et al. (2009) reported peak KEM (deceleration phase) ($0.087 \pm 0.029$ Nm * $BW^{-1}$ * $Ht^{-1}$); Sell et al. (2007) reported KEM at Peak Posterior Ground Reaction Force ($0.056 \pm 0.044$ Nm * $BW^{-1}$ * $Ht^{-1}$); whereas KEM for the impact phase in the current study was $0.011 \pm 0.003$ Nm * $BW^{-1}$ * $Ht^{-1}$. Our results are more similar to those reported by Zhang et al. (2000) in a study of male kinetics during a step off landing from a 32 cm box. The authors report average peak knee moment right after the peak in ground reaction force of $2.04 \pm 0.38$ Nm* $BW^{-1}$ (Zhang, Bates et al. 2000) which is very similar to the $1.86 \pm 0.42$ Nm* $BW^{-1}$ observed in the current study. The difference between Shultz et al. (2009) and the current data may be explained by the fact that they report kinetics observed through the whole deceleration phase between ground contact and deepest knee flexion whereas we observed the impact phase only.

Inter-individual differences in the duration of the impact phase may have again affected the findings. Specifically, as individuals perform a landing maneuver, the amplitude of the knee extensor moment is largely determined by the stiffness of the muscular system with greater instantaneous quadriceps torques resulting in higher peaks in the knee extensor moment and shorter impact phases (Devita and Skelly 1992; Zhang, Bates et al. 2000). Conversely, those landing in a softer fashion, and with lower peaks in the knee extensor moment, would have longer durations of the impact phase (Decker, Torry et al. 2003). Using the impulse may therefore fail to differentiate between the stiff and soft landings because the higher moment produced during a stiff landing may be hidden by the longer duration of the impact phase in those with a soft landing. This is exemplified in the following Figure 22.

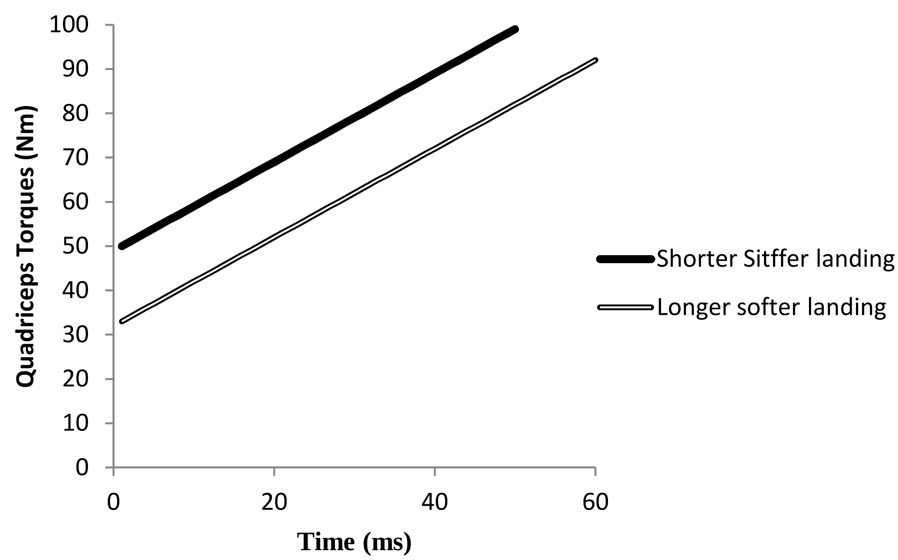



Figure 22: *Exemplification of the Confounding Effect of Using The Impulse of Quadriceps Torques.*
*When the length of the impact phase is not accounted for, higher instantaneous torques (Black Line) may result in similar impulses than lower instantaneous torques.*

In the above Figure 22, the artificially created data shows two individuals with similar impulses (3.72 Vs. 3.75 Nm*s), one landing in stiff fashion and experiencing a 7 % higher quadriceps torque (99 Vs. 92 Nm) and a 20 % shorter impact phase (50 Vs. 60 ms). This suggests that using the impulse of the quadriceps torque may fail to differentiate between two different landing styles. The average quadriceps torque (74 Vs. 62 Nm) highlights a difference in quadriceps torques between the stiff and soft landings and as such may better represent the neuromuscular characteristics of the quadriceps for the purpose of this study.

Second, joint stiffness is a function of the relative contribution of the quadriceps and hamstrings to the net knee joint moment (Decker, Torry et al. 2003). In the current study, we observed the magnitude of the quadriceps and hamstrings torques separately without regards to their relative magnitudes. Co-contraction ratios of different forms are commonly found in the literature assessing neuromuscular strategies (Doorenbosch and Harlaar 2003; Kellis, Arabatzi et al. 2003). It is plausible that the mechanical outcome of a specific hamstrings torque will be different depending on the magnitude of the simultaneous quadriceps torque.

To summarize, muscle torques impulses estimated using the current method did not predict the peak in internal knee extension moment, this finding may be related to the nature of impulses and the fact that they are largely affected by the duration of the impact phase or to the necessity to look at the ratio between hamstrings and quadriceps torques.

***Peak Anterior Shear Force***

Anterior shear force is the most direct ACL loading mechanism (Butler, Noyes et al. 1980). The forces created by the thigh muscles are thought to be major contributors to anterior shear forces with quadriceps torques increasing anterior shear forces (Grood, Suntay et al. 1984; Li, Rudy et al. 1998) and hamstrings forces decreasing them (Pandy and Shelburne 1997; Mesfar and Shirazi-Adl 2006). Therefore, we hypothesized that greater quadriceps torque impulse together with lesser hamstrings torque impulse would predict increased peak anterior shear forces. Contrary to our hypothesis, current findings showed that once accounting for individual differences in weight ($R^2$= 39.2%, P <0.001) adding the estimated muscle torque impulses did not significantly increase the proportion of variance in anterior shear force explained by the model ($R^2$ change = 8.1%, P =0.285). In the final model, none of the muscle torque impulses were significant predictors of the peak in anterior shear force.

Other biomechanical studies of landings have reported the ability to predict peak anterior shear forces. Sell et al. (2007), found that the integrated EMG of the vastus lateralis was a significant positive predictor of anterior shear force in a model, including peak posterior ground reaction force, knee flexion moment, knee flexion angle and gender, that in total accounted for 86% of the variance of anterior shear force. Shultz et al. (2009) report similar findings where knee flexion excursion, hip flexion excursion, knee extensor moment and quadriceps activation post ground contact were collectively significant predictors of anterior shear force ($R^2$= 53.8 %) and the activation of the quadriceps post ground contact after accounting for all of the afore-mentioned variables, explained an additional 7.3% of the variance in the peak anterior shear force. Both of these previous studies included sagittal plane kinematic and kinetic variables, in the current model however we did not. It is very likely that these previous significant findings are due to the inclusion of the net knee extensor moment, as it is highly correlated to anterior shear force (Current data: $R^2$= 59%, P<0.001).

Other factors that may explain why our model did not predict the peak anterior shear force was that ground reaction forces likely contribute to anterior shear force and ACL load through two main mechanisms (Li, Rudy et al. 1998). The vertical component of ground reaction forces induces an anterior shift of the tibia relative to the femur because of the posterior tibial slope (Li, Rudy et al. 1998). The posterior component on the other hand, contributes to the external flexion moment that needs to be counteracted by an extensor moment to avoid collapsing of the knee (Devita and Skelly 1992). The quadriceps create this extensor moment but also contribute to shear and compressive forces that may also result in shear (Grood, Suntay et al. 1984; Li, Rudy et al. 1998). However, the effect of the ground reaction forces on knee joint mechanics and the extent to which it varies across individuals is largely unknown and may have affected our ability to predict peak anterior shear forces from neuromuscular variables.

Based on prior in-vivo and in-vitro evidence it appears that failing to include specific information about the individual’s landing style such as peak knee flexion, duration of the impact phase or peak ground reaction force into the regression model may have contributed to our inability to distinguish the contribution of quadriceps and hamstrings to sagittal plane knee joint mechanics as observed through inverse dynamics.

**Frontal Plane**

Frontal plane knee joint mechanics are commonly described as part of a multi-planar ACL injury mechanism (Markolf, Burchfield et al. 1995; Ireland, Gaudette et al. 1997; Ireland 1999; Koga, Nakamae et al. 2010) and external valgus moments during landing have been shown to predict ACL injury (Hewett, Myer et al. 2005). Hewett et al. (2005) suggest that increased external valgus moments are due to a lack of neuromuscular control characterized by a heightened reliance on the quadriceps. This is based on reports that quadriceps forces induce valgus moments (Zhang, Wang et al. 2003) and excursions (DeMorat, Weinhold et al. 2004) whereas hamstrings forces are known to create varus motions (Kwak, Ahmad et al. 2000).

***Frontal Plane Excursions***

Knee valgus collapse is thought to be part of the ACL injury mechanism (Ireland, Gaudette et al. 1997; Koga, Nakamae et al. 2010) and increased knee valgus excursions are often associated with a greater risk of ACL injury (Neptune, Wright et al. 1999; Boden, Dean et al. 2000; Olsen, Myklebust et al. 2004). The quadriceps and hamstrings are thought to influence frontal plane knee joint mechanics not only because of their large size but also because of their respective valgus and varus moment arms (Lloyd and Buchanan 2001). Therefore we hypothesized that greater vastus lateralis and medialis torque impulses together with lesser bicep femoris and semitendinous torque impulses would predict greater valgus excursions. Contrary to our hypothesis, none of the anthropometric and neuromuscular variables studied in the final model were significant in explaining the variance in the knee frontal plane excursion.

As one would expect, excursions observed in the current study (Frontal plane excursion: -3.99 (Varus) ± 5.12°) were of lesser magnitude than those observed during the 40ms post ground contact of actual injury events (~12° Valgus) (Koga, Nakamae et al. 2010). A prospective study of females reported that females who went on to sustain a non-contact ACL injury had similar valgus excursions than those who did not get injured (Injured: 4 °; Non injured: 4.8°) but that they reached greater peak valgus angle because they contacted the ground already in a valgus position (ground contact front plane angle; Injured: 5 ° (Valgus); Non injured: -10.4 ° (Varus)) (Hewett, Myer et al. 2005). In the current study the participants landed in initial valgus (2.8 ± 3°) which is similar to other values reported previously for females landing from a volleyball block jump (1.6 ± 2.8°)(Hughes, Watkins et al. 2008). However, they moved towards varus which is more comparable to frontal plane knee joint kinematics reported for males during landing from a volleyball block jump (Frontal plane excursion: -3.5 (Varus) ± 9.6 °) (Hughes, Watkins et al. 2008). The reason for the difference between the reported literature and the current study in terms of frontal plane excursions during landing is unclear as the tasks and populations were very similar, however the current study investigated a shorter period after ground contact which may likely explain the difference in the magnitude of the results.

Our model attempted to explain the variance in frontal plane excursion based on sagittal plane estimations of muscle torques. However, the link between those two variables may be mitigated by the angle specific moment arms of the thigh muscles in the frontal plane. For example, the extent to which quadriceps forces induce valgus rotation largely increases when the knee is flexed to more than 30° whereas the capability of the hamstrings to counteract this rotation does not increase to the same extent at those angles (Kwak, Ahmad et al. 2000).

The non significance of the current results may also be explained by the fact that other factors, such as external (ground reaction) and internal (muscle) forces are likely to contribute to the moment found using an inverse dynamics analysis. For example ground reaction forces are known to induce valgus motions (Meyer and Haut 2005) and the gastrocnemius may contribute to posterior shear force (O'connor 1993) and neutral frontal plane alignment (Lloyd and Buchanan 2001). Not accounting for those may have affected our ability to predict frontal plane excursion based on estimated muscle torque impulses.

To summarize, our participants displayed frontal plane kinematics similar to those reported in the literature for ground contact position, but on average moved into varus rather than valgus as is commonly observed which may be explained by the fact that we studied a shorter phase than previous studies. Our inability to predict frontal plane excursions during the impact phase of the initial landing of a drop jump maneuver may be explained by the limitations of our approach in using the sEMG/Torque method and not accounting for other factors that may influence frontal plane knee excursion, such as ground contact frontal plane knee joint position, peak ground reaction forces or the contributions of other structures. Because of the importance of frontal plane kinematics, further investigation is required to determine the factors contributing to increase valgus excursions.

***Peak Internal Frontal Plane Moment***

Hewett et al. (2005) found that increased external valgus moments were predictive of injury and suggested that increased valgus moments may be due to a heightened reliance on quadriceps during deceleration. Therefore, we hypothesized that greater quadriceps estimated torque impulses together with lesser hamstrings estimated torque impulses will predict greater internal knee valgus moments. The findings show that after accounting for anthropometric ($R^2$= 4.6%, P=0.416) inclusion of the muscle torques significantly increased the amount of variance explained ($R^2$ change = 29.4%, P=0.014). The final regression model predicted 34% of the variance in the peak frontal plane moment with lesser weight (Beta= -0.540), greater magnitudes of bicep femoris (Beta= -0.636), and vastus medialis (Beta= 0.514) predicted greater peak internal valgus moment. Mechanically, those findings can be explained by quadriceps torques creating frontal plane moments (Zhang, Wang et al. 2003), and by the valgus moment arm of bicep femoris (Lloyd and Buchanan 2001).

In comparison to the literature, our participants displayed lesser internal valgus moments (8.7 ± 17.3 Nm) than reported in other studies. In Hewett et al. (2005), prospectively screened females who go on to injure their ACL had much greater magnitudes of valgus moments during a drop jump maneuver off of a 30 cm box than those who did not get injured (Relative: + 31%, Injured: -45.3 ± 28.5 Nm Vs. Non Injured – 18.4 ± 15.6 Nm) (Hewett, Myer et al. 2005). This suggests that on average our participants displayed lesser high risk mechanics in the frontal plane. The graphical representation of the abduction moment as a function of the percent landing phase proposed in McLean et al. (2007) provides further insight into this. Specifically, it was shown that the peak in external valgus moment occurs late (80 to 100%) in the landing phase and that early in the landing (15-30%), a phase corresponding to the impact as defined in the current study, frontal plane moments are characterized by neutral or external varus moments (Between 0.01-

0.05 Nm* $BW^{-1}$ * $Ht^{-1}$), more similar to the current results (0.09 ± 1.6 Nm* $BW^{-1}$ * $Ht^{-1}$). The differences in phases investigated may therefore explain the magnitude differences of the current findings.

Our limited ability to predict peak frontal plane moments during the impact phase of the initial landing of a drop jump maneuver (> 70% of the variance remained unexplained in our model) may be explained by the limitations of the current method. The angle and action specific sEMG/Torque ratio does not account for internal and external forces, as well as differences in structural characteristics, that may influence the peak in frontal plane knee moment. For example ground reaction forces and frontal plane position at ground contact is highly correlated with valgus moment (r=0.69) (Sigward and Powers 2007). Higher loading rates, due to a 15% shorter impact phase duration, are also observed in females with higher knee valgus moments (Hewett et al. (2005) possibly imparting a greater load upon the knee joint structures, including the ligaments and joint capsule (Lloyd and Buchanan 1996). This may have in total affected the sensitivity of the model to predict the variance in the peak in frontal plane moment since those variables were not accounted for in the current approach.

To summarize, greater magnitudes of vastus medialis torque impulses and greater magnitudes of bicep femoris torque impulses were predictive of greater internal valgus moment magnitudes about the knee. This is likely due to the muscles anatomical arrangement and frontal plane mechanical influences upon the knee joint (Lloyd and Buchanan 1996). In the current method, not accounting for kinematic and kinetic variables may explain the limited predictivity of the regression model to distinguish the contribution of the muscle torque impulses to frontal plane knee joint mechanics.

### Transverse Plane

In the transverse plane, internal rotation excursions (Koga, Nakamae et al. 2010) and peak external internal rotation moment (Meyer and Haut 2008), are commonly related to the ACL injury mechanism (Koga, Nakamae et al. 2010). Both internal (muscle) (Zhang, Wang et al. 2003) and external (ground reaction) forces (Meyer and Haut 2005) can contribute to transverse plane knee joint mechanics. This is determined in part by the interaction of forces and knee joint geometry, specifically the more concave shape of the medial menisci, that facilitates internal tibial rotation during axial loading (Blankevoort and Huiskes 1996; Vedi, Williams et al. 1999; Meyer and Haut 2008; Stijak, Herzog et al. 2008).

#### *Transverse Plane Excursion*

Since quadriceps and hamstrings forces induce internal (Hirokawa, Solomonow et al. 1992; MacWilliams, Wilson et al. 1999; Li, DeFrate et al. 2004) and external (Kwak, Ahmad et al. 2000) rotations, respectively, we hypothesized that greater quadriceps torque impulses and lesser hamstrings torque impulses would predict greater internal tibial rotation excursions. Contrary to the hypothesis, our results suggest that neither anthropometric nor neuromuscular variables as investigated in the current study predicted internal rotation excursion ($R^2$= 5.3%, P= 0.927).

The magnitude of internal rotation excursions during the impact phase in the current study (7.3°) are slightly smaller than those observed during actual injury events (8.0°) (Koga, Nakamae et al. 2010). As such the impact phase of the initial landing of a drop jump maneuver appears to be a relevant time epoch and task to reproduce high risk transverse plane kinematics.

While the quadriceps and hamstrings, in isolation, are thought to respectively act as internal (Zhang, Wang et al. 2003) and external rotators (Li, Rudy et al. 1998; MacWilliams, Wilson et al. 1999; Kwak, Ahmad et al. 2000) of the tibia, current findings suggest that their mechanical contributions to internal tibial rotation excursion were not significant. A reason may be that the extent to which thigh muscles torques, estimated in the sagittal plane, affect transverse plane knee joint mechanics is largely dependent on knee flexion angle (Kwak, Ahmad et al. 2000). In a cadaveric model, quadriceps forces were found to induce the greatest magnitudes of internal rotation between 30 and 60° of knee flexion (Li, Rudy et al. 1998; Kwak, Ahmad et al. 2000) whereas the ability of the hamstrings to counter those motions was most pronounced with the knee flexed to more than 30°. The interpretation of the estimated torque impulses in the current method in comparison with in-vitro studies necessitates for those torques to be expressed as forces. This conversion, of torque to force, is angle specific as it includes the sagittal plane moment arms of the quadriceps and hamstrings (Herzog and Read 1993). Reports show that the quadriceps moment arm is relatively independent of the knee flexion (± 1cm between 0 and 90 ° of flexion, moment arm at full extension= 4 to 6 cm) whereas the hamstrings moment arms largely increase with knee flexion (+ 2 to 3 cm between 0 and 90 ° of knee flexion, moment arm at full extension= 0.5 to 1.5 cm) (Herzog and Read 1993). Thus, failure to account for initial knee joint angles may have contributed to our inability to predict internal rotation excursions.

As an exploratory follow up, we compared the quadriceps estimated torques impulses for those landing in internal rotation (or neutral) (n=14, Mean angle at ground contact= - 3°) Vs. those landing in external rotation (n=26, Mean angle at ground contact = 6°) and compared their kinematic and neuromuscular variables. Taken together the findings suggest that the initial internal rotation group experienced larger peak internal rotation angles (Internal rotation group -10° Vs. External rotation group -1°), apparently because they landed already in internal rotation and not because they underwent greater excursions (Transverse plane excursion (Internal rotation <0) Internal rotation group -7° Vs. External rotation group -7°)(P=0.98). The internal rotation group, however, appear to have relied on a neuromuscular strategy characterized by 46% greater vastus medialis ($P<0.05$), 57% greater vastus lateralis ($P<0.05$) extensor torques and 87% ($P<0.05$) greater bicep femoris and 37% greater semitendinous ($P>0.05$) flexor torques. This difference in neuromuscular strategy suggests that those landing in internal rotation adopted a generalized co-contraction strategy to maintain knee position in the transverse plane (Lloyd and Buchanan 2001).

Regardless of neuromuscular strategy used, it appears that the current participants were relatively homogenous in the amount of internal rotation excursion. Therefore, it may be more appropriate to investigate the relation between estimated muscle torque impulses and transverse plane peak position at the end of the impact phase ("peak") rather than excursion through the impact phase. The advantage of proceeding in this fashion is that differences in ground contact knee alignment in the transverse plane would be reflected in the predicted outcome. Therefore we reran the regression model, including, internal rotation angle at ground contact, duration of the impact phase and the muscle torque impulses. The results suggest that landing in internal rotation (Beta=0.687, $P<0.001$) and having a shorter impact phase (Beta=-0.299, $P=0.019$) predicted a greater peak transverse plane position, while the contribution of estimated muscle torque impulses were non-significant.

Based on this information it is difficult to draw a conclusion regarding the possible mechanical outcomes of the muscle torques, but to summarize, those individuals landing in internal rotation experienced greater peak internal rotation angles, had greater quadriceps and bicep femoris torque impulses, but went through the same amount of excursion as those landing in external rotation. This may be a conditioned response strategy to control internal rotation excursions as demonstrated by greater overall muscle torque impulses (Lloyd and Buchanan 2001).

***Peak Internal Transverse Plane Moment***

In isolation, the application of large internal rotation moments at the tibia can result in ACL rupture (Meyer and Haut 2008). Since quadriceps and hamstrings forces contribute to internal (Zhang, Wang et al. 2003) and external tibial rotation moments (Kwak, Ahmad et al. 2000) respectively, we hypothesized that greater quadriceps torque impulses and lesser hamstrings torque impulses would predict greater peak internal rotation moments during the impact phase of the initial landing of a drop jump maneuver. Our results show that the final regression model neared statistical significance in explaining the variance in peak transverse moment ($R^2$= 26.6 %, P= 0.095). Increased vastus lateralis torque impulse (Beta= -0.442), decreased vastus medialis torque impulse (Beta= 0.578), decreased bicep femoris torque impulse (Beta= -0.588), and increased semitendinous torque impulse (Beta= 0.570) were significant predictors of increased internal rotation moments.

Mechanically, those findings can be explained by the anatomical arrangement and the mechanical contribution of the afore mentioned muscles to knee joint transverse plane mechanics. Vastus lateralis, vastus medialis and semitendinous are demonstrated internal rotators (Buford, Ivey et al. 2001; Zhang, Wang et al. 2003) whereas bicep femoris is an external rotator (Buford, Ivey et al. 2001).

The magnitude of internal rotation moment observed in the current study (8.15 Nm) is smaller than that observed elsewhere during the drop jump (Current study: (0.08 Nm * $BW^{-1}$ * $Ht^{-1}$ Vs. 0.11 Nm* $BW^{-1}$ * $Ht^{-1}$) (McLean, Fellin et al. 2007) which may be explained, as mentioned previously, by the fact that McLean et al. (2007) investigated the entire stance phase whereas we looked at the impact phase. Based on the graphical representations of group data in the afore mentioned report, lower moments than those reported as peaks can be observed to occur within the first 25 % of the landing phase. Internal tibial rotation moments of around 0.03 Nm * $BW^{-1}$ *

$Ht^{-1}$ can be estimated from the graphs presented, which is more comparable to the values we presented (0.08 Nm * $BW^{-1}$ * $Ht^{-1}$). The importance of a 8 Nm internal rotation torque upon ACL loading mechanics is unclear as Meyer et al. (2005) in a cadaveric set up, demonstrated that ~33 Nm of internal tibial rotation torque resulted in ACL rupture.

Several factors may have affected our ability to explain a significant proportion of the variance in peak transverse moments during the impact phase of the initial landing of a drop jump maneuver because of other forces and moments that also contribute to the knee joint moments observed using an inverse dynamics analysis. We did not account for ground reaction force magnitude. This may be an important factor since ground reaction forces have been found to increase internal tibia rotation (Meyer and Haut 2005) possibly because of shape differences between the medial and lateral joint structures/anatomy (Blankevoort and Huiskes 1996; Vedi, Williams et al. 1999; Meyer and Haut 2008; Stijak, Herzog et al. 2008). When considering internal forces not accounted for in the current method, we did not include more proximal muscles such as muscles attaching to the illiotibial band (Kwak, Ahmad et al. 2000) or hip extensors and abductors (Powers 2010), that are suggested to influence knee rotation mechanics. In the current method and regression we did not account for the extent to which the contribution of those structures to transverse plane knee joint mechanics varies between individuals. In turn, this may have affected our ability to predict the extent to which muscles contributed to transverse plane moments.

To summarize findings in the transverse plane; muscle torque impulses did not predict excursions, potentially because of the small variability of the internal rotation excursion values. We found that individuals landing in internal rotation went through the same amount of knee excursion as those who landed in external rotation but used a neuromuscular strategy characterized by overall increased muscle torques. We also found that, in a non significant model predicting 26 % of the variance in internal rotation moment, increased vastus lateralis, decreased vastus medialis, decreased bicep femoris, and increased semitendinous torque impulses were all significant independent predictors of increased internal rotation moments. Further studies are needed to investigate the exact extent to which the variance in peak internal rotation moment can be explained by isolated muscle torque impulses or by their interactions.

#### *Implications & Future Directions*

Current findings revealed that greater vastus medialis torques do predict higher risk mechanics in the frontal and transverse plane and that those individuals landing with the knee close to full extension and in internal rotation are more likely to have to rely on greater muscle torques to control knee motion. Based on this, it appears important that prevention programs should include components that promote ground contact in a flexed position and neutral knee alignment in the frontal and transverse planes together with the ankle slightly dorsiflexed. The direction of this practice should be guided by the findings relating increased impact phase duration and relatively externally rotated knee at ground contact with lesser peak internal rotation moment. Those findings together with injury mechanism descriptions relate that the short time frame in which the injury occurs may not allow for the response to a specific stimuli from the passive structures. This highlights the importance of learned strategies to perform safer landing and as such, practicing landing as described above should be performed from different types of approaches that are specific to the game played which may provide for central adaptations (Pascual-Leone, Amedi et al. 2005) decreasing the potential for injury during actual match play (Hewett, Lindenfeld et al. 1999; Powers and Fisher 2010). To increase the ecological validity of this practice, having the individuals reacting to a specific command should be gradually included once the individual can perform the maneuver in an adequate fashion (assessed by the visual inspection of a double legged landing, checking for sufficient knee flexion and frontal plane neutral alignment throughout the descent phase of a drop jump). Gradually, motor skills related to the task of landing may also transfer to the game through increased physical performance (Hewett, Lindenfeld et al. 1999) and awareness of situational demands presented in the reactive practice.

The findings of hypothesis 1 suggest that in females, using the angle and action specific muscle torque impulses provides a better estimation of sagittal plane knee joint mechanics estimated through inverse dynamics than the classically used isometric method. Therefore in future studies looking at the relation between sEMG and biomechanical measures in the sagittal plane, angle and action specific sEMG/Torque ratio models should be implemented, to better understand the role of sagittal muscles during landing activities.

The findings of hypothesis 2 are the first to report a direct estimation of muscle torques during a landing maneuver based on angle and action specific sEMG/ratio. Based on this method we found that vastus medialis and bicep femoris were significant predictors of peak internal moments in the frontal and transverse planes. The roles of vastus medialis and bicep femoris were, however, dependent on the plane studied. Vastus medialis contributed to increased frontal plane high risk mechanics and decreased transverse plane high risk mechanics, whereas bicep femoris contributed to increased frontal plane high risk mechanics and decreased transverse plane high risk mechanics. In summary, this suggests the contribution of muscle torques to high risk mechanics is largely a function of the plane of motion studied. This differential effect of muscle upon knee joint mechanics may be due to other factors such as ground contact knee position in the three planes of motion as it largely determines the influence of the applied forces upon the joint. Further studies should investigate the dependency of muscles torques to ground contact knee position.

The current project has raised several issues with regard to advancement of the work in future investigations. In the discussion of hypothesis 1, it was pointed out that not knowing the exact action undertaken by the hamstrings may be driving some of the error observed when the current method was compared to inverse dynamics. Therefore it would be relevant to assess how hamstrings length actually changes during the deceleration phase of a landing maneuver in order to use the most valid calibration action to calculate the sEMG/Torque for the hamstrings.

Another issue highlighted in the discussion was the difference in filtering methods between the isometric and angle and action specific sEMG/Torque ratio methods, further studies should investigate how changing the filtering parameters actually affects the directionality of the findings. This would allow further insight into the usefulness of the angle and action specific sEMG/Torque ratio compared to the isometric method.

We have also suggested that the contribution of proximal and distal muscles may have affected our ability to more closely match the inverse dynamics analysis. One way in which we could investigate this issue using the current data, is to include the energy absorption at the hip in the regression model as a representation of the contribution of the hip to the overall muscular work. This may also bring further insight into the simultaneous contributions of the hip and knee muscles to high risk knee joint mechanics.

With regards to predicting high risk knee joint mechanics, the current study focused on the impact phase of the landing because of its relevance to commonly report ACL injury mechanism. Future studies should look at high risk knee joint mechanics over the whole deceleration phase and determine the extent to which the muscle torques created during the impact phase contribute the subsequent mechanics of the joint. This would allow for the data extracted from the sEMG/Torque ratio to be compared to the relatively large body of evidence addressing landing mechanics over the entire landing phase. Given the seemingly important nature of ground contact knee joint position in determination of landing mechanics, future research should also focus on a better understanding of the neuromuscular factors occurring during the flight phase that potentially contribute to knee joint position at ground contact and duration of the impact phase. Those appear to be important to understand the contribution of muscle torque impulses to high risk knee joint mechanics during the impact phase of the initial landing of a drop jump maneuver.

## Limitations

It is recognized that there are limitations associated with the current study. The angle and action specific sEMG/Torque ratio is limited in validity because the calibration actions were performed on an isokinetic dynamometer, in an open chain, whereas the impact phase of the drop jump is a closed chain activity. To a certain extent this may have affected the validity of the angle and action specific sEMG/Torque ratio because of the differences in the direction of the resistance vector of forces, and how they contribute to shear and compressive forces, between the calibration actions and the impact phase of the landing maneuver (Kaufman, An et al. 1991; Fleming, Ohlen et al. 2003).

This study was further limited as we did not account for knee position at ground contact in our regression models. This may be important to understand the contribution of the torque impulses to knee joint mechanics, specifically in the sagittal plane due in part to the effect of knee flexion on the moment arm of the quadriceps and hamstrings but also because medial condyle displacement is limited between 0 and 30 ° of knee flexion (Johal, Williams et al. 2005). This may have contributed to our limited ability to explain excursions based on muscle torque impulses during the impact phase of landing where the knee is in a relatively extended position.

Finally, including both the lateral and medial components of the quadriceps and hamstrings in the regression models may have affected the directionality of the regression coefficients findings. Since the medial and lateral components of the quadriceps are innervated by the femoral nerve (Thiranagama 1990), they were highly correlated (Correlation medial/lateral quadriceps: 0.48) in the current study. Given this correlation there is a possibility that this could explain some of the differential directionality of the regression coefficients for the medial and lateral components when assessing their contribution to the peak in the internal rotation moment.

## Summary and Conclusions

Based upon current findings, the angle and action specific sEMG/Torque ratio appears to provide a better estimation of the net knee joint moment than that estimated using the isometric sEMG/Torque ratio. However, there were inter-individual differences in the magnitude of error of the estimated net knee joint moment that may be improved upon through minor changes in the methods. Those misestimations were likely influenced by other external and internal forces influencing the net knee joint moment determined via inverse dynamics and not accounted for in the current method. Collectively, it was concluded that the use of angle and action specific sEMG/Torque ratio provides a better estimation of sagittal joint moments than does the traditional isometric approach to sEMG normalization.

In spite of the afore-mentioned potentially confounding factors, vastus medialis and bicep femoris muscle torque impulses, estimated using the angle and action specific sEMG/Torque ratio, were significant predictors of increased frontal plane and of increased and decreased, respectively, transverse peak moments about the knee. Current findings suggest that future studies should include impact phase duration, ground contact kinematics and ground reaction forces as those are shown in the current study to be determinant to understanding the contribution of muscle torque impulses to transverse and frontal plane moments during the impact phase of the initial landing of a drop jump maneuver.

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# APPENDIX A.

## INSTITUTIONAL REVIEW BOARD CONSENT FORM

**UNIVERSITY OF NORTH CAROLINA AT GREENSBORO**

***CONSENT TO ACT AS A HUMAN PARTICIPANT: LONG FORM***

Project Title: Strength, co-activation and kinematics during athletic tasks
Project Directors: Randy Schmitz & Jerome Sauret

Participant's Name: _____

**What is the study about?**
**The purpose of the project is to evaluate the day to day repeatability of strength, muscle activation and joint motions of a healthy population during athletic tasks**

**Why are you asking me?**
You are a healthy male or female between 18 and 30 without current lower extremity joint injuries or past knee surgeries. You are fairly active (2 to 10 hours a week) and you do not smoke.

**What will you ask me to do if I agree to be in the study?**
We will ask you to fill in physical activity and menstrual history questionnaires.
We will ask you to perform knee extension and flexion maximal efforts on a machine that assesses your muscle strength.
We will have you perform several athletic tasks (jumping and landing tasks) during which we will observe knee motion as well as muscle activity. Those activities are not very strenuous.
Finally, we will take body size measurements using a standard measuring tape.

**Is there any audio/video recording?**
No.

**What are the dangers to me?**
There is minimal danger associated with the participation in this study. Anytime physical activity is performed there is a small chance of joint or muscle injury.

If you have any concerns about your rights or how you are being treated please contact Eric Allen in the Office of Research and Compliance at UNCG at (336) 256-1482. Questions about this project or your benefits or risks associated with being in this study can be answered by Dr Randy Schmitz who may be contacted at 336.334.3031 or at rjschmit@uncg.edu.

**Are there any benefits to me for taking part in this research study?**
There is no direct benefit to you as a subject.

**Are there any benefits to society as a result of me taking part in this research?**
Society may benefit from a greater understanding of how people activate their muscles during physical activity.

**Will I get paid for being in the study? Will it cost me anything?**
There are no costs to you or payments made for participating in this study.

**How will you keep my information confidential?**
All information obtained in this study is strictly confidential unless disclosure is required by law. The information will be kept in a secured office and electronic information will be stored on a password protected computer.

**What if I want to leave the study?**
You have the right to refuse to participate or to withdraw at any time, without penalty. If you do

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Approved Consent Form
Valid 4/1/11 to 3/30/12

withdraw, it will not affect your in any way. If you choose to withdraw, you may request that any of your data which has been collected be destroyed unless it is in a de-identifiable state.

**What about new information/changes in the study?**
If significant new information relating to the study becomes available which may relate to your willingness to continue to participate, this information will be provided to you.

**Voluntary Consent by Participant:**
By signing this consent form you are agreeing that you read, or it has been read to you, and you fully understand the contents of this document and are openly willing consent to take part in this study. All of your questions concerning this study have been answered. By signing this form, you are agreeing that you are 18 years of age or older and are agreeing to participate, or have the individual specified above as a participant participate, in this study described to you by _____.

Signature: ________________________ Date: ________________

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Approved Consent Form
Valid 4/1/11 to 3/30/12

APPENDIX B.

MENSTRUAL HISTORY QUESTIONNAIRE

**FEMALE MENSTRUAL HISTORY**

This questionnaire asks questions about your menstrual cycle. As a reminder, this information is strictly confidential. None of this information will be shared with anyone besides the study investigators. Your survey uses a coded identification number in substitution for your name. If you have any questions, or do not understand any of the questions, please let us know.

**Subject Code:_________________** Date:___________ Date of Birth:________________

How old were you when you started your menstrual periods (Years of Age)? ________________

When was the first day of your last period (month/day)?________________________________

On average, how many days are there between your menstrual periods (i.e. 21 days, 28 days, 32 days, etc.)? _________________

How many menstrual periods have you had in the last 12 months? _______________________

Does your cycle length vary more than 1-2 days per month? YES NO

Have you missed any menstrual periods within the last 12 months (Please Circle)? YES NO

Since starting your menstrual periods, has there ever been an extended time where you did not have a menstrual period (Please Circle)? YES NO

If YES, when was the most recent, and for how long? _________________________________

When do you expect your next menstrual period will start (month/day)? ___________________

Do you experience premenstrual symptoms (Please Circle)? YES NO

If you experience premenstrual symptoms, please indicate their severity on a scale of 0-10:

**0 (None) 5 (Moderate) 10 (Severe)**

| | | | | |
|---|---|---|---|---|
| Check all that apply: | Bloating: | ________ | Spotting: | _______ |
| | Irritability: | ________ | Mood Swings: | _______ |
| | Food Cravings: | ________ | Other: | _______ |

Are you currently taking hormones (e.g. birth control pills, estrogen therapy, etc) for any reason (Please Circle)? YES NO

Have you taken hormones (e.g. birth control pills, estrogen therapy, etc) for any reason in the past (Please Circle)? YES NO

If Yes, When did you take these, and for how long?______________________________

Do you have plans to become pregnant in the near future? YES NO

# APPENDIX C.

# ACTIVITY RATING SCALE

**PHYSICAL ACTIVITY AND HEALTH HISTORY**

**Do you have any General Health Problems or Illnesses? (e.g. diabetes, respiratory disease) Yes____ No____**

**Do you have any vestibular (inner ear) or balance disorders? Yes____ No____**

**Do you smoke? Yes____ No____**

**Do you drink alcohol? Yes____ No____ If yes, how often?__________________________**

**Do you have any history of connective tissue disease or disorders? (e.g. Ehlers-Danlos, Marfan's Syndrome, Rheumatoid Arthritis) Yes____ No____**

**Has a family member of yours ever been diagnosed with breast cancer? Yes____ No____ (if no, please skip next question.)**

**If yes, please put a check next to the types of relatives that have been diagnosed. You may check more than one box:**

Mother________ Sister_______ Grandmother _______ Aunt ______.
Male relative (father, brother, grandfather, or uncle) _____________.
Other type of relative (please write in) ________________________.

**Please list any medications you take regularly: ____________________________________**

**Please list any previous injuries to your lower extremities. Please include a description of the injury (e.g. ligament sprain, muscle strain), severity of the injury, date of the injury, and whether it was on the left or right side.**

| Body Part | Description | Severity | Date of Injury | L or R |
|---|---|---|---|---|
| Hip | | | | |
| Thigh | | | | |
| Knee | | | | |
| Lower Leg | | | | |
| Ankle | | | | |
| Foot | | | | |

**Please list any previous surgery to your lower extremities (Include a description of the surgery, the date of the surgery, and whether it was on the left or right side)**

| Body Part | Description | Date of Surgery | L or R |
|---|---|---|---|
| | | | |
| | | | |
| | | | |
| | | | |

**Please list all physical activities that you are currently engaged in. For each activity, please indicate how much time you spend each week in this activity, the intensity of the activity (i.e. competitive or recreational) and for how long you have been regularly participating in the activity.**

| Activity | #Days/week | #Minutes/Day | Intensity | Activity Began When? |
|---|---|---|---|---|
| | | | | |
| | | | | |
| | | | | |
| | | | | |
| | | | | |
| | | | | |
| | | | | |

**What time of day do you generally engage in the above activities?**____________________

________________________________________________________________________

**Please list other conditions / concerns that you feel we should be aware of**:____________

________________________________________________________________________

________________________________________________________________________

________________________________________________________________________

________________________________________________________________________

# The Activity Rating Scale (partcode =--------)

Please indicate how often you performed each activity in your healthiest and most active state, **in the past year.**

| | **Less than one time in a month** | **One time in a month** | **One time in a week** | **2 or 3 times in a week** | **4 or more times in a week** |
|---|---|---|---|---|---|
| Running: running while playing a sport or jogging | | | | | |
| Cutting: Changing directions while running | | | | | |
| Decelerating: coming to a quick stop while running | | | | | |
| Pivoting: turning your body with your foot planted while playing a sport; For example: skiing, skating, kicking, throwing, hitting a ball (golf, tennis, squash), etc. | | | | | |

**Investigator Comments:**

APPENDIX D.

ANOVA SPSS OUTPUTS

**Within-Subjects Factors**

Measure:MEASURE_1

| Means_of_estimation | Dependent Variable |
|---|---|
| 1 | Iso_Net_Joint |
| 2 | Dyn_Net_ Joint |
| 3 | ID_Net_Joint |

**Multivariate Tests[b]**

| Effect | | Value | F | Hypothesis df |
|---|---|---|---|---|
| Means_of_estimation | Pillai's Trace | .434 | 14.589[a] | 2.000 |
| | Wilks' Lambda | .566 | 14.589[a] | 2.000 |
| | Hotelling's Trace | .768 | 14.589[a] | 2.000 |
| | Roy's Largest Root | .768 | 14.589[a] | 2.000 |

a. Exact statistic

b. Design: Intercept
Within Subjects Design: Means_of_estimation

**Multivariate Tests[b]**

| Effect | | Error df | Sig. |
|---|---|---|---|
| Means_of_estimation | Pillai's Trace | 38.000 | .000 |
| | Wilks' Lambda | 38.000 | .000 |
| | Hotelling's Trace | 38.000 | .000 |
| | Roy's Largest Root | 38.000 | .000 |

b. Design: Intercept
Within Subjects Design: Means_of_estimation

**Mauchly's Test of Sphericity**[b]

Measure:MEASURE_1

| Within Subjects Effect | Mauchly's W | Approx. Chi-Square | df | Sig. |
|---|---|---|---|---|
| Means_of_estimation | .867 | 5.432 | 2 | .066 |

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

b. Design: Intercept
Within Subjects Design: Means_of_estimation

**Mauchly's Test of Sphericity**[b]

Measure:MEASURE_1

| Within Subjects Effect | Epsilon[a] | | |
|---|---|---|---|
| | Greenhouse-Geisser | Huynh-Feldt | Lower-bound |
| Means_of_estimation | .882 | .921 | .500 |

Tests the null hypothesis that the error covariance matrix of the orthonormalized transformed dependent variables is proportional to an identity matrix.

a. May be used to adjust the degrees of freedom for the averaged tests of significance. Corrected tests are displayed in the Tests of Within-Subjects Effects table.

b. Design: Intercept
Within Subjects Design: Means_of_estimation

**Tests of Within-Subjects Effects**

Measure:MEASURE_1

| Source | | Type III Sum of Squares | df | Mean Square |
|---|---|---|---|---|
| Means_of_estimation | Sphericity Assumed | 31.543 | 2 | 15.772 |
| | Greenhouse-Geisser | 31.543 | 1.765 | 17.872 |
| | Huynh-Feldt | 31.543 | 1.842 | 17.122 |
| | Lower-bound | 31.543 | 1.000 | 31.543 |
| Error (Means_of_estimation) | Sphericity Assumed | 113.330 | 78 | 1.453 |
| | Greenhouse-Geisser | 113.330 | 68.831 | 1.646 |
| | Huynh-Feldt | 113.330 | 71.847 | 1.577 |
| | Lower-bound | 113.330 | 39.000 | 2.906 |

**Tests of Within-Subjects Effects**

Measure:MEASURE_1

| Source | | F | Sig. |
|---|---|---|---|
| Means_of_estimation | Sphericity Assumed | 10.855 | .000 |
| | Greenhouse-Geisser | 10.855 | .000 |
| | Huynh-Feldt | 10.855 | .000 |
| | Lower-bound | 10.855 | .002 |

**Tests of Within-Subjects Contrasts**

Measure:MEASURE_1

| Source | Means_of_estimation | Type III Sum of Squares | df | Mean Square |
|---|---|---|---|---|
| Means_of_estimation | Linear | 27.023 | 1 | 27.023 |
| | Quadratic | 4.520 | 1 | 4.520 |
| Error (Means_of_estimation) | Linear | 60.991 | 39 | 1.564 |
| | Quadratic | 52.339 | 39 | 1.342 |

**Tests of Within-Subjects Contrasts**

Measure:MEASURE_1

| Source | Means_of_estimation | F | Sig. |
|---|---|---|---|
| Means_of_estimation | Linear | 17.280 | .000 |
| | Quadratic | 3.368 | .074 |

**Tests of Between-Subjects Effects**

Measure:MEASURE_1
Transformed Variable:Average

| Source | Type III Sum of Squares | df | Mean Square | F | Sig. |
|---|---|---|---|---|---|
| Intercept | 1941.454 | 1 | 1941.454 | 215.918 | .000 |
| Error | 350.673 | 39 | 8.992 | | |

```
[DataSet1] C:\Users\Jerome\Desktop\data diss\stats\dataset\hyp10912.sav
```

**Paired Samples Statistics**

| | | Mean | N | Std. Deviation | Std. Error Mean |
|---|---|---|---|---|---|
| Pair 1 | Iso_Net_Joint | 3.303854176 | 40 | 1.87362149E0 | .2962455704 |
| | Dyn_Net_Joint | 4.296759061 | 40 | 2.18075473E0 | .3448075996 |
| Pair 2 | ID_Net_Joint | 4.466242962 | 40 | 1.90561560E0 | .3013042826 |
| | Dyn_Vast_Lat | 2.232384675 | 40 | 1.26094024E0 | .1993721577 |
| Pair 3 | ID_Net_Joint | 4.466242962 | 40 | 1.90561560E0 | .3013042826 |
| | Iso_Net_Joint | 3.303854176 | 40 | 1.87362149E0 | .2962455704 |

**Paired Samples Correlations**

| | | N | Correlation | Sig. |
|---|---|---|---|---|
| Pair 1 | Iso_Net_Joint & Dyn_Net_Joint | 40 | .779 | .000 |
| Pair 2 | ID_Net_Joint & Dyn_Vast_Lat | 40 | .469 | .002 |
| Pair 3 | ID_Net_Joint & Iso_Net_Joint | 40 | .562 | .000 |

# APPENDIX E.

# REGRESSIONS OUTPUTS

**Regression: Sagittal plane excursion**

**Model Summary**

| Model | R | R Square | Adjusted R Square | Std. Error of the Estimate | Change Statistics | | | | |
|---|---|---|---|---|---|---|---|---|---|
| | | | | | R Square Change | F Change | df1 | df2 | Sig. F Change |
| 1 | .208[a] | .043 | -.008 | 6.88834 | .043 | .840 | 2 | 37 | .440 |
| 2 | .458[b] | .210 | .066 | 6.62979 | .166 | 1.736 | 4 | 33 | .166 |

a. Predictors: (Constant), Height, Weight

b. Predictors: (Constant), Height, Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

**ANOVA[c]**

| Model | | Sum of Squares | df | Mean Square | F | Sig. |
|---|---|---|---|---|---|---|
| 1 | Regression | 79.734 | 2 | 39.867 | .840 | .440[a] |
| | Residual | 1755.624 | 37 | 47.449 | | |
| | Total | 1835.358 | 39 | | | |
| 2 | Regression | 384.871 | 6 | 64.145 | 1.459 | .223[b] |
| | Residual | 1450.487 | 33 | 43.954 | | |
| | Total | 1835.358 | 39 | | | |

a. Predictors: (Constant), Height, Weight

b. Predictors: (Constant), Height, Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

c. Dependent Variable: FlexionExcursion

**Coefficients[a]**

| Model | | Unstandardized Coefficients | | Standardized Coefficients | t | Sig. | Correlations | | |
|---|---|---|---|---|---|---|---|---|---|
| | | B | Std. Error | Beta | | | Zero-order | Partial | Part |
| 1 | (Constant) | -24.887 | 26.875 | | -.926 | .360 | | | |
| | Weight | .231 | .182 | .268 | 1.269 | .212 | .128 | .204 | .204 |
| | Height | -.203 | .199 | -.216 | -1.023 | .313 | -.043 | -.166 | -.165 |
| 2 | (Constant) | -20.236 | 27.874 | | -.726 | .473 | | | |
| | Weight | .244 | .194 | .283 | 1.257 | .217 | .128 | .214 | .195 |
| | Height | -.226 | .203 | -.240 | -1.113 | .274 | -.043 | -.190 | -.172 |
| | VastusLateralImpulse | -.160 | 1.155 | -.027 | -.138 | .891 | .015 | -.024 | -.021 |
| | VastusMedialisImpulse | .496 | 1.127 | .102 | .440 | .662 | -.039 | .076 | .068 |
| | BicepFemorisImpulse | -2.507 | 4.645 | -.144 | -.540 | .593 | .167 | -.094 | -.084 |
| | SemitendinousImpulse | 14.830 | 7.335 | .524 | 2.022 | .051 | .366 | .332 | .313 |

a. Dependent Variable: FlexionExcursion

**Regression: Peak extensor moment**

**Model Summary**

| Model | R | R Square | Adjusted R Square | Std. Error of the Estimate | Change Statistics | | | | |
|---|---|---|---|---|---|---|---|---|---|
| | | | | | R Square Change | F Change | df1 | df2 | Sig. F Change |
| 1 | .597[a] | .357 | .322 | 25.60413 | .357 | 10.266 | 2 | 37 | .000 |
| 2 | .683[b] | .467 | .370 | 24.67975 | .110 | 1.706 | 4 | 33 | .172 |

a. Predictors: (Constant), Height, Weight

b. Predictors: (Constant), Height, Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

**ANOVA[c]**

| Model | | Sum of Squares | df | Mean Square | F | Sig. |
|---|---|---|---|---|---|---|
| 1 | Regression | 13459.830 | 2 | 6729.915 | 10.266 | .000[a] |
| | Residual | 24256.143 | 37 | 655.571 | | |
| | Total | 37715.973 | 39 | | | |
| 2 | Regression | 17616.004 | 6 | 2936.001 | 4.820 | .001[b] |
| | Residual | 20099.969 | 33 | 609.090 | | |
| | Total | 37715.973 | 39 | | | |

a. Predictors: (Constant), Height, Weight

b. Predictors: (Constant), Height, Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

c. Dependent Variable: PeakKEM

**Coefficients[a]**

| Model | | Unstandardized Coefficients | | Standardized Coefficients | t | Sig. | Correlations | | |
|---|---|---|---|---|---|---|---|---|---|
| | | B | Std. Error | Beta | | | Zero-order | Partial | Part |
| 1 | (Constant) | -59.037 | 99.896 | | -.591 | .558 | | | |
| | Weight | 2.183 | .676 | .559 | 3.229 | .003 | .596 | .469 | .426 |
| | Height | .245 | .738 | .057 | .331 | .742 | .419 | .054 | .044 |
| 2 | (Constant) | -23.446 | 103.763 | | -.226 | .823 | | | |
| | Weight | 2.111 | .722 | .540 | 2.923 | .006 | .596 | .454 | .372 |
| | Height | -.039 | .756 | -.009 | -.052 | .959 | .419 | -.009 | -.007 |
| | VastusLateralImpulse | 6.538 | 4.301 | .248 | 1.520 | .138 | .364 | .256 | .193 |
| | VastusMedialisImpulse | -.344 | 4.194 | -.016 | -.082 | .935 | .469 | -.014 | -.010 |
| | BicepFemorisImpulse | 28.189 | 17.291 | .358 | 1.630 | .113 | -.091 | .273 | .207 |
| | SemitendinousImpulse | -50.883 | 27.305 | -.397 | -1.864 | .071 | -.249 | -.309 | -.237 |

a. Dependent Variable: PeakKEM

**Regression: Peak anterior shear force**

**Model Summary**

| Model | R | R Square | Adjusted R Square | Std. Error of the Estimate | Change Statistics | | | | |
|---|---|---|---|---|---|---|---|---|---|
| | | | | | R Square Change | F Change | df1 | df2 | Sig. F Change |
| 1 | .626[a] | .392 | .377 | 62.26358 | .392 | 24.551 | 1 | 38 | .000 |
| 2 | .688[b] | .474 | .396 | 61.26377 | .081 | 1.313 | 4 | 34 | .285 |

a. Predictors: (Constant), Weight

b. Predictors: (Constant), Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

**ANOVA[c]**

| Model | | Sum of Squares | df | Mean Square | F | Sig. |
|---|---|---|---|---|---|---|
| 1 | Regression | 95177.899 | 1 | 95177.899 | 24.551 | .000[a] |
| | Residual | 147316.620 | 38 | 3876.753 | | |
| | Total | 242494.519 | 39 | | | |
| 2 | Regression | 114884.042 | 5 | 22976.808 | 6.122 | .000[b] |
| | Residual | 127610.477 | 34 | 3753.249 | | |
| | Total | 242494.519 | 39 | | | |

a. Predictors: (Constant), Weight

b. Predictors: (Constant), Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

c. Dependent Variable: peakASF

**Coefficients[a]**

| Model | | Unstandardized Coefficients | | Standardized Coefficients | t | Sig. | Correlations | | |
|---|---|---|---|---|---|---|---|---|---|
| | | B | Std. Error | Beta | | | Zero-order | Partial | Part |
| 1 | (Constant) | 26.214 | 75.874 | | .345 | .732 | | | |
| | Weight | 6.208 | 1.253 | .626 | 4.955 | .000 | .626 | .626 | .626 |
| 2 | (Constant) | -.728 | 80.091 | | -.009 | .993 | | | |
| | Weight | 6.306 | 1.468 | .636 | 4.296 | .000 | .626 | .593 | .534 |
| | VastusLateralImpulse | 17.403 | 10.669 | .260 | 1.631 | .112 | .361 | .269 | .203 |
| | VastusMedialisImpulse | -11.621 | 10.268 | -.209 | -1.132 | .266 | .328 | -.191 | -.141 |
| | BicepFemorisImpulse | 37.284 | 42.151 | .187 | .885 | .383 | -.229 | .150 | .110 |
| | SemitendinousImpulse | -111.162 | 67.649 | -.342 | -1.643 | .110 | -.270 | -.271 | -.204 |

a. Dependent Variable: peakASF

**Regression: Frontal plane excursion**

**Model Summary**

| Model | R | R Square | Adjusted R Square | Std. Error of the Estimate | Change Statistics | | | | |
|---|---|---|---|---|---|---|---|---|---|
| | | | | | R Square Change | F Change | df1 | df2 | Sig. F Change |
| 1 | .386[a] | .149 | .103 | 4.85079 | .149 | 3.240 | 2 | 37 | .050 |
| 2 | .422[b] | .178 | .028 | 5.04865 | .029 | .289 | 4 | 33 | .883 |

a. Predictors: (Constant), Height, Weight

b. Predictors: (Constant), Height, Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

**ANOVA[c]**

| Model | | Sum of Squares | df | Mean Square | F | Sig. |
|---|---|---|---|---|---|---|
| 1 | Regression | 152.495 | 2 | 76.247 | 3.240 | .050[a] |
| | Residual | 870.617 | 37 | 23.530 | | |
| | Total | 1023.112 | 39 | | | |
| 2 | Regression | 181.979 | 6 | 30.330 | 1.190 | .336[b] |
| | Residual | 841.133 | 33 | 25.489 | | |
| | Total | 1023.112 | 39 | | | |

a. Predictors: (Constant), Height, Weight

b. Predictors: (Constant), Height, Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

c. Dependent Variable: FrontalExcursion

**Coefficients[a]**

| Model | | Unstandardized Coefficients | | Standardized Coefficients | t | Sig. | Correlations | | |
|---|---|---|---|---|---|---|---|---|---|
| | | B | Std. Error | Beta | | | Zero-order | Partial | Part |
| 1 | (Constant) | -50.536 | 18.926 | | -2.670 | .011 | | | |
| | Weight | -.043 | .128 | -.066 | -.333 | .741 | .209 | -.055 | -.050 |
| | Height | .299 | .140 | .426 | 2.139 | .039 | .383 | .332 | .324 |
| 2 | (Constant) | -51.546 | 21.227 | | -2.428 | .021 | | | |
| | Weight | -.082 | .148 | -.127 | -.554 | .583 | .209 | -.096 | -.087 |
| | Height | .312 | .155 | .444 | 2.014 | .052 | .383 | .331 | .318 |
| | VastusLateralImpulse | .044 | .880 | .010 | .050 | .961 | .111 | .009 | .008 |
| | VastusMedialisImpulse | .172 | .858 | .048 | .200 | .842 | .233 | .035 | .032 |
| | BicepFemorisImpulse | -.814 | 3.537 | -.063 | -.230 | .819 | -.086 | -.040 | -.036 |
| | SemitendinousImpulse | -2.207 | 5.586 | -.104 | -.395 | .695 | -.164 | -.069 | -.062 |

a. Dependent Variable: FrontalExcursion

**Regression: Peak frontal plane moment**

**Model Summary**

| Model | R | R Square | Adjusted R Square | Std. Error of the Estimate | Change Statistics: R Square Change | F Change | df1 | df2 | Sig. F Change |
|---|---|---|---|---|---|---|---|---|---|
| 1 | .215[a] | .046 | -.005 | 15.25416 | .046 | .898 | 2 | 37 | .416 |
| 2 | .583[b] | .340 | .220 | 13.43781 | .294 | 3.670 | 4 | 33 | .014 |

a. Predictors: (Constant), Height, Weight

b. Predictors: (Constant), Height, Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

**ANOVA[c]**

| Model | | Sum of Squares | df | Mean Square | F | Sig. |
|---|---|---|---|---|---|---|
| 1 | Regression | 417.720 | 2 | 208.860 | .898 | .416[a] |
| | Residual | 8609.504 | 37 | 232.689 | | |
| | Total | 9027.224 | 39 | | | |
| 2 | Regression | 3068.253 | 6 | 511.376 | 2.832 | .025[b] |
| | Residual | 5958.970 | 33 | 180.575 | | |
| | Total | 9027.224 | 39 | | | |

a. Predictors: (Constant), Height, Weight

b. Predictors: (Constant), Height, Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

c. Dependent Variable: PeakKVM

**Coefficients[a]**

| Model | | Unstandardized Coefficients: B | Unstandardized Coefficients: Std. Error | Standardized Coefficients: Beta | t | Sig. | Correlations: Zero-order | Partial | Part |
|---|---|---|---|---|---|---|---|---|---|
| 1 | (Constant) | -82.772 | 59.515 | | -1.391 | .173 | | | |
| | Weight | -.391 | .403 | -.204 | -.970 | .338 | -.023 | -.157 | -.156 |
| | Height | .586 | .440 | .281 | 1.332 | .191 | .148 | .214 | .214 |
| 2 | (Constant) | -76.914 | 56.498 | | -1.361 | .183 | | | |
| | Weight | -1.033 | .393 | -.540 | -2.626 | .013 | -.023 | -.416 | -.371 |
| | Height | .687 | .412 | .329 | 1.667 | .105 | .148 | .279 | .236 |
| | VastusLateralImpulse | -.274 | 2.342 | -.021 | -.117 | .908 | .260 | -.020 | -.017 |
| | VastusMedialisImpulse | 5.527 | 2.284 | .514 | 2.420 | .021 | .322 | .388 | .342 |
| | BicepFemorisImpulse | -24.484 | 9.415 | -.636 | -2.601 | .014 | -.301 | -.412 | -.368 |
| | SemitendinousImpulse | 21.880 | 14.867 | .349 | 1.472 | .151 | -.209 | .248 | .208 |

a. Dependent Variable: PeakKVM

**Regression: Transverse Plane excursion**

**Model Summary**

| Model | R | R Square | Adjusted R Square | Std. Error of the Estimate | Change Statistics | | | | |
|---|---|---|---|---|---|---|---|---|---|
| | | | | | R Square Change | F Change | df1 | df2 | Sig. F Change |
| 1 | .193[a] | .037 | -.015 | 4.80708 | .037 | .718 | 2 | 37 | .494 |
| 2 | .231[b] | .053 | -.119 | 5.04733 | .016 | .140 | 4 | 33 | .966 |

a. Predictors: (Constant), Height, Weight

b. Predictors: (Constant), Height, Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

**ANOVA[c]**

| Model | | Sum of Squares | df | Mean Square | F | Sig. |
|---|---|---|---|---|---|---|
| 1 | Regression | 33.204 | 2 | 16.602 | .718 | .494[a] |
| | Residual | 854.995 | 37 | 23.108 | | |
| | Total | 888.199 | 39 | | | |
| 2 | Regression | 47.507 | 6 | 7.918 | .311 | .927[b] |
| | Residual | 840.691 | 33 | 25.475 | | |
| | Total | 888.199 | 39 | | | |

a. Predictors: (Constant), Height, Weight

b. Predictors: (Constant), Height, Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

c. Dependent Variable: TransverseExcursion

**Coefficients[a]**

| Model | | Unstandardized Coefficients | | Standardized Coefficients | t | Sig. | Correlations | | |
|---|---|---|---|---|---|---|---|---|---|
| | | B | Std. Error | Beta | | | Zero-order | Partial | Part |
| 1 | (Constant) | -25.436 | 18.755 | | -1.356 | .183 | | | |
| | Weight | -.130 | .127 | -.217 | -1.024 | .313 | -.061 | -.166 | -.165 |
| | Height | .158 | .139 | .241 | 1.138 | .262 | .101 | .184 | .184 |
| 2 | (Constant) | -24.542 | 21.221 | | -1.156 | .256 | | | |
| | Weight | -.128 | .148 | -.213 | -.865 | .393 | -.061 | -.149 | -.147 |
| | Height | .156 | .155 | .238 | 1.005 | .322 | .101 | .172 | .170 |
| | VastusLateralImpulse | -.498 | .880 | -.123 | -.566 | .575 | -.104 | -.098 | -.096 |
| | VastusMedialisImpulse | .265 | .858 | .079 | .309 | .759 | -.019 | .054 | .052 |
| | BicepFemorisImpulse | -1.326 | 3.536 | -.110 | -.375 | .710 | .093 | -.065 | -.064 |
| | SemitendinousImpulse | 3.120 | 5.584 | .158 | .559 | .580 | .098 | .097 | .095 |

a. Dependent Variable: TransverseExcursion

**Regression: Peak transverse plane moment**

**Model Summary**

| Model | R | R Square | Adjusted R Square | Std. Error of the Estimate | Change Statistics | | | | |
|---|---|---|---|---|---|---|---|---|---|
| | | | | | R Square Change | F Change | df1 | df2 | Sig. F Change |
| 1 | .244[a] | .059 | .008 | 4.97115 | .059 | 1.167 | 2 | 37 | .323 |
| 2 | .516[b] | .266 | .132 | 4.65008 | .207 | 2.321 | 4 | 33 | .077 |

a. Predictors: (Constant), Height, Weight

b. Predictors: (Constant), Height, Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

**ANOVA[c]**

| Model | | Sum of Squares | df | Mean Square | F | Sig. |
|---|---|---|---|---|---|---|
| 1 | Regression | 57.674 | 2 | 28.837 | 1.167 | .323[a] |
| | Residual | 914.357 | 37 | 24.712 | | |
| | Total | 972.031 | 39 | | | |
| 2 | Regression | 258.465 | 6 | 43.078 | 1.992 | .095[b] |
| | Residual | 713.566 | 33 | 21.623 | | |
| | Total | 972.031 | 39 | | | |

a. Predictors: (Constant), Height, Weight

b. Predictors: (Constant), Height, Weight, SemitendinousImpulse, VastusLateralImpulse, VastusMedialisImpulse, BicepFemorisImpulse

c. Dependent Variable: peakKRM

**Coefficients[a]**

| Model | | Unstandardized Coefficients | | Standardized Coefficients | t | Sig. | Correlations | | |
|---|---|---|---|---|---|---|---|---|---|
| | | B | Std. Error | Beta | | | Zero-order | Partial | Part |
| 1 | (Constant) | -36.388 | 19.395 | | -1.876 | .069 | | | |
| | Weight | -.132 | .131 | -.210 | -1.004 | .322 | -.003 | -.163 | -.160 |
| | Height | .219 | .143 | .320 | 1.528 | .135 | .184 | .244 | .244 |
| 2 | (Constant) | -28.813 | 19.551 | | -1.474 | .150 | | | |
| | Weight | -.230 | .136 | -.367 | -1.692 | .100 | -.003 | -.282 | -.252 |
| | Height | .203 | .143 | .297 | 1.427 | .163 | .184 | .241 | .213 |
| | VastusLateralImpulse | -1.871 | .810 | -.442 | -2.308 | .027 | -.140 | -.373 | -.344 |
| | VastusMedialisImpulse | 2.041 | .790 | .578 | 2.582 | .014 | .180 | .410 | .385 |
| | BicepFemorisImpulse | -7.425 | 3.258 | -.588 | -2.279 | .029 | .010 | -.369 | -.340 |
| | SemitendinousImpulse | 11.747 | 5.145 | .570 | 2.283 | .029 | .088 | .369 | .341 |

a. Dependent Variable: peakKRM