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**MUSCLE CONTRIBUTIONS TO KNEE JOINT STABILITY: EFFECTS OF ACL  
INJURY AND KNEE BRACE USE**

by

**Aaron Derouin**

**A Thesis  
Submitted to the Faculty of Graduate Studies and Research  
through Kinesiology  
in Partial Fulfillment of the Requirements for  
the Degree of Master of Human Kinetics at the  
University of Windsor**

**Windsor, Ontario, Canada**

**2006**

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## ABSTRACT

### MUSCLE CONTRIBUTIONS TO KNEE JOINT STABILITY: EFFECTS OF ACL INJURY AND KNEE BRACE USE

Aaron Joseph Derouin  
University of Windsor, 2006

The purpose of this study was to develop an EMG-based biomechanical model of the lower extremity for the purpose of evaluating the rotational joint stiffness of ACL-deficient (n=9) and reconstructed (n=8) knees, with and without a functional knee brace. Kinematic, EMG, and kinetic data were measured, while subjects performed stable and unstable quasi-static squats, and were incorporated into biomechanical and anatomical models. Individual and total muscle contributions to knee joint rotational stiffness were calculated about the flexion-extension axis, using the method developed by Potvin and Brown (2005) for four independent variables: Squat task; Leg condition (uninjured, unbraced, braced); Knee angle (5-10°, 20-25°, 40-45°); Injury Status. Subjects had significantly higher ( $p < 0.05$ ) total knee joint stiffness values while wearing the brace compared to the control leg. The difference in knee joint stiffness values in the unstable trials compared to stable trials was significantly greater at 5-10° versus 20-25° and 40-45°.

## ACKNOWLEDGEMENTS

The completion of this project would not have been possible without the valuable contributions and input from several significant individuals. I would like to first acknowledge God, for blessing me with the opportunity and curiosity to undertake such a complex study and the ability, determination, and perseverance to finish.

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“What lies behind us and what lies before us are tiny matters compared to what lies within us.”

Ralph Waldo Emerson

*I Love This Game!!!*

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## LIST OF ABBREVIATIONS

AAOS: American Academy of Orthopaedic Surgeons

ACL: Anterior Cruciate Ligament

AIS: Anterior Inferior Iliac Spine

AMB: Anterior Medial Band

ANOVA: Analysis of Variance

ATT: Anterior Tibial Translation

BF: Biceps Femoris

BF-L: Biceps Femoris Long Head

BF-S: Biceps Femoris Short Head

BMI: Body Mass Index

CCI: Co-contraction Index

CMRR: Common Mode Rejection Ratio

COM: Center of Mass

COP: Center of Pressure

EMG: Electromyography

FKB: Functional Knee Brace

GA: Gastrocnemius

GA-L: Gastrocnemius Lateralis

GA-M: Gastrocnemius Medialis

GR: Gracilis

GTO: Golgi Tendon Organs



IKDC: International Knee Documentation Committee

$I_O$ : Moment of Inertia

ITT: Iliotibial Tract

$JS_R$ : Joint Rotational Stiffness

k: Stiffness

KOS: Knee Outcome Survey

LCL: Lateral Collateral Ligament

MCL: Medial Collateral Ligament

$M_K$ : Net Knee Reaction Moment

MRI: Magnetic Resonance Imaging

MVC: Maximum Voluntary Contraction

OA: Osteoarthritis

PCL: Posterior Cruciate Ligament

PF: Patellofemoral

PL: Patellar Ligament

PLB: Posterior Lateral Band

QF: Quadriceps Femoris

RF: Rectus Femoris

$r_z$ : Functional Moment Arm about Z-axis

SA: Sartorius

sEMG: Surface Electromyography

SM: Semimembranosus

$Sm_z$ : Overall Muscle Stability

ST: Semitendinosus

TF: Tibiofemoral

TFL: Tensor Fascia Latae

V: Potential Energy

VI: Vastus Intermedius

VL: Vastus Lateralis

VM: Vastus Medialis

## CHAPTER 1

# INTRODUCTION

### 1.1 BACKGROUND

Disruption to the anterior cruciate ligament (ACL) is the most frequently occurring ligament injury of the knee (Feagin and Curl, 1976) and is one of the most debilitating injuries of the lower extremity. "The increasing popularity of physical fitness, combined with activities that place the cruciate ligament at risk, seems to be causing an almost exponential increase in the number of serious knee disruptions"(Ryder et al., 1997; Johnson, 1988). In 1982, knee injuries in the United States were defined as a national health problem (Feagin and Lambert, 1985), and today the incidence of serious knee injuries continues to escalate. The estimated annual cost of surgical treatment is about \$1.5 billion, alone, without consideration of the cost of the initial evaluation and treatment of those injured, the nonsurgical care of remaining patients, or the future of medical treatment for those who develop post-traumatic arthritis (Kao et al., 1995). Disruption to the ACL has become so commonplace that the seriousness of the injury is often overlooked. In fact, the careers of thousands of athletes have been hampered or completely terminated as a result of ACL disruption. The vulnerability of the knee to injury may be accounted for by the fact that "no other joint in the leg depends less for its stability on the shapes of its constituent articulating surfaces" (Laurence and Strachan, 1970) and that the knee joint is between the two longest bones in the body.

Each year in the United States, it is estimated that there are approximately 250,000 ACL injuries, or 1 in 3,000 in the general population (Miyasaka et al., 1991). In

an analysis of athletic population segments, such as football, skiing and soccer, the incidence of ACL injury is even higher (Smith et al., 1993).

Functionally, the ACL is the primary restraint to anterior translation of the tibia, as it provides an average of 86 per cent of the total resisting force to anterior drawer (Butler et al., 1980). The intersection of ACL together with the posterior cruciate ligament defines the axis of rotation of the knee joint. Furthermore, the changing position and relative orientation of the cruciate ligaments throughout the range of knee motion is responsible for the rolling and gliding functions of the knee. “The ACL, in particular, serves a highly specialized role in guiding knee motion, one that is vital to joint stability and the maintenance of normal knee function and overall kinematics” (Smith et al., 1993). In addition to its structural stabilizing role, the ACL is thought to provide a stabilizing function through a neurological feedback mechanism (Goldstein and Bosco, 2001). Mechanoreceptors, such as Golgi tendon-like organs, Pacinian and Ruffini corpuscles, have been identified within the ACL and other joint structures. These receptors are thought to be responsible for this feedback mechanism. It is postulated that these receptors, when stimulated, provide afferent input along with information supplied by muscle spindles and joint Golgi tendon organs (GTO's), to the alpha and gamma motor pathways to modulate muscle stiffness.

A loss of integrity to the ACL causes significant alterations to knee joint kinematics and the recruitment patterns of muscles that surround the knee. The primary concerns following injury to this ligament are: functional impairment and the development of symptomatic knee instability (Evans and Stannsih, 2001). Instability rates, according to Frank and Jackson (1997), range from 16% to nearly 100%, with

instability defined as giving way during activity (Lysholm and Gillquist, 1982). Without appropriate treatment, a complete rupture of the ACL may result in progressive knee instability, which in turn leads to recurrent intra-articular damage and eventual osteoarthritis (OA) (Daniel et al., 1994). Furthermore, it is difficult to select patients for a specific treatment protocol because the relationship between quantitatively measured laxity, such as that determined by a knee joint arthrometer, is not at all related to the functional status of the patient. This inability to predict the functional outcome of ACL deficient patients is further compounded by the fact that not every patient displays symptoms of instability and that not all knees with ongoing ACL insufficiency progress to developing osteoarthritis (Evans and Stannish, 2001).

Typically, the focus of treatment for the ACL deficient patient is the restoration of normal knee joint mechanics in an effort to relieve pain, reduce instability, and minimize the risk of post-traumatic osteoarthritis (MacWilliams et al., 1999). The treatment options available to the ACL injured individual are surgical reconstruction and non-operative treatment. Reconstruction is generally advocated for the young active individual, whereas non-operative management is recommended to patients with low athletic demands or sedentary occupations (Segawa et al., 2001). Recently, the term non-operative management has been clarified and described as a progressive exercise based program which may include the following components: isometric, isotonic, isokinetic, closed chain exercises, and functional exercises involving perturbation training and sport-specific agility exercises. An additional treatment option prescribed to both surgically and non-surgically managed ACL-deficient patients is the use of functional knee braces. In a recent survey of the American Academy of Orthopaedic Surgeons on the topic of brace

prescription practices, only 13% and 3% of physicians never prescribed functional braces to ACL-reconstructed and ACL-deficient patients, respectively.

Despite the overwhelming recommendations of functional knee brace use by physicians that manage ACL injured patients, there exists significant controversy in the literature regarding the effectiveness of these braces in stabilizing the knee. Cawley et al. (1991) reviewed the subjective response data from several studies and found that 90% of subjects reported fewer episodes of giving way (instability) and were functionally improved with brace use. Subjective response, although a valuable tool to the researcher, must always be correlated to biomechanical measures (Cawley et al., 1991). The ability of functional knee braces to passively restrain anterior tibial translation in cadaveric specimens has been demonstrated by several authors to occur at relatively low loads (140 N) (Beynon et al., 2004), well below estimated physiological loads of 400N (Noyes et al., 1980). Despite these consistent findings, the inability of functional knee braces to control tibial displacement under high physiological loads has yet to be validated in vivo. Loading of the knee in vivo with physiological loads is a significant ethical constraint that has yet to be resolved.

Recently, electromyography (EMG) has been used to evaluate the effectiveness of knee braces in modifying muscle activity in favourable manner to stabilize the ACL-deficient knee. The effects of functional knee bracing have been demonstrated by a limited number of studies to exert its influence on the neuromuscular system through various mechanisms. Compensatory recruitment of the hamstrings and quadriceps is required following ACL injury to maintain stability of the knee joint. Although, several investigators (Acierno et al., 1995; Branch et al., 1989; Ramsey et al., 2003; Smith et al.,

2003; Wojtys et al., 1996b) have established that bracing the ACL deficient patient alters the EMG activity of knee musculature, no conclusive evidence exists that these changes actually enhance joint stability. Unfortunately, these modifications to muscular activity are speculatively postulated as favourable or unfavourable based on qualitative data. Favourable responses to knee bracing, as described by Smith et al. (2003), were considered as: earlier activation (shorter onset latency) of either the hamstrings or gastrocnemius muscles groups relative to the quadriceps muscle groups. The results of this study, along with the findings of some other investigations (Wojtys and Huston, 1994; Wojtys et al., 1996a), suggest that inter-individual differences exist in muscle reflex patterns and that there is a lack of a universal recruitment pattern in response to anterior tibiofemoral shear. To date, no known study has quantitatively assessed how functional knee bracing alters mechanical knee joint stability in ACL-deficient patients.

Doorenbosch and Harlaar (2003) quantified co-activation of the hamstrings and quadriceps using a co-contraction index and found that this index was significantly higher in ACL-deficient subjects compared to normal subjects performing isokinetic contractions about the knee. It was suggested that quantifying co-contraction is a useful parameter to evaluate clinical interventions and rehabilitation processes. The usefulness of this approach in assessing joint stability is limited by one central assumption.

Rotational equilibrium of the joint is achieved when flexor and extensor torques are applied simultaneously and if the flexing and extending moments of the muscle forces are equal. However, this does not necessarily mean that the components of the flexor and extensor forces parallel to the tibial plateau will also be equal and opposite (O'Connor, 1993). Any difference in these shear forces has to be taken up by ligament forces. In the

absence of the ACL, the collateral ligaments and joint capsule must resist anterior tibial displacement that occurs when the shear force generated by the quadriceps is greater than that of the hamstrings. O'Connor (1993) identified a critical flexion angle of 22 degrees, where these respective parallel components exactly balance each other and no ligament forces are required. Although, static equilibrium is one of the required conditions to calculate joint stability, it does not guarantee that a particular musculoskeletal system is stable from a mechanical perspective. No known study to date has calculated the mechanical stability of the healthy or ACL-deficient knee.

In 1989, Bergmark's quantitative assessment of lumbar spine stability was the first study to fully examine and define the mechanical stability of a muscular system. He affirmed that when the Potential Energy ( $V$ ) of an entire system is at a relative minimum then the muscular system can be described as stable. Thus, a system in a stable state of equilibrium must always be capable of returning to its original state of equilibrium in response to a perturbation around this state. Two conditions must be satisfied in order for a system to be classified as stable. Initially, the system must be in a state of rotational or mechanical equilibrium, that is the first derivative of  $V$  (net moment) with respect to angle ( $\theta$ ) must be zero. Next, the second derivative of  $V$  must be positive definite (greater than zero) indicating that  $V$  is at some minimum.

Muscular contributions, to the potential energy of a system during a perturbation, are dependent on how this perturbation changes the length of the muscle and the orientation of the muscles relative to the joint being examined. The ability of a muscle to store or release elastic energy is related to its stiffness, which in turn depends on the length and the pre-activation force of the muscle. The stiffness provided by muscles plays



a very significant role in stabilizing the muscles surrounding a joint and/or system of joints.

A novel approach to calculating stability about any specified joint has recently been proposed by (Potvin and Brown, 2005). This method is suggested to be mathematically less complex than the approach used by Bergmark and subsequent researchers (Cholewicki and McGill, 1996; Crisco and Panjabi, 1991; Granta and Wilson, 2001). The method used by Bergmark is thought to be limited to only few researchers, as the mathematical analysis required to determine stability limits its application. Additionally, this new approach allows the determination of stability for a multi-muscle system, overcoming the shortcomings of the method used by Cholewicki et al. (1999) and Granata and Orishimo (2001), which is limited to one single-equivalent flexor and one single upright single-equivalent extensor muscle. According to Potvin and Brown (2005), the stability of any specified joint may be assessed with an EMG-driven biomechanical model given the coordinates of the individual muscles that cross the joint, the pre-activation level of the muscle prior to being perturbed, the muscles functional moment arm, the length of the muscle, the stiffness of the muscle and the potential energy of the system being examined.

To date, the efficacy of functional knee braces in stabilizing the ACL deficient knee is equivocal. Unfortunately, the scientific validation of functional knee braces has not kept pace with the proliferation of knee brace companies and their innovative designs. The evaluation of the effectiveness of braces using knee arthrometers on cadaveric specimens has proven to be redundant, as the loads employed and the testing mode do not approximate the normal physiologic condition (Cawley et al., 1991). Functional knee

braces are designed for athletically active individuals wishing to return to their pre-morbidity status and, consequently, should be evaluated under conditions that duplicate their normal environment. Up to now, relatively few studies have attempted to elucidate the effects of functional knee braces on muscle control in the ACL-deficient knee.

Numerous researchers (Acierno et al., 1995; Branch et al., 1989; Ramsey et al., 2003; Smith et al., 2003; Wojtys et al., 1996b) have all documented notable changes to EMG amplitudes and patterns of ACL-deficient patients with brace use. However, the stability that functional knee braces potentially provide, thus far, has only been qualitatively described. The new approach to calculating joint stability proposed by Potvin and Brown (2005) will enable a quantitative evaluation of the stabilizing potential of functional knee braces.

## **1.2 PURPOSE**

The goal of this study was to develop an EMG-based biomechanical model of the lower extremity for the purpose of evaluating the rotational joint stiffness ( $JS_R$ ) of the ACL injured knee with and without a functional knee brace.  $JS_R$  was evaluated for two single-leg squat tasks, a stable isometric squat and an unstable isometric squat, performed on a custom fabricated fitter-like wobble board. The individual contributions, and total contribution, of leg muscles to knee  $JS_R$  were assessed at three knee joint angle ranges (5-10, 20-25, and 40-45 degrees) that the subject was instructed to maintain over the course of a trial.

### 1.3 HYPOTHESES

#### 1.3.1a Overall Knee Joint Rotational Stiffness

*Total muscle contributions to knee flexion-extension  $JS_R$  will show a statistically significant main effect ( $p < 0.05$ ) between leg conditions. Post hoc analysis will reveal that the brace involved leg condition will have significantly lower total knee  $JS_R$  compared to the unbraced involved leg.*

In two studies, a statistically significant decrease in hamstrings torque or muscle activity has been demonstrated while wearing a functional knee brace (Acierno et al., 1995; Wojtys et al., 1996b). It is hypothesized that the proprioceptive feedback that the brace provides will optimize the gain on neuromotor control of the hamstrings and will decrease its activity. Preliminary biomechanical modelling has shown that particular knee flexors, such as biceps femoris long and short head, are destabilizing and have negative stability values when the model is positioned in a neutral posture (neutral hip extension and full knee extension). The brace may provide the nervous system with sufficient proprioceptive information to decrease the gain on the hamstrings throughout a range of joint angles that are destabilizing.

#### 1.3.1b Overall Knee Joint Rotational Stiffness

*Total muscle contributions to knee flexion-extension  $JS_R$  will show a statistically significant interaction ( $p < 0.05$ ) between leg condition and squat task. Further Post hoc analysis will reveal that the highest total  $JS_R$  will be displayed for the unbraced involved leg performing the unstable single leg squat, compared to any other leg x squat task combination.*

Typically, squat exercises performed on a wobble board provide a greater challenge to the neuromuscular system, and result in increased co-activation of the flexors and extensors of the knee, compared to similar stable squat exercises. Kean *et al.* (2006) found that a fixed foot balance training program on a wobble board significantly increased rectus femoris muscle activation during the landing phase of a jumping test, compared to functionally directed training (supervised jump training). Additionally, static balance improved 33% more with the fixed foot protocol compared to functionally directed training. In a different study, both conservatively managed ACL deficient and surgically treated subjects had significantly higher hamstrings co-activation patterns compared to that of the normal-control subjects (Grabiner and Weiker, 1993). It is hypothesized that the proprioceptive feedback that the brace provides will optimize the gain on neuromotor control of the hamstrings and will decrease its activity. The brace may provide the nervous system with sufficient proprioceptive information to decrease the gain on the hamstrings throughout a range of joint angles that are destabilizing. It is expected that the unstable squat task will elicit greater reliance on the proprioceptive information provided by the brace, as it has a greater requirement for mechanical stability.

### 1.3.2 Individual Muscle Contributions to Knee Joint Rotational Stiffness

*The vastus lateralis will exhibit the greatest contribution to knee flexion-extension  $JS_R$  of all the extensor muscles, while the semimembranosus will display the highest  $JS_R$  of all the knee flexors. The ratio of vastus lateralis and semimembranosus EMG will be used to quantify hamstring co-activation and to determine if any qualitative relationship exists*

*between co-contraction and rotational joint stiffness. This will be calculated by dividing the percent maximum voluntary contraction (%MVC) for the semimembranosus by the %MVC for of the vastus lateralis.*

In the literature, it is widely regarded that co-activation of the hamstrings and quadriceps is the primary mechanism that is used by the neuromuscular system to enhance stability about the knee joint in ACL injured patients. According to Delp et al. (1990), the vastus lateralis and the semimembranosus have the greatest maximal force capacity of the muscles in the extensor and flexor compartments, respectively. Additionally, preliminary biomechanical modelling (Derouin and Potvin, 2005) has shown that the vastus lateralis and semimembranosus have the greatest potential to stabilize the knee joint in a squat posture.

## CHAPTER 2

### REVIEW OF LITERATURE

In order to understand the destabilizing effects of injury to the anterior cruciate ligament it is first necessary to describe the anatomy of the largest and the most complex joint in the body, the knee (Tortora and Grabowski, 1996)

#### 2.1 ANATOMY OF THE KNEE

##### 2.1.1 Passive Tissues

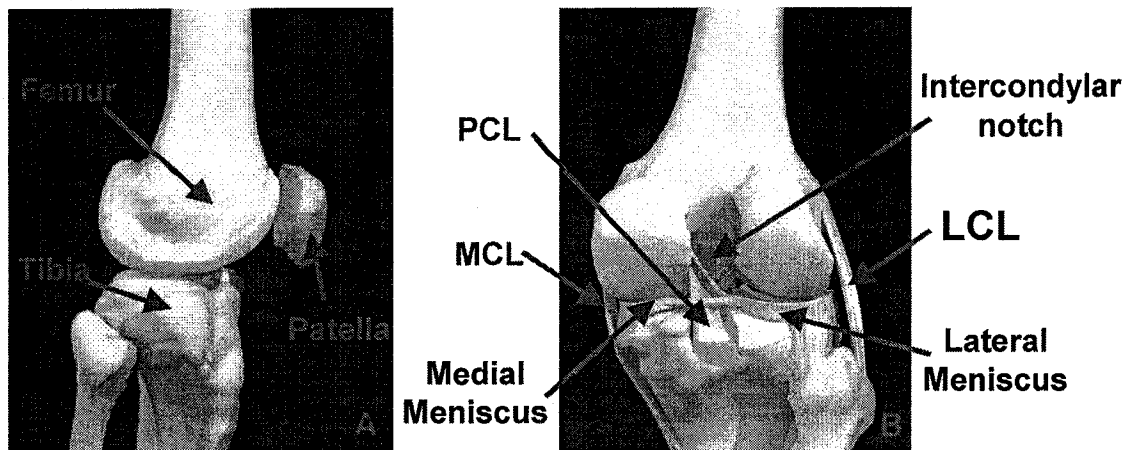
The knee is composed of 3 bones, 3 joints, 5 primary ligaments and several other accessory structures. The distal end of the femur, proximal portion of the tibia and the patella make up the knee. The inherent complexity of the knee may be attributed to the uniquely shaped articular surfaces of the femoral condyles, the tibial plateau and to the configuration of the cruciate ligaments.

##### *2.1.1.1 Bone Anatomy*

The femur is the longest, strongest bone in the body and its distal end is broadened for articulation with the tibia (Moore, 1992). The articular surfaces of the condyles are pulley shaped and are convex in both the sagittal and frontal planes (Kapandji, 1987) (Figure 1A). The articular surface of the medial femoral condyle is somewhat larger than that of the lateral condyle (Brantigan and Voshell, 1941). The neck of the pulley is represented by the central groove on the patellar surface anteriorly and by the intercondylar notch posteriorly (Kapandji, 1987). The curvature of the femoral condyles associated with the tibial condyles is distinctively smaller than the

superior/anterior surfaces of femoral condyles related to the patella and is primarily responsible for limiting hyperextension of the knee (Brantigan and Voshell, 1941).

In an examination of 100 cadaveric knees, Brantigan and Voshell (1941) found that the knee will go into greater than 90 degrees of hyperextension after cutting away the femoral condyles and leaving the ligaments intact.



**Figure 1.** A) Sagittal view of bone anatomy of knee joint. B) Posterior view of ligament anatomy of knee joint. The ACL is not shown

The tibia is the second largest bone of the skeleton and is situated on the anteromedial side of the leg (Moore, 1992). The proximal surface is expansive compared to the rest of the tibia, due to its articulation with the large condyles of the femur. The tibial surfaces are reciprocally curved to the femoral condyles and consist of two curved and concave parallel gutters that are separated by the intercondylar eminence (Kapandji, 1987). More precisely, the medial condyle of the tibia is biconcave in both the sagittal and frontal plane, while the lateral condyle is concave in the frontal plane and convex in the sagittal plane. The medial femoral condyle is relatively stable inside the concave medial tibial condyle, whereas the lateral femoral condyle is unstable as it travels on the convex surface of the lateral tibial condyle, with its stability during movement dependant

on the integrity of the anterior cruciate ligament. The intercondylar eminence is oriented in an anterior-posterior direction and fits into the intercondylar notch between the femoral condyles. The planed shape of the intercondylar eminence acts as a pivot, allowing axial rotation of the tibia through a vertical axis that passes through the medial intercondylar spine (Kapandji, 1987). An additional noteworthy feature of the tibia is the prominent tibial tuberosity, located anteriorly, where the patellar ligament inserts (Moore, 1992).

The patella, or kneecap, is a triangular shaped sesamoid bone embedded in the quadriceps femoris tendon with its apex directed inferiorly (Moore, 1992). The posterior surface of the patella, which articulates with the femur, is comprised of two biconcave facets. The lateral and medial facets are separated by the median vertical ridge.

The 3 joints that comprise the knee are: 1) the patellofemoral joint between the inner surface of the patella and the patellar surface of the femur; 2) the lateral tibiofemoral joint between the lateral condyle of the femur, lateral meniscus, and lateral condyle of the tibia; and 3) the medial tibiofemoral joint between the medial condyle of the femur, medial meniscus, and medial condyle of the tibia (Tortora and Grabowski, 1996). Tortora and Grabowski (1996) describe the patellofemoral joint as a gliding joint. Generally, the tibiofemoral joint may be considered a modified hinge joint, referring to the ability of the tibiofemoral joint to produce moderate amounts of transverse rotational motion, small amounts of varus-valgus or frontal plane motion, in addition to the flexion-extension motion in the sagittal plane.

#### *2.1.1.2 Ligament Anatomy*

Ligaments are relatively inelastic and highly adapted tissues comprised of short bands of dense fibrous tissue organized into parallel bundles of fibres (Pope, 1994),



which connect bone to bone. Ligaments are obligated to function in a state of tension, with its functional position considered as the position in which a vast majority of its fibres are taut (Fuss, 1989). This state of ligament tension is dependent on whether or not the distance between the origin and insertion of the ligament fibres remains constant or variable as the knee joint moves throughout its normal range of motion (Fuss, 1989). The variability of this distance is dependent both on the osseous attachment of the ligaments and the surface anatomy of the tibiofemoral articular surfaces. They function at lower applied loads to guide joint motion, at higher loads to limit joint motion and to assist other joint structures in the protection of periarticular soft tissues during both normal and pathological knee motions (Smith *et al.*, 1993).

The 5 primary structural ligaments in the knee are the two collateral ligaments (medial and lateral), the two cruciate ligaments (anterior and posterior) and the patellar ligament/tendon (Figure 1B). These ligaments along with the joint capsule and the menisci (articular cartilage) are responsible for stabilizing the knee during movement. According to Ryder *et al.*(1997), the cruciate ligaments form the central pivot and serve as the key components to stability of the femoral-tibial articulation. Several other accessory structures not as significant to stability of the knee include: the medial and lateral patellar retinaculae, the infrapatellar fat pad, the oblique and arcuate popliteal ligaments, the transverse ligaments of the menisci, and the bursae.

### **Collateral Ligaments**

The collateral ligaments are mostly responsible for providing medial and lateral translational and rotational stability to the knee. The tibial (medial) collateral ligament (MCL) is a broad flat ligament on the medial surface of the knee tibiofemoral joint that

extends from the medial condyle of the femur to the medial condyle of the tibia (Tortora and Grabowski, 1996). The MCL functions primarily as a passive medial joint stabilizer, as it is taut throughout flexion. It also aids in restraining hyperextension and provides the main valgus stability in these positions (Pope, 1994).

The fibular (lateral) collateral ligament (LCL) is positioned on the lateral aspect of the tibiofemoral joint and originates on the lateral condyle of the femur and inserts onto the lateral side of the head of the fibula. The LCL is a rounded strong ligament that is tense in knee extension but relaxed in knee flexion. Its primary role is to prevent abnormal (varus) lateral motion when the knee is extended.

### **Patellar Ligament**

The patellar ligament or tendon is a connective tissue structure that connects the patella to the tibia. Pope (1994) classified this structure as either a ligament or a tendon on the characterization of the patella, as either a sesamoid bone or as a separate bone. Normally, the patellar ligament is the same length as the patella.

### **The Cruciate Ligaments**

The cruciate ligaments are intracapsular and extrasynovial structures that traverse the knee joint, attaching to the tibia and the femur (Tortora and Grabowski, 1996). The cruciates were originally named due to their crossed arrangement (Goldblatt and Richmond, 2003). The posterior cruciate ligament (PCL) originates on the anterior aspect of the medial surface of the medial condyle of the femur and courses in a lateral, posterior, and inferior direction to insert onto a depression located between the medial and lateral tibial plateaus. The primary function of the PCL is to resist posterior translation of the tibia on the femur and to act as a secondary stabilizer to varus-valgus

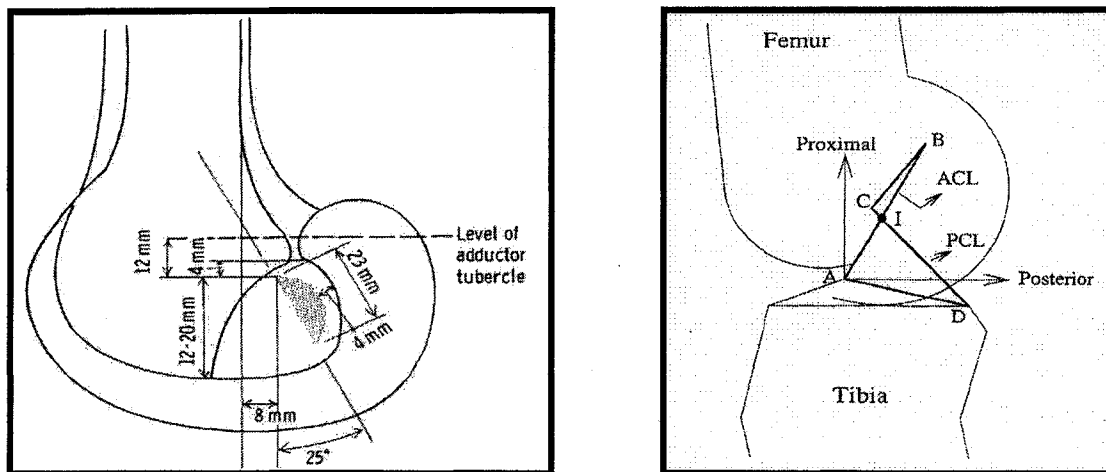
movement and external rotation (Girgis *et al.*, 1975). According to Gollehon *et al.* (1987), the PCL is the only ligament in isolation that provides primary restraint to the posterior translation at all angles of flexion. The PCL provides no resistance to anterior drawer (Butler *et al.*, 1980), but does provide a check against extreme hyperextension only after the ACL has been severed (Girgis *et al.*, 1975).

The anterior cruciate ligament originates on the posterior aspect of the medial aspect of the lateral femoral condyle and runs medially, anteriorly and inferiorly to insert onto an area anterior to the intercondylar eminence of the tibia (Tortora and Grabowski, 1996). The ACL functions primarily to resist anterior movement of the tibia on the femur and control internal rotation, but also acts in conjunction with the PCL to resist varus-valgus movement.

Normal functioning of the knee is very dependent on the anatomy of the ACL, specifically, the position, shape and dimensions of its areas of attachment (Fuss, 1989). Its spatial orientation within the joint can be directly related to its function as a constraint of joint motion. Dienst *et al.* (2002) precisely defined the location of the origin of the ACL as slightly inferior to the most superoposterior quadrant of the intercondylar fossa. The orientation and shape of this area has been described, by several authors, as a segment/arc of a circle or an ellipse (Arnockzky, 1983; Ellison and Berg, 1985; Fuss, 1989; Girgis *et al.*, 1975; Goldblatt and Richmond, 2003; Moore, 1992), and may be seen in Figure 2A.

The ACL consist of fibres of varying lengths that attach to the tibia and femur, not as a singular unit, but as a collection of individual fascicles that spread out over a broad area (Fuss, 1989). The ACL has been described by numerous investigators as

having 2 or 3 distinct divisions called fascicles, while as other anatomical studies have found the ACL to contain no distinct divisions (Butler *et al.*, 1980; Odensten and Gillquist, 1985). The two fascicular bundles have been described as the anterior medial band (AMB) and the posterolateral band (PLB) (Arnockzky, 1983; Dienst *et al.*, 2002; Ellison and Berg, 1985; Furman *et al.*, 1976; Girgis *et al.*, 1975; Smith *et al.*, 1993), in reference to their relative tibial attachments. An additional fascicular bundle called the intermediate band has been identified by the work of Amis and Dawkins (1991) and is described by Kapandji (1987). Unfortunately, no consensus exists regarding the fascicular anatomy of the ACL. However, a two bundle description has been accepted for understanding the function of the ACL (Dienst *et al.*, 2002) and will be utilized throughout the remainder of this document.



**Figure 2.** A) Relative orientation and position of the ACL on the medial surface of the right lateral femoral condyle (Girgis *et al.*, 1975). B) Schematic representation of the four-bar linkage. The instant center of rotation is defined at the intersection of AB and CD (Imran and O'Connor, 1998).

According to Dienst *et al.* (2002), the fascicles of the anteromedial band originate from the most posterior and proximal aspect of the femoral attachment and insert at the anteromedial aspect of the tibial attachment. In contrast, the fascicles of the posterolateral

band originate at the distal aspect of the femoral attachment and insert at the posterolateral aspect of the tibial attachment. The PLB represents the largest portion of the ligament, with a smaller number of fascicles making up the AMB (Smith *et al.*, 1993). As well, Girgis *et al.* (1975) found the AMB to be relatively shorter compared to the PLB. The relative position of the AMB and the PLB attachment sites is critical in their respective functions during flexion and extension of the knee joint. Several investigators have reported a functional difference between these two bundles, noting that the AMB is tight in flexion, whereas the PLB is tight in extension (Arnockzky, 1983; Dienst *et al.*, 2002; Ellison and Berg, 1985; Furman *et al.*, 1976; Girgis *et al.*, 1975; Smith *et al.*, 1993). The functional significance of the relative attachment of the two bundles during knee flexion is very well described by Girgis *et al.* (1975).

Since the femur flexes 120° - 130°, the vertical attachment of the anterior cruciate ligament becomes horizontal. This change brings the bulk of the attachment of the anterior cruciate ligament closer so that it becomes loose. Only the fascicles from the maximum convexity of the femoral attachment which form a thin anterior medial band become taut. The reason that these fascicles become taut during flexion is that the maximum convexity of the femoral attachment moves inferiorly and posteriorly rather than anteriorly as does the bulk of the anterior cruciate, and thus this part of the ligament becomes taut.

Smith *et al.* (1993) described this functional relationship as a continuum, with different fascicles being taught at different angles of knee flexion as the knee moves throughout its full range of motion. The reciprocal relationship of the anteromedial and the posterolateral bundles, is described by Ellison and Berg (1985), “as a four bar linkage within the anatomy of this single ligament and provides stability throughout the entire arc of knee joint motion”.

Isometric fibres of both the posterior and anterior cruciate ligaments act together as guiding bundles to control the rolling and gliding motion of the tibiofemoral joint (Fuss, 1989) and prevent unphysiologic or excessive motion (Ellison and Berg, 1985). The kinematics of the tibiofemoral joint are uniquely complex and are a result of the interaction of the shape of the femoral condyles and the guiding and restraining motion provided by the cruciate ligaments. Ellison and Berg (1985) stated that “kinematic guidance is therefore determined by the relative tension exhibited in the various collagenous fascicles of a given ligament”. During flexion of the knee, the ratio of rolling to gliding movement changes and as result alters the position of the axis of rotation of the knee in the sagittal plane (Ellison and Berg, 1985). Muller (1983) revealed that the variable axis of rotation of the knee can be simulated from crossed four bar link, where as Fuss (1989) described the four bars as the femur, tibia and the two guiding bundles of each respective cruciate ligament (Figure 2B). The distal attachments of the ACL and PCL are fixed and the distance between the two ends does not change in flexion or extension. According to Dienst *et al.*(2002), the gliding intersection of the crossed bars (i.e. the cruciate ligaments) represents the instant center of rotation of the knee joint. The relationship of the cruciate ligaments is nearly perpendicular when comparing the tibial attachment of the ACL to the line of attachment of the PCL on the posterior aspect of the tibia. In the sagittal plane both ligaments are separate from each other and also cross each other in the frontal plane.

#### *2.1.1.3 Accessory Structures*

The articular capsule of the knee is a fibrous sleeve that covers the distal end of the femur and the proximal end of the tibia. The posterior invagination of the capsule

along with the medial and lateral menisci partitions the synovial cavity into almost separate and distinct medial and lateral halves. According to Tortora and Grabowski (1996), the capsule is not complete or independent but is continuous with the muscle tendons and their expansions and is closely associated with the cruciate ligaments. The capsule consists of two layers: an outer layer is made up of dense, irregular connective tissue and an inner layer called the synovial membrane. This membrane secretes synovial fluid, which lubricates and reduces friction in the knee joint (Tortora and Grabowski, 1996). The articular disc or menisci of the knee are pads of fibrocartilage interposed between the joint surfaces of the femur and tibia. The relative size and thickness of lateral meniscus is greater than that of its medial counterpart. This is due to the greater lack of congruency of articular surfaces between the lateral femoral and tibial femoral condyles compared to the medial side. The lateral joint space is an average of 2.95 mm larger than its medial counterpart. As described earlier, the medial meniscus is relatively immobile compared to its lateral counterpart, as it is bound to the medial tibial plateau (Mackenzie and Dixon, 2001) and has fibres blended with the MCL and joint capsule. This relative immobility of the medial meniscus makes it more liable for damage during various forms of trauma (Mackenzie and Dixon, 2001). The lateral meniscus is required to be relatively mobile as it moves with the lateral femoral condyle during the screw home mechanism as the knee approaches full extension. The menisci function to correct the lack of congruency between the tibiofemoral joint surfaces and maintain joint stability. Additional functions of the menisci include: transmission of compressive loads, shock absorption, stress reduction, joint lubrication, and nutrient distribution (Allen *et al.*, 2000).

### 2.1.2 Neuroanatomy and Mechanoreceptors of the ACL and other joint structures

The ACL, typically, is thought of as a purely mechanical structure. However, its role in knee joint proprioception, kinaesthesia and recruitment of agonistic muscles during knee destabilizing activities has been investigated by a plethora of researchers. Early histological studies (Boyd, 1954; Freeman and Wyke, 1967; Halata, 1977) of various knee joint structures, primarily on cats, found the presence of several types of mechanoreceptors in the ACL and other surrounding structures such as the joint capsule, PCL, collateral ligaments and the menisci. Ruffini and pacinian corpuscles, golgi-tendon like organs, and free nerve endings have been reported in these structures. Later histological studies on humans (Haus and Halata, 1990; Schultz *et al.*, 1984; Schutte *et al.*, 1987; Zimny *et al.*, 1986; Zimny and Wink, 1991) confirmed the sensory and proprioceptive potential of the ACL and other connective tissues of the knee, reporting the presence of these same mechanoreceptors, as in cat preparations. Understanding the proprioceptive and reflexive potential of the ACL and the various other joint structures is of a paramount importance to gaining insight into the mechanisms of mechanical joint stability. For example, injury to the ACL may disrupt its afferent feedback potential, which will require neural adaptation and increased reliance on the mechanoreceptors in the remaining intact structures. Ultimately, this will affect the level of muscular recruitment and the stiffness of the muscles supporting the knee, directly influencing the joint's rotational stability.

Mechanoreceptors play a fundamental role in the organization of the central nervous system, functioning as transducers, converting physical energy expressed as tension into a nervous signal (Zimny *et al.*, 1986). Their overall function has been



described by Kennedy *et al.* (1982) as a structure that has the ability “to initiate a reflex which protects the joint by muscular splinting in situations of abnormal stress”. The 3 mechanoreceptors described thus far, each adapt to mechanical stimuli at different thresholds and are categorized as either slow or rapidly adapting. Functionally, Ruffini corpuscles and Golgi tendon organs are slowly adapting, but the Ruffini corpuscles have a low threshold while the GTO’s have a high threshold to physical strain (Freeman and Wyke, 1967; Zimny *et al.*, 1986).

These 3 mechanoreceptors when stimulated provide afferent feedback to divergent neurons and interneurons in the spinal cord to regulate the control of movement along with input from supraspinal centers. The afferent signals of these mechanoreceptors along with feedback from muscle spindles and musculotendinous GTO’s provide input to the alpha and gamma motor neuron pathways to regulate muscle stiffness. The alpha motor neurons innervate extrafusal muscle fibres, while the gamma motor neurons supply the smaller diameter intrafusal muscle fibres. Activation of gamma motor neurons results in shortening of the intrafusal fibres and a concomitant elevation of muscle spindle sensitivity. Kandel *et al.* (2000) describe the synergistic activation of the gamma motor neurons in a nearly parallel fashion with the alpha motor neurons as *alpha-gamma coactivation*. In cat preparations, spindle responsiveness when muscles are stretched are preset at a fairly steady level, but vary according to the specific task or context (i.e. is a perturbation expected or unexpected??; is the task complexity high or low??) Kandel *et al.* (2000). Significant controversy exists regarding the exact mechanisms in which afferent feedback from ligament and joint mechanoreceptors influence the skeletomuscular and gamma muscle spindle systems. Regardless of the exact mechanism

of reflexive (i.e. monosynaptic or polysynaptic input of joint receptors) muscular activation, strain or electrical stimulation to the ACL results in the antagonistic activation of the hamstrings and the inhibition of the quadriceps (Solomonow *et al.*, 1987; Tsuda *et al.*, 2001). Injury to the ACL may dramatically alter the level of muscle spindle activity, which ultimately affects knee joint stability by augmenting muscle stiffness, which serves as an integral role in rotational joint stability. This neuromuscular control mechanism acts in conjunction with the ACL's passive stabilizing role, as the joint approaches the end range of knee extension.

### 2.1.3 Muscles of the Knee

Activation of the muscles that cross the knee produces movement at the joint and function synergistically along with ligamentous restraints to stabilize the knee, throughout its full range of motion. Therefore, it is necessary to describe in detail the muscles examined in this thesis. The following section has been adapted from two primary sources (Tortora and Grabowski, 1996; Moore, 1992). Additional resources, when required to add detail or highlight supplementary information not provided by the above two sources, will be cited accordingly.

#### *2.1.3.1 Anterior (Extensor) Compartment*

The anterior compartment of the thigh is comprised of the quadriceps femoris, the sartorius, and the tensor fasciae latae (Figure 3A).

#### **Quadriceps Femoris**

The quadriceps femoris (QF) is a composite muscle comprised of four distinct muscle groups: the rectus femoris, the vastus lateralis, the vastus medialis, and the vastus

intermedius. The QF heads have a common insertion onto the patella and their overall function is to extend the leg on the thigh or extend the knee joint.

The rectus femoris (RF) originates on the anterior inferior iliac spine (AIIS) and also functions as a hip flexor. Its fibres course straight down the thigh. The vastus lateralis (VL) originates on the greater trochanter and lateral lip of the linea aspera of the femur. Its fibres are obliquely oriented in a lateral to medial direction. The vastus medialis (VM) originates on the intertrochanteric line and medial lip of the linea aspera of the femur. Its fibres are also obliquely oriented, but in medial to lateral direction. The vastus intermedius (VI) originates on the anterior and lateral surfaces of the body of the femur. It lies between VL and VM and is deep to RF. The VI and RF are both bipennate muscles and each of their respective tendons run straight down.

### **Sartorius**

The sartorius (SA) attaches proximally onto the anterior superior iliac spine (ASIS) and the superior part of the notch inferior to it. It inserts onto the superior aspect of the medial surface of the tibia. The SA is responsible for flexing the lower leg on the thigh (knee flexion) and additionally acts at the hip to flex and laterally rotate the thigh. Its fibres are oriented in an oblique fashion, running in a lateral to medial direction. The SA is most superficial muscle in the anterior compartment.

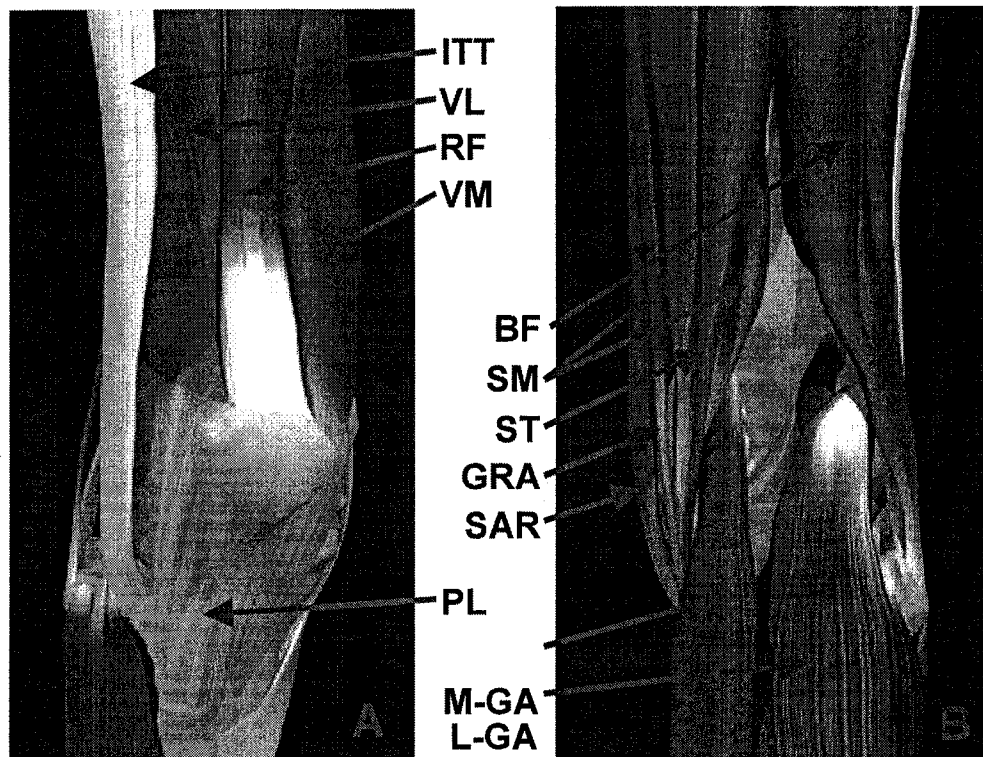
### **Tensor Fasciae Latae**

The tensor fasciae latae (TFL) originates on the ASIS and anterior aspect of the external lip of the iliac crest. Its fibres attach distally by way of the iliotibial tract (ITT) to the lateral condyle of the tibia. It acts across two joints. Contraction of the TFL enables the QF muscles to act more powerfully. Additionally, it tightens the ITT and allows the

gluteus maximus muscle to maintain the knee in an extended position. At the hip, the TFL is responsible for abduction, medial rotation, and flexion of the thigh.

### 2.1.3.2 Posterior (Flexor) Compartment

The posterior compartment, the hamstrings, is a collective name for three distinct muscles: the biceps femoris, the semitendinosus, and the semimembranosus (Figure 3B). Overall the hamstrings act to flex the knee and extend the hip. All of these muscles have a common origin on the ischial tuberosity, with the exception of the short head of the biceps femoris.



**Figure 3.** A) Anterior view of knee musculature: ITT- iliotibial tract, VL- vastus lateralis, RF- rectus femoris, VM- vastus medialis, PL- patellar ligament. B) Posterior view: BF- biceps femoris, SM- semimembranosus, ST- semitendinosus, GRA- gracilis, SAR- sartorius, M-GA- medial gastrocnemius, L-GA- lateral gastrocnemius

The biceps femoris (BF) is composed of two divisions, a short and long head that have a common insertion onto the head of the fibula and the lateral condyle of the tibia. The biceps femoris short head (BF-S) originates on the lateral lip of the linea aspera, while the biceps femoris long head (BF-L) originates on the ischial tuberosity. The BF-S only acts at the knee, flexing the leg on the thigh. Both BF muscle groups are oriented in a medial to lateral direction. Due to its orientation, the BF also is responsible for laterally rotating the leg.

The semitendinosus (ST) inserts on the medial surface of the superior aspect of the tibia. Its fibres course in a slight lateral to medial direction and account for its ability to medially rotate the knee when both the leg and thigh are flexed.

The semimembranosus (SM) inserts onto the posterior aspect of the medial condyle of the tibia. Its fibres course in a lateral to medial direction, similar to the ST, and also functions to medially rotate the leg on the thigh.

### **Gracilis**

The gracilis (GR) is part of the medial (adductor) compartment of the thigh and originates on the pubic symphysis and pubic arch. It inserts to the medial surface of the body of the tibia. Its fibres run along the medial side of the thigh and knee and it functions as a knee flexor, medial rotator and as a hip adductor.

### **Gastrocnemius**

The gastrocnemius (GA) is the most superficial of all the posterior compartmental muscles of the lower leg. It consists of two heads, crosses two joints, and has a common insertion via the Achilles tendon onto the posterior surface of the calcaneus. The gastrocnemius medial head (GA-M) originates on the popliteal surface of the femur,

superior to the medial condyle. The gastrocnemius lateral head (GA-L) attaches proximally onto the lateral aspect of the lateral condyle of the femur. Collectively, both heads function to plantarflex the ankle and flex the knee.

## **2.2 NORMAL KNEE BIOMECHANICS**

### **2.2.1 Passive Knee Kinematics**

The intricate three dimensional movement of the tibiofemoral joint is a result of complex interactions between the changing orientation of the cruciate ligaments and the changing radii of curvature of the femoral condyles that simultaneously occur during flexion and extension motions. Understanding the joint kinematics of the knee is further compounded when the motion of the patellofemoral joint is considered.

#### ***2.2.1.1 Tibiofemoral Kinematics***

The role of the cruciate ligaments and the geometry of the articular surfaces of the tibiofemoral joint in determining its movements are inseparable and do not act in isolation (Jakob and Staubli, 1990). “The shapes of the articular surfaces must fulfill the requirement that they move in contact with one another while maintaining the neutral fibers of the cruciate ligaments at constant length (Daniel *et al.*, 1990).

The cruciate ligaments assist in guiding the knee through 6 potential degrees of freedom: 3° of translation (anterior-posterior, medial-lateral, and proximal-distal) and 3° of rotation (flexion-extension, external-internal rotation, and abduction). According to Butler *et al.* (1980), the ACL provides an average of 86% of the total resisting force to anterior drawer test *in vitro* knee specimens, while other supporting structures (all other

ligaments and capsular structures) provide the remaining secondary restraint of approximately 3 % each. The geometry of the articular surfaces, along with the guiding action of the cruciates, results in obligatory axial rotation, gliding and rolling movements of the femur on the tibia during knee flexion and extension.

The radius of curvature of the femoral condyles is not uniform, and varies, according to its elliptical shape (Kapandji, 1987). The circumferential length of the femoral condyles are two times greater than that of the tibial condyles and, therefore, a simultaneous forward gliding and backward rolling motion of the femur is necessary to maintain contact of the articular surfaces and satisfy the constraints of the posteriorly directed changing axis of rotation of the cruciate ligaments intersection that occurs during knee flexion about the sagittal plane. As a result, the distance between successive contact points on the femur is 3 to 4 times greater than the corresponding distance on the tibia (Daniel *et al.*, 1990). This point of contact moves backward during flexion and prevents impingement of the femur on the posterior surface of the tibial condyle (Kapandji, 1987). Knee flexion occurs about the instantaneous center of rotation (zero velocity) or geometric center of the knee and is a function of the changing shape of the cruciate ligaments. Obligatory internal femoral rotation (screw home mechanism during knee extension of approximately 20° to 25° (Daniel *et al.*, 1990) occurs as function of the inequality of the lateral femoral condyle circumference relative to its medial counterpart. The ACL during various activities acts primarily to restrain anterior translation and internal rotation of the tibia.

### *2.2.1.2 Patellofemoral Kinematics*

Movement of the patella on the femur occurs primarily in the sagittal plane along a path constrained by the relatively inextensible patellar ligament, the line of action of the QF and contact with the femur. Functionally, the patella acts to increase the lever arm of the quadriceps femoris. This lever arm changes length as the patella moves along a predetermined path during flexion-extension motion of the tibiofemoral joint. As the knee approaches full flexion the patella moves into the intercondylar groove closer to the axis of rotation of the tibiofemoral joint, resulting in a smaller effective moment arm. The maximum effective lever arm distance is attained at approximately  $45^\circ$  as the knee is extended and decreases thereafter (Lehmkuhl and Smith, 1983).

Functionally, the patellofemoral joint is kinematically modelled as a two dimensional mechanism acting in the sagittal plane and consist of several components: 1) the attachments of the patellar ligament on the tibial tuberosity and inferior aspect of the patella, 2) the patellar ligament, 3) the patella, and 4) the line of action of the QF. The relative movement of these elements is dependent on the movement of an assumed single point of contact that occurs between the articular surfaces of the femur, which may be described given the orientation of the patellar axis. This axis is defined as, the intersection of the inferior attachment of the patellar ligament on a circumscribed radius, determined by the overall length of the patellar ligament and the relative position of its tibial insertion. Again, as with the QF lever arm length, the relative position of most of these components is contingent on the angle of flexion or extension of the tibiofemoral joint.



The orientation of the patellar ligament relative to the tibial axis is directed anteriorly during the first 70° of knee flexion and posteriorly between 70° and 120°. This alteration in orientation of the patellar ligament changes the QF from an ACL antagonist during the first 70° or 80 of knee flexion to an ACL agonist over the remaining flexion range of motion.

The patellofemoral (PF) contact point varies at different flexion-extension angles. In full extension, the contact point is located on the inferior aspect of the patella (Baratta *et al.*, 1988) and makes a rolling-gliding motion along the patellar surface of the femoral condyle (van Eijden *et al.*, 1986). This contact point moves upward along the patellar profile from full extension to 90° and then makes a small reversal between 90° and 120°. The contact point on the femoral condyle moves inferiorly throughout the entire flexion range. The average amount of patellar gliding is approximately 0.65 mm/deg of TF between 0° and 80° and 0.45 mm/deg between 80° and 120°.

Also noteworthy is the smaller angular rotation of the patella compared to the tibia. The patella rotates 0.7°/deg of lower leg flexion due to the backward rotation of the patellar ligament relative to the tibial axis. No angular changes between the patellar ligament and patella have been measured (van Eijden *et al.*, 1986).

### 2.2.2 Muscular Contributions to Knee Kinematics

The ability of the muscles to stabilize the knee joint is dependent on several factors, such as: 1) relative cross-sectional area of the muscle, 2) alteration of insertion sites as a function of variation of the TF joint instant center of rotation during flexion-extension, 3) variation of muscle moment arm as a function of joint angle, and 4) force length properties of muscles. The larger relative cross sectional area of the quadriceps,

hamstrings and gastrocnemius allow them to significantly contribute to joint stability, compared to smaller muscles like the sartorius. The lines of actions of these muscles vary systematically, along with ligament and contact forces during flexion and extension, as the tendons and ligaments rotate about their points of insertion. Baratta *et al.* (1988) defined, muscle moment arm, as the perpendicular distance from the muscle's line of action and the instant center of rotation of the joint. They suggest that both the variable line of action of a muscle, and variation in the instant center of TF rotation, contribute toward a certain pattern of muscle moment arm as a function of joint angle. This variable pattern has been described for the patellofemoral extensor mechanism. Typically, the tension generated in the QF is presumed to be of equal magnitude to that of the patellar ligament.

#### *2.2.2.1 Hamstrings*

The hamstrings, an ACL protagonist, has a posteriorly directed shear component that is only effective at resisting anterior tibial translation after approximately 10 degrees of knee flexion has been attained. From full extension to 10° of flexion, the hamstrings are oriented almost parallel to the tibia and are incapable of providing an effective restraint to anterior tibial drawer. As knee flexion increases beyond 10°, this backward shear component increases due to the increasing angle of pull of the hamstrings on the tibia.

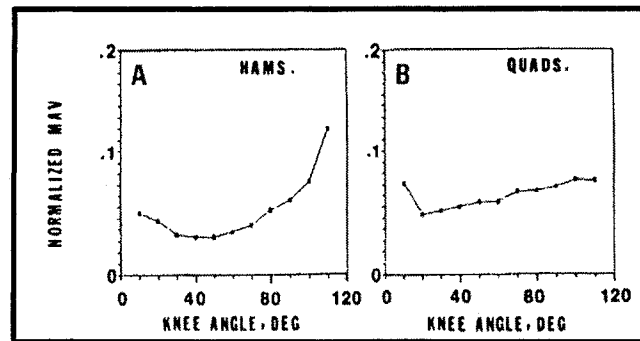
#### *2.2.2.2 Gastrocnemius*

The line of action of the gastrocnemius produces an anteriorly directed shear force throughout the entire knee range of motion, acting as an ACL antagonist like the QF.

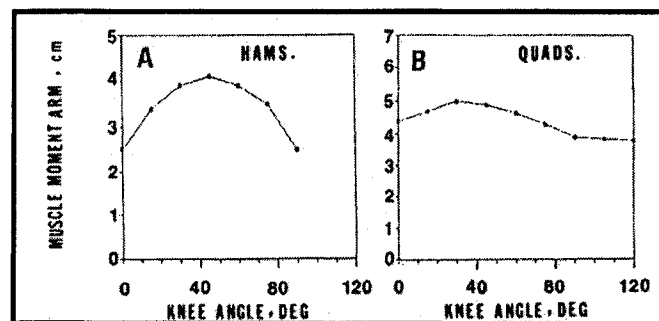
### 2.2.3 Neuromuscular Recruitment and Reflexes

Various activities of daily living and sport, like walking, climbing stairs, running, and cutting or cross-over manoeuvres, elicit different patterns and levels of activation of knee joint musculature. The level of muscle activation is dependent of several factors: 1) type of contraction, 2) velocity and force of contraction 3) position, orientation, and magnitude of an external perturbation relative to the knee center, 4) moment arm length variation as a function of joint angle, and 5) training effects (untrained vs. trained).

Baratta *et al.* (1988) demonstrated that hamstring (QF antagonist) level of activity is nearly inversely related to its moment arm over the joint range of motion (Figure 4 and Figure 5). They postulated that the muscle compensates for its moment arm variability while attempting to maintain constant torque. The hamstrings and quadriceps exerts a nearly constant opposing torque over the joint range of motion.. This evidence demonstrates the ability of the neuromuscular system to organize itself to balance muscular forces acting at the knee, thus effectively minimizing shear forces and evenly distributing compressive loads between the TF articular surfaces. If agonist (extensor) activity is considered independent of co-contraction of the hamstrings, then a single point along the anterior aspect of its articular surface will be selected as a center about which the torques develop (Figure 6). “This configuration will create a focused stress point that will rapidly contribute toward focal deterioration along the articular surface and result in early tissue damage and osteoarthritis” (Baratta *et al.*, 1988).



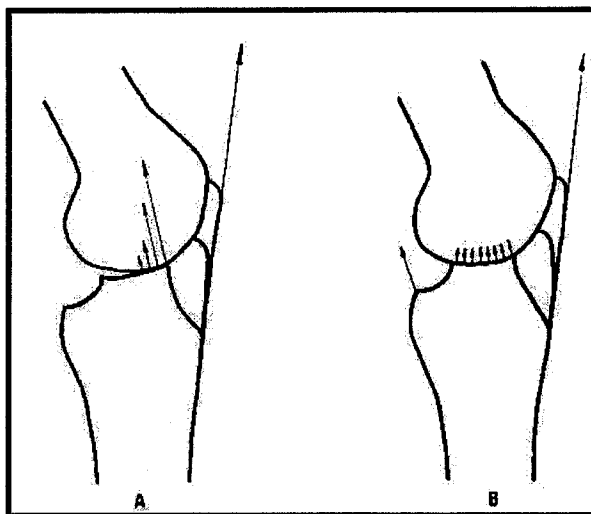
**Figure 4.** Average normalized antagonistic mean amplitude value versus joint angle for hamstrings and quadriceps (Baratta *et al.*, 1988).



**Figure 5.** Muscle moment arm variations with joint angle. Note the striking inverse pattern of muscle moment arm and MAV between the hamstrings (4A & 5A) and quadriceps (4B & 5B) (Baratta *et al.*, 1988).

They suggest that antagonistic (hamstrings) activity is the most critical factor facilitating safe pressure distribution over articular surface interface throughout the joint range of motion. The recruitment of the quadriceps and hamstrings occurs in an almost synchronous fashion and serves to enhance stability of the joint provided by the knee ligaments. As alluded to earlier, the mechanoreceptors of the ACL, joint capsule and other knee ligaments function in an elaborate neuropathway that coordinates the activities of the quadriceps and hamstrings muscles. MacWilliams *et al.* (1999) discovered that co-contraction of the hamstrings during physiological loaded flexion, in cadaveric knee specimens, reversed the anterior/posterior shear force, increased joint compression force

and increased the quadriceps force required for equilibrium. Thus, the drawback to hamstring co-activation is an increase in the physiological cost required to maintain the knee in a flexed position.



**Figure 6.** A) Localized focal high pressure to the anterior aspect of the articular surface due to the absence of hamstrings antagonistic activity. B) Evenly distributed low articular pressure due to fully activated hamstrings co-contraction. (Baratta *et al.*, 1988)

Rotational equilibrium, where the knee joint will not flex or extend, is achieved when extensor and flexor muscle forces are applied simultaneously and their respective net moments are equal in magnitude. However, the shear components of the flexors and extensors that act parallel to the tibial plateau are not necessarily equal and opposite throughout all joint angles. Therefore, co-activation of the hamstrings and quadriceps will, in most circumstances, load one or more of the cruciate ligaments. O'Connor (1993) identified a critical flexion angle of  $22^\circ$ , where the parallel muscle components exactly balance each other and no ligament forces are required. The magnitude of the posteriorly directed shear component produced by the hamstrings is dependent on its line of action which is altered with changes in joint angle. Hirokawa *et al.* (1991), in a in-vitro study of

cadaveric knees, demonstrated that hamstrings were ineffective at reducing the anterior shift of the tibia in the range of 0° to 15° of flexion. Between 15° and 80°, this posterior shear force was capable of reducing the anterior tibial displacement created by the quadriceps. After 80° of flexion, the shear component generated by the quadriceps reverses direction and produces a backward shift of the tibia along with the hamstrings.

## **2.3 ACL INJURY**

Injury to the ACL severely compromises the stability of the knee and represents the most permanent disability of any athletic injury, with the longest duration of recovery (Thacker *et al.*, 2003).

### **2.3.1 Diagnosis and Classification**

Acute ACL injuries can be graded or classified based on the degree of damage to the ACL (partial or complete disruption) and the presence or absence of damage to other knee structures (isolated or combined) (Evans and Stannish, 2001). ACL injuries are graded on a 4 point scale: grade 0 = intact ligament, grade 1 = partial tear with less than half of ligament disrupted, grade 2 = partial tear with more than half of ligament disrupted, grade 3 = complete tear (Hong *et al.*, 2003). According to Evans and Stannish (2001), 85% of ACL injuries are complete ruptures and isolated injury occurs in 25% of all injury cases. For example, meniscal tears are associated with 41% to 68 % of acute ACL injuries (Moore, 2002).

Diagnosis of an acute ACL injury may be achieved through several methods, including: history of injury, physical examination, arthroscopy, KT-1000 arthrometer, radiographic evaluation and MRI.

Physical examination involves a subjective comparison of the end feel of the joint between the injured and uninjured leg, with the knee positioned at a predetermined knee angle. During the Lachman and anterior drawer test the clinician manipulates the tibia in an anterior direction to evaluate the integrity of the ACL at knee angles of 15°-30° and 90°, respectively. The pivot shift test is performed with the internal rotation of the leg and a valgus directed force to a slightly flexed knee. The anterior drawer test is specific for AMB disruption, while the Lachman and pivot shift test are more specific for determining PLB integrity. The sensitivity and specificity of each of these test ranges from 41-82% and 95-98%, respectively (Katz and Fingerhuth, 1986). A positive Lachman or anterior drawer test with a firm end-feel indicates a partial (grade 2) ACL sprain, particularly when the pivot shift test is negative. Due to the qualitative and subjective nature of these tests, confirmation of the diagnosis is acquired through more quantitative test or diagnostic tools, like MRI or arthroscopy.

MRI, considered the “gold standard”, is 95% accurate in diagnosing ACL and concomitant injuries, such as meniscal damage (Goldstein and Bosco, 2001; Moore, 2002). KT arthrometer is a quantitative, instrumented test that measures the anterior tibial displacement by a determined 6.82 kg (15 lb), 9.09 kg (20 lb), and 13.64 kg (30 lb) pull on the proximal tibia, a maximal anterior pull by the examiner and an active quadriceps contraction (Larson and Grana, 1993). A complete ACL disruption is indicated by a side-to-side difference of 5 mm or greater (Frank and Jackson, 1997). Arthroscopy is used

systematically to visually assess and diagnose injury to the ACL and other knee joint structures.

### 2.3.2 Treatment and Prognosis

Proper management of acute injury to the ACL requires very careful consideration of a plethora of patient and injury variables, including: patient age, occupation, activity/sports participation level (sedentary vs. elite athlete; low vs. high risk pivoting sports), combined vs. isolated injury, patient reported “giving way”, functional instability, degree of injury (complete vs. partial rupture), possibility of failure with conservative treatment, risk of osteoarthritis development, and patient expectations and treatment preference. The decision to proceed with surgical reconstruction of the ACL or non-operative (conservative) treatment requires thoughtful evaluation of all these factors and is somewhat influenced by the bias and experience of the orthopaedic surgeon treating the injury. Significant controversy exists among orthopaedic surgeons, who treat ACL injuries, regarding the indications to perform surgery and its long-term benefits over conservative treatment. A recent survey on the treatment of ACL injury of members of the American Academy of Orthopaedic Surgeons (AAOS), revealed disagreement on 4 patient characteristics (age over 40, presence of pain, irreparable meniscal damage, and injury involving Workers’ Compensation) on the decision to perform surgery (Marx *et al.*, 2003). According to Marx *et al.* (2003), the fundamental rationale for surgical reconstruction of a disrupted ACL is to prevent an unfavourable cascade of progressive knee instability, which leads to recurrent intra-articular damage and eventual osteoarthritis. Jackson *et al.* (1993) proposed the “ACL injury cascade” as a conceptual framework for understanding the events that may follow (Figure 11). Although, two



recent studies (Loosli and Oshimo, 1994; Fink *et al.*, 2001) failed to show any differences in radiographic changes between ACL injured recreationally athletic patients treated surgically or by non-operative means in follow-up assessments performed 5-7 years or 10-13 years following injury. Participation in high-risk pivoting sports, like soccer and basketball, was shown to have a significant correlation with osteoarthritic changes seen in the non-operative group (Fink *et al.*, 2001). Prevention of this progressive sequale of meniscal damage and articular deterioration is dependent on restoration of the knee stability.

The development of symptomatic knee instability is unpredictable with instability rates ranging from 16 to almost 100% and is more likely to occur with combined damage to the menisci, articular cartilage, or other ligaments (Evans and Stannsih, 2001). Knee instability is defined as “giving way” during activity (Lysholm and Gillquist, 1982). Some patients are disabled for sports, while others have insignificant impairment. These discrepancies may be a result of varying degrees of ACL damage, different combinations of injury, and may represent the diverse physical demands of different demographic populations.

Previous studies have attempted to classify an ACL injured patients' level of dysfunction by the presence or absence of ACL rupture (Eastlack *et al.*, 1999). However, ACL rupture does not always lead to instability and functional impairment. In fact, laxity measures between two functionally distinct ACL injury patient groups, compensators (copers) and noncompensators (non-copers), has shown no correlation to functional outcome measures for the knee (Eastlack *et al.*, 1999; Snyder-Mackler *et al.*, 1997). Both patient populations, in numerous studies, have been shown to have similar side to side

arthrometer displacements, but score significantly different in terms of overall knee function. In 2001, Johnson and Smith (2001) identified the Lysholm (I and II) knee scoring scale, the Tegner activity score, the International Knee Documentation Committee (IKDC) subjective knee evaluation form, Mohtadi's ACL quality of life outcome measure, and the Knee Injury and Osteoarthritis form, among 54 knee outcome measures, as having been appropriately validated prior to their use. The Tegner activity score and Lysholm II knee scoring scale are considered the "gold standard" for clinical follow-up of ACL injured patients. In a two part study, (Noyes *et al.*, 1983a; 1983b), patients' were evaluated based on the symptoms experienced during functional activities and through question items. Numerous studies to date (Fitzgerald *et al.*, 2000; Rudolph *et al.*, 2000) have included and advocate the use of functional performance outcome measures, like the single leg hop, cross-over hop, triple hop, and timed hop test described by Noyes *et al.* (1991) to distinguish copers from non-copers.

In 1983, Noyes *et al.*, found that, out of 84 patients with chronic ACL laxity treated by rehabilitation and activity modification, one-third of the patients improved, one-third stayed the same, and one-third became worse and required reconstructive surgery. In a later study, by Snyder-Mackler *et al.* (1997), 10 of 20 subjects with confirmed (MRI and arthroscopy) ACL rupture, were classified as compensators (copers) as they were able to make a full return to sports participation. The remaining 10 subjects, described as noncompensators or non-copers, were unable to return to sports activities due to recurrent episodes of instability and were later scheduled for surgery.

Re-establishing stability of the knee after ACL trauma is of paramount importance and may be achieved either through surgical reconstruction or aggressive

non-operative treatment consisting of physical therapy and bracing. In the AAOS study of Marx *et al.* (2003), surgeons preferred the autogenous patellar tendon reconstruction method (79.1%) over the hamstrings (semitendinosus tendon) graft method (12%). In the same survey, clinical agreement (> than 80%) was found, advocating the use of bracing and physical therapy in the non-operative treatment of an ACL-deficient knee. The term *non-operative* treatment has yet to be clarified and could mean anything from no treatment at all to a vigorous program of rehabilitation (Giove *et al.*, 1983). Recent reviews and studies on rehabilitation of the ACL-deficient knee describes non-operative management as, a progressive exercise based program including: isometric, isotonic, and isokinetic exercises, closed chain exercises, and functional exercises involving perturbation training and sport-specific agility exercises (Chmielewski *et al.*, 2002; Fitzgerald *et al.*, 2000a; Fitzgerald *et al.*, 2000b). Perturbation training and agility exercises are designed to assist the athlete in re-establishing neuromuscular control through compensation of sensory loss of the ACL by secondary structures, thus maintaining dynamic joint stability. The ultimate goal of non-operative treatment is to return patients to full participation in high-level physical activities. In reaching this objective, several patient parameters have been identified as critical components to the ACL-deficient patient's successful return to sport. Some of these parameters are: restoration of full knee ROM, quadriceps index (involved maximum voluntary contraction (MVC)/uninvolved MVC \*100) (Eastlack *et al.*, 1999), and the hamstrings: quadriceps strength ratio or (hamstrings %MVC/quadriceps %MVC) (Eastlack *et al.*, 1999; Fitzgerald *et al.*, 2000a; Giove *et al.*, 1983), where the hamstrings are at least of equal strength to the quadriceps on the involved side.

### 2.3.3 ACL-Deficient Pathomechanics

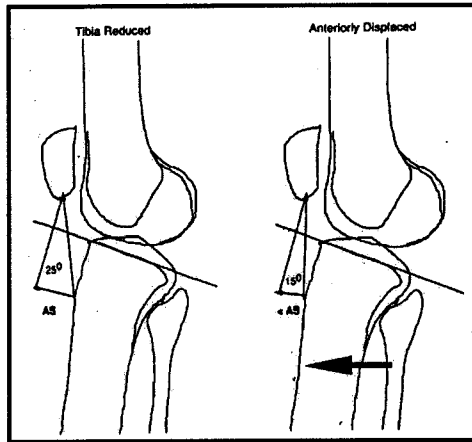
#### *2.3.3.1 Altered Kinematics*

The loss of the guiding motion and restraining action provided by the ACL following injury, results in increased anterior-posterior translation of the tibiofemoral joint and a shift in the fulcrum of axial rotation from the medial intercondylar spine of the tibia.

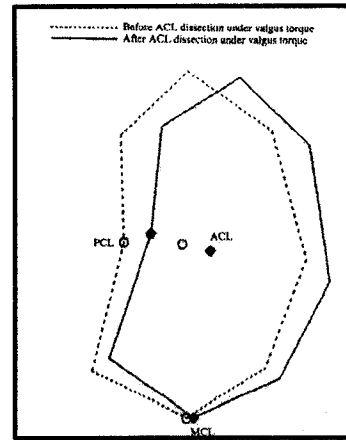
A three-dimensional motion analysis of ACL-deficient patients compared to normal subjects during level walking, by Marans *et al.* (1989), revealed an increase in anterior tibial translation, specifically during early and mid-swing phase. Additionally, Gross *et al.* (1993) found a significant increase in displacement of the anterior tibial plateau of ACL-deficient subjects involved knee compared to the uninvolved knee during isometric contraction. Anterior translation for each isometric contraction was assessed at various knee angles (15°, 45° and 75° of flexion) by measuring the anterior displacement of the tibial plateau on the resting roentgenogram relative to the isometrically resisted roentgenogram (Gross *et al.*, 1993). However, the decrease in the patellar ligament insertion angle that occurs during knee flexion was observed to decrease more in ACL-deficient knee compared to the control knee (Howell and Usafr, 1990). It seems as though the anterior drawer force produced by the patellar ligament is self-limiting in the case of the ACL deficient subject. Any increase in anterior tibial translation reduces the angle that the patellar ligament makes with the long axis of the tibia, thus effectively reducing the length of the anterior shear force lever arm (Figure 7).

In a cadaveric study investigating the rotational instability of the knee joint after selective sectioning of the ACL, the amount of internal tibial rotation was seen to increase from approximately  $10^{\circ}$  to greater than  $20^{\circ}$  within a range of knee flexion from  $5^{\circ}$  to  $25^{\circ}$  (Gross *et al.*, 1993). Additionally, the axis of this increased axial rotation was observed to shift from the medial intercondylar spine of the tibia to the medial collateral ligament (Figure 8).

Disruption to the ACL has serious implications for the mechanical stability of the knee from both a passive and dynamic standpoint. The loss of proprioceptive afferent information supplied by the mechanoreceptors of the ACL to the alpha and gamma motor systems results in several deficits and adaptations in the neuromuscular system of the knee. Some of these alterations observed in ACL deficient patients, include: increased reflex latency of the hamstrings to anterior tibial translation (Beard *et al.*, 1993; Beard *et al.*, 1994; Wojtys and Huston, 1994); decreased ipsilateral and contralateral proprioception and kinaesthetic awareness (Barrack *et al.*, 1989; Carter *et al.*, 1997; Reider *et al.*, 2003; Roberts *et al.*, 1999; Roberts *et al.*, 2000); earlier activation of hamstrings contraction during level and uphill walking on a treadmill (Kalund *et al.*, 1990); larger amplitude and duration of hamstrings contraction, increased co-contraction, quadriceps avoidance gait patterns, quadriceps inhibition, and an altered order of recruitment from hamstrings-quadriceps-gastrocnemius to quadriceps-gastrocnemius-hamstrings.



**Figure 7.** The angle of the patellar tendon acquires a more perpendicular orientation ( $25^{\circ}$  to  $15^{\circ}$ ) with quadriceps contraction in the ACL deficient knee. This effectively reduces the quadriceps anterior shear component (Howell and Usafr, 1990).



**Figure 8.** After sectioning of the ACL, the axis of tibial rotation is located about the MCL (Matsumoto and Seedhom, 1993).

### 2.3.3.2 Altered Neuromuscular and Recruitment Patterns

Beard *et al.* (1994) assessed the reflex contraction latency of the hamstrings of ACL-deficient subjects by measuring the elapsed time from initial tibial movement caused by an anteriorly directed shear force until surface electromyography recorded the onset of hamstrings activity. The contralateral uninjured limb of the ACL-deficient subjects and a control group of subjects were used to determine the effects of ACL-deficiency on the reflexive activation of the hamstrings. The ACL-deficient subjects had a mean reflex hamstring contraction latency of 90.4 ms compared to 49.1 ms for the uninjured limb, and 47.9 ms for the control limb.

Several investigators have found an inhibition mediated decrease in quadriceps activation in ACL deficient subjects during tasks, like walking or knee extension exercises (Berchuck *et al.*, 1990; Limbird *et al.*, 1988; Snyder-Mackler *et al.*, 1994; Suter *et al.*, 2001). Suter *et al.* (2001) explain that distension of the joint capsule produces a dose dependant inhibition of the

quadriceps. Previous work by Ekholm *et al.* (1960) on cat preparations revealed that increasing intra-articular pressure or pinching the anterior aspect of knee joint capsule activated joint receptors, causing an inhibition of the knee extensors and a facilitation of the knee flexors. In humans, Kennedy *et al.* (1982) demonstrated similar inhibition of the quadriceps when electrically stimulated (H-reflex), with knees infused with saline. No inhibition was produced when the joint was infused with Lidocaine, indicating the reflex potential of the knee joint capsule. The stretching of the joint capsule that occurs with quadriceps activation results in the deformation of mechanoreceptors which have been shown to inhibit quadriceps activity (Iles *et al.*, 1990). These findings suggest an association between muscle inhibition in the quadriceps and the quadriceps avoidance gait pattern described in ACL-deficient subjects.

In a recent study by Snyder-Mackler *et al.* (1994), 3 groups of ACL subjects had their quadriceps femoris strength and quadriceps inhibition assessed using a burst-superimposition technique during voluntary isometric contraction of the quadriceps. Group 1 consisted of patients with a surgically reconstructed ACL, one to six months after injury. Groups 2 and 3 consisted of non-operative ACL-injured subjects injured an average of three months (subacute) and 2 years (chronic), respectively, prior to participating in the study. The ACL reconstructed patients (group 1) and the chronic ACL-deficient patients were able to produce a maximal voluntary contraction that resulted in less than a 5% increase in force after the additional superimposed twitch was imposed. However, the subacute ACL-injured patients had an inhibition of the quadriceps of more than 5% that could not be eliminated by the additional superimposed twitch. It is evident, that the recruitment of the quadriceps and hamstrings in ACL-deficient patients is significantly influenced by several injury and treatment variables.

Several factors have been identified in the literature that lead to different neuromuscular patterns of activation of the muscles surrounding the knee in ACL-injured subjects, such as:

operative vs. non-operative treatment (Ciccotti *et al.*, 1994), chronic vs. acute or subacute (Snyder-Mackler *et al.*, 1994), functional bracing vs. no bracing, and type of rehabilitation program for non-operative treatment and whether or not this program was supervised (Ageberg *et al.*, 2001).

In non-operatively treated ACL-deficient patients, dichotomous groups, copers and non-copers, have been identified by their distinctly different neuromuscular patterns. Rudolph *et al.* (2000) recently revealed that copers move in an almost identical fashion in a single leg hopping test as do uninjured subjects and stabilize their knee with significantly greater contributions from the ankle extensors. No significant findings in activation levels of the vastus lateralis were found between the copers and uninjured subjects. In a preliminary study by Kirkendall *et al.* (2002), ACL deficient subjects performed a pre-planned side step cutting manoeuvre while running. The coper subject displayed 53% more semitendinosus activity and decreases in activity of 26% for rectus femoris and 18% for gastrocnemius in the ACL-deficient leg compared to the uninvolved leg. The results of this study supported the authors' hypothesis that copers alter their control strategy to compensate for their ACL-deficient knee. On the other hand, results from Rudolph *et al.* (2000) suggest that distinct movement patterns were not able to be recognized for noncopers. There was a tendency for these subjects to shift the distribution of their support moments from the knee to the hip. "Non-copers rely on an invariable stabilization strategy rather than stabilization strategies to suit the demands of various tasks".



## 2.4 FUNCTIONAL KNEE BRACES

“Functional bracing of the knee is a relatively recent phenomenon “ (Cawley *et al.*, 1991), and earlier brace designs have spawned significant controversy, as they often provided individual testimonials but have had very little scientific evidence to support their efficacy (Ott and Clancy, Jr., 1993). In 1984, the American Academy of Orthopaedic Surgeons formed a Sports Medicine Committee to objectively evaluate the design and effectiveness of knee braces by obtaining data from the brace manufacturers, physicians, and bioengineers (Podesta and Sherman, 1988). Knee braces have been classified into 3 or 4 categories as determined by this committee and other investigators (Paluska, 2000): prophylactic, rehabilitative, patellofemoral and functional. This thesis is concerned with the effectiveness of functional knee braces in providing stability to the ACL-deficient knee and, therefore, no further discussion of the other brace types will be undertaken. In 1985, the AAOS published a seminar report that concluded the following: controversy exist regarding the effectiveness of the knee braces, further epidemiological and biomechanical studies are required, and that more effective designs of these braces seemed imperative to improve efficacy. Unfortunately, the scientific validation of functional knee braces has not kept pace with the proliferation of knee brace companies and their innovative designs and such growth is thought to be corollary with the emergence of the *sports medicine* phenomenon.

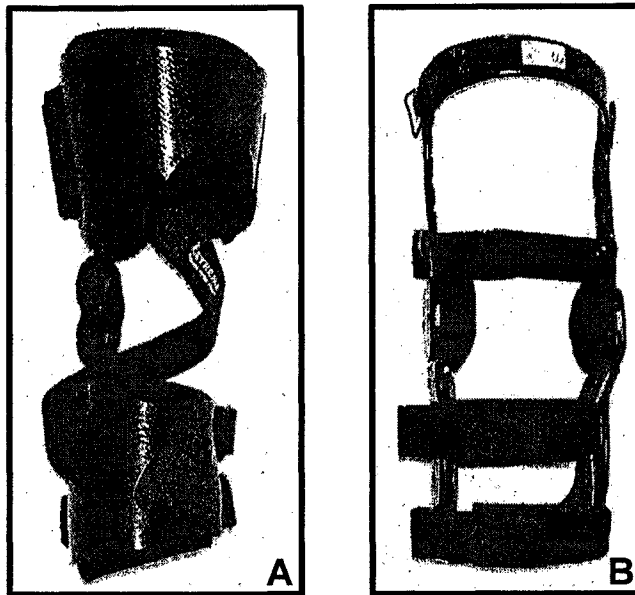
Today, there are over 30 models of marketed functional braces in both custom and off-the-shelf designs. Despite the numerous studies, over the last 20 years, that have attempted to define the efficacy of functional knee braces, significant controversy still exist regarding their mechanisms of action and stability offered to the ACL-deficient knee.

### 2.4.1 Brace Design: Types, Features and Other Considerations

Functional knee braces (FKB's) are designed to resist abnormal translation and rotation, to minimize the envelope of pathological knee motion, and to restore functional stability to the knee joint. Mechanical joints, integral components of the FKB, attempt to mimic the intricate motion of the tibiofemoral joint and act to guide and align the knee throughout its range of motion. Unfortunately, control of axial rotation (internal-external rotation) is ineffective due to the current functional brace design deficiencies imposed by the soft-tissue-brace interface. Many claims made by brace manufacturers, such as control of pathological axial rotation, are scientifically and subjectively unsubstantiated. However, most brace designs provide a small degree of resistance to anterior translation. This restraint is typically produced as a coupling effect with flexion and extension motion control (France and Paulos, 1994).

Liggins and Bowker (1991) quantitatively assessed 24 commonly used knee braces, including simple knee sleeves, hinged knee sleeves, sprung knee sleeves and frame (functional) knee braces. They found that the mechanical performance of any individual brace was dependant on three factors: the mechanical characteristics of the individual components from which the brace was constructed, the structural integrity of its design, and the interaction of the limb during loading.

Typically, each brace manufacturer and each FKB they design incorporates patented knee hinges into either one of the two different basic construction designs: the hinge-post-shell (shell) design and the hinge-post-strap (strap) design (Paluska, 2000; Vandertuin and Grant, 2004).



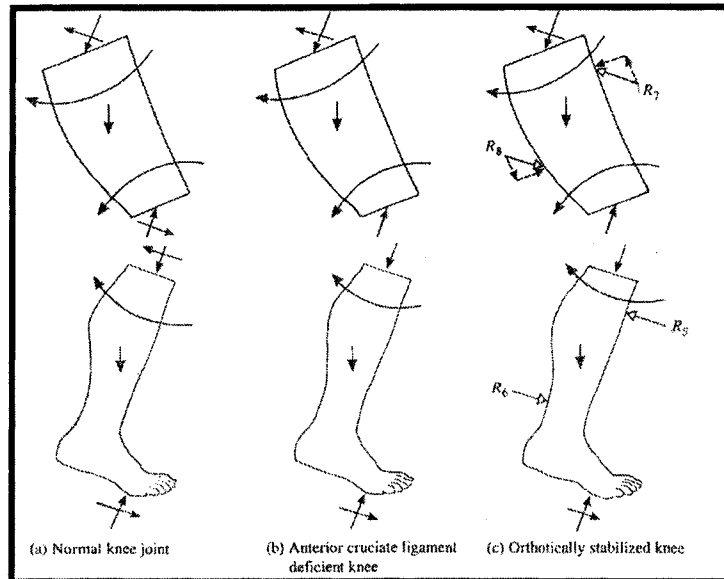
**Figure 9.** A) A typical hinge-post-shell brace. B) A typical hinge-post-strap brace (Vandertuin and Grant, 2004).

The shell design utilizes a semirigid or rigid plastic-carbon fiber shell that covers a large portion of the anterior thigh and tibia (Figure 9), while the strap configuration uses a rigid carbon-fiber frame. To date, no conclusive evidence supports the superiority of one brace construction design over the other. The hinge on FKB's may be placed either unilaterally or more frequently on both the medial and lateral sides of the knee and may be either one of the following designs: simple (monocentric, uniaxial), biaxial, polycentric, or multiaxial cam action (Podesta and Sherman, 1988). Additionally, the hinges typically incorporate variable flexion - extension stops to allow any range of motion desired (France and Paulos, 1994; Podesta and Sherman, 1988). The ideal kinematically designed hinge system would incorporate medial and lateral hinge components that would exhibit distinct patterns of motion, reflecting the differences in shapes of medial and lateral condyles and accommodating normal automatic rotation (Cawley *et al.*, 1991). Brace manufacturers' claims of superior simulation of natural knee motion provided by their knee hinge design over other designs have yet to be validated. Regalbuto *et al.* (1989) evaluated 4 different hinge designs (fixed axis, gear-on-gear, rack-and-pinion, and natural

3-D) using instrumented force transducers on each of the four uprights of the simulated knee braces and found that accurate hinge placement was more critical to reducing the mismatch between knee motion and brace motion, in terms of forces and moments at the hinges, than the hinge design kinematics.

Functional knee braces resist pathological knee motion either by passive or active means by incorporating straps and pads into a rigid framework, as four controlling points (Liggins and Bowker, 1991). These four controlling points are necessary to restore translational and rotational equilibrium at the calf and thigh (Figure 10). The posteriorly directed R5 external force restores translational equilibrium to the lower leg, while the R6 force re-establishes the rotational equilibrium. The forces of the brace on the thigh, R7 and R8, act in conjunction with external forces in the calf to restore equilibrium. This four point pressure system may be applied to the leg in either a passive or active manner. Passive braces apply resistive loads only during abnormal motion, while active braces apply active (dynamic) resistance at all times. Dynamic bracing, according to (Bledsoe, 2004), uses the act of knee extension to place an increasing anteriorly directed shear force across the tibia as the knee is extended. Unfortunately, few studies have classified the available knee braces on the market as either active or passive nor determined the functional advantages of one design over the other. The effectiveness of any functional brace, regardless of its design, is determined by the efficiency with which the controlling forces are transmitted from the orthosis/skin interface to the underlying bone structure. The amount of surface contact area that the 4 point force system makes with the leg along with the soft tissue characteristics of the braced individual, ultimately determine the efficiency of force transmission from the brace to the skeletal structure of the leg. The ability of a knee brace to accommodate soft tissue deflection is the greatest limiting factor in controlling knee motion. The shell FKB design attempts to provide greater tissue contact area. This serves to

enhance brace suspension, minimize soft tissue deflection and reduce brace contact pressure compared to the strap design.



**Figure 10.** The forces acting on the weight-bearing lower extremity with the knee flexed for the normal leg (A), the ACL deficient knee (B) and the orthotically stabilized knee (C) (Liggins and Bowker, 1991).

The numerous design features that comprise any FKB make it unique in its overall function in controlling anterior tibial translation. Cawley *et al.* (1991) noted that there is a tendency to generalize the results of testing one or two braces to all functional braces. “With a wide variety of functional knee braces available, it is a gross mistake to generalize the performance of one brace in a given test condition to all functional braces” (Cawley *et al.*, 1991). Unfortunately, brace manufacturers’ provide little scientific evidence to justify the claims of superior functionality of their brace designs over their competitors. However, a few studies have comparatively examined the more popular functional knee braces on the market by loading the knee of cadaveric or surrogate knees and measuring the strain of the tibia plateau relative to the femur (Beck *et al.*, 1986; Hoffman *et al.*, 1984; Liu *et al.*, 1994). Braces manufactured by DonJoy, Generation II, MKS II (Marquette Knee Stabilizer), Townsend, 3D 3-way, and Cti ranked among the best out of numerous brace designs, of the three papers, in providing the

greatest resistance to instrumented anterior loading of the knee. FKB design features suggested to account for these measured differences in ability to stabilize the ACL-deficient knee include: bilateral (medial and lateral) uprights vs. a single hinged bar/upright and increased side-bar stability (Hoffman *et al.*, 1984). France *et al.* (1990) stressed that knee braces with bilateral hinge bars were more stable than those with a single hinge bar.

#### 2.4.2 Functional Considerations

The significant controversy regarding the efficacy of functional knee bracing comes from a lack of agreement of the potential mechanisms by which ACL braces act to stabilize the knee. The potential mechanisms by which FKB's may act to improve knee stability has been investigated by numerous researchers and may include: enhanced proprioception (joint position sense) that the brace provides, increased passive resistance to anterior tibial translation and internal tibial rotation, improved reflex reaction time of muscles, and altered activity of musculature surrounding the knee, including reduced quadriceps inhibition and decreased hamstrings co-contraction. In a comprehensive review of literature related to proprioception and knee bracing, Beynnon *et al.* (2002) reported that the application of a functional brace to the ACL-deficient limb does not improve the joint's threshold to detection of passive movement.

##### *2.4.2.1 Passive Restraint Provided by Functional Knee Braces*

Passive restraint to anterior tibial loading provided by functional knee braces has been demonstrated by numerous authors to only occur at relatively low loads (below physiologic), as this reduction in displacement disappears as the applied load increases (Beynnon *et al.*, 2004; Branch *et al.*, 1988; Colville *et al.*, 1986; Liu *et al.*, 1994; Ott and Clancy, Jr., 1993). Markolf *et al.* (1986) stated that forces greater than 200N are required to produce an accurate measure of absolute laxity, which is much lower than the value of 400N (Noyes *et al.*, 1980) that occurs

during strenuous activity. Unfortunately, loads required to functionally challenge knee braces may not be applied *in vivo* without risk of injury (Liu *et al.*, 1994). The use of cadaveric knees to test knee braces passively confounds the validity of the results obtained, as “decreased tissue integrity and compliance, prevents firm application of the brace to the knee, while the rapidly changing mechanical properties of the periarticular tissue” (Liu *et al.*, 1994), that occurs with repetitive testing compromises the evaluation of knee instability. Additionally, braces may be fit on anatomic specimens at strap tensions not tolerated *in vivo* (Ott and Clancy, Jr., 1993). Despite these methodological limitations, functional knee braces tested *in vivo* show a decreased ability to restrain anterior directed tibial loading as this force approaches 200N.

In an *in vivo* investigation of the effects of functional knee bracing on anterior cruciate strain, by Beynnon *et al.* (2004), the restraining capacity of the brace diminished as the applied load approached the maximum load of 140N. This study evaluated ACL strain, directly through the use of an implanted strain transducer on the anterior medial band of the ACL, in patients requiring minor arthroscopic surgery. The DonJoy Goldpoint brace significantly reduced ACL strain for both anterior loading of the tibia and for internally-externally applied torques, compared to the unbraced condition.

In a similar study comparing the efficacy of DonJoy, Townsend, C.T.i., Lenox Hill, Bledsoe, Lerman, and 3D braces using a Hall-effect strain transducer applied to the ACL to quantify their respective abilities to resist applied anterior tibial and rotary loading, Beynnon *et al.* (1992) found that only the Don Joy and Townsend braces were associated with a significant reduction in strain as compared with no brace. The DonJoy, Townsend, C.T.i., and Lenox Hill braces significantly reduced an internally applied tibial torque. The DonJoy and Townsend braces reduced ACL strain in anterior tibial loading of relatively low magnitudes (100N), but failed to provide significant reductions in ACL strain at higher loads (180N). Although some of

the braces tested were successful at reducing ACL strain at low loads, no one brace was able to decrease ACL strain during isometric contraction of the quadriceps or active flexion-extension of the knee compared to unbraced conditions. In one of the earlier quantitative evaluations of the efficacy of FKB's, Colville *et al.* (1986) evaluated the Lenox Hill brace and determined that, despite the brace failing to significantly resist anterior tibial translation elicited by a maximal active drawer, patients experienced significant reductions in episodes of giving way, improved athletic performance by 69%, and 91% of patients examined reported satisfaction and beneficial use of the brace. Additionally, the authors found that despite the brace not reducing maximum anterior subluxation of the tibia that resistance to displacement did increase. Similarly, Liu *et al.* (1994) explain that resistance to anterior tibial displacement depends directly on the brace design and is inversely related to the applied force.

#### 2.4.2.2 Neuromuscular Effects of Functional Knee Braces

The effects of functional bracing has been demonstrated by numerous investigators to exert its influence on the neuromuscular system through various mechanisms, including: delayed onset of quadriceps activation (*vastus lateralis*) (Smith *et al.*, 2003); a favourable alteration in firing pattern of the quadriceps, hamstrings and gastrocnemius with “favourable” being determined as: significantly earlier activation of the hamstrings or gastrocnemius relative to the quadriceps or a significant delay in activation of the quadriceps (Smith *et al.*, 2003); increased amplitude and/or duration of hamstrings activation (Branch *et al.*, 1989; Diaz *et al.*, 1997; Nemeth *et al.*, 1997); alteration of co-contraction levels in the quadriceps, hamstrings and gastrocnemius (Acierno *et al.*, 1995; Ramsey *et al.*, 2003); decreased activation of the uninjured leg in expert downhill skiers (Nemeth *et al.*, 1997). Caution is suggested by several of these authors regarding a direct comparison of the results obtained from one study on the effects of FKB to the next investigation. In Ramsey *et al.* (2003), examination of the exclusive use of the



DonJoy Legend functional knee brace on EMG and kinematics, “one may not be certain that the results can be applicable to other functional braces”. Additionally, Acierno *et al.* (1995), stated that results of their maximal effort concentric isokinetic extensions testing protocol using EMG to analyse the stability producing effect of the Bledsoe ProShifter knee brace, “may not be directly applicable to other types of motion, such as eccentric contractions or complex contractions such as those involved in pivoting movements”. Due to the lack of literature on the dynamic or neuromuscular effects of FKB, it is difficult to make substantiated conclusions regarding the ability of braces to stabilize the knee. It appears as though the neuromuscular response of ACL-deficient patients to functional knee bracing is dependant on the functional stability of the patient (coper vs. noncoper) and the type of functional knee brace design (passive vs. dynamic).

In a study evaluating the effects of dynamic functional knee bracing on muscle activation patterns of ACL deficient patients, Acierno *et al.*(1995), revealed that symptomatic patients had quite different activation patterns than that of asymptomatic patients. Twelve subjects, (n=5 for high activity or asymptomatic and n=7 for low activity or symptomatic), performed maximum isokinetic knee extension on a KinCom isokinetic machine, with the Bledsoe Pro Shifter knee brace and without the brace. Symptomatic subjects exhibited an increased quadriceps activity and decreased hamstrings activity, and displayed a minor increase in force in the mid-range of flexion (80° to 40° flexion). Conversely, asymptomatic, or high activity subjects, showed no change in muscle activity, but demonstrated a decrease in extension force throughout the active range of motion. The results of this study suggest that dynamic bracing prevents quadriceps inhibition in symptomatic subjects by exerting a posteriorly directed force to the proximal aspect of the tibia, acting externally to compensate for the loss of the ACL. The increased stability provided by the brace enabled the quadriceps to contract more forcefully and reduced the need

for the hamstrings antagonistic intervention, as the hamstrings EMG was reduced to near normal levels.

In a more recent study of “coper” ACL-deficient subjects, Smith *et al.* (2003) found that the application of the C.Ti. brace, a passive functional knee brace, resulted in a significant delay in the average onset of vastus lateralis activation before landing in a single leg hopping task. The use of this brace significantly altered the onset latency of one or more muscles in 9 of the 10 subjects. The results of this study demonstrated significant interindividual variation in firing patterns observed and changes in the average onset latency of the quadriceps, hamstrings, and gastrocnemius, with brace use. Research (Wojtys and Huston, 1994; Beard *et al.*, 1993) has indicated that interindividual differences in muscle recruitment patterns exist. A lack of a universal recruitment pattern suggests that several adaptive strategies may exist to stabilize the knee in response to anterior tibiofemoral shear forces.

In 1989, Branch *et al.* examined the effects of the Lenox Hill and the C.Ti., (both passive knee braces), and found that ACL-deficient knees had 38% more and 32% higher lateral hamstrings activity with the braces compared to normal subjects in the swing phase of a cutting task. During the stance phase quadriceps activity decreased 13% in area under the curve and 11% in peak EMG activity, and the gastrocnemius decreased 12% in area under the curve and 9% in peak EMG activity. The medial hamstrings, however, increased by 46% in area under the curve and 36% in peak EMG activity. The ACL-deficient subjects in this study were not classified according to functional stability (i.e. coper vs. non-coper), therefore it is difficult to make comparisons between the results of this study and more recent studies, which dichotomize ACL-deficient subjects.

Due to the apparent paucity of research on the efficacy of functional knee braces on dynamic muscle activity assessed through EMG, it is difficult to make generalizations to one

specific ACL deficient patient population group or one type of brace versus another (static vs. dynamic brace).

## 2.5 JOINT STABILITY

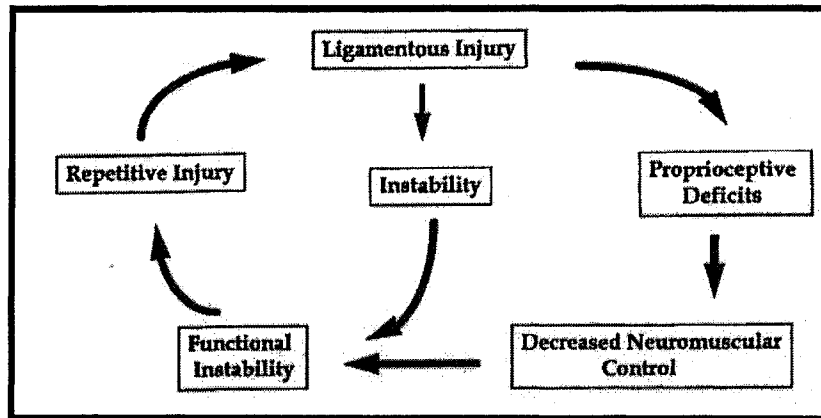
The term *stability* in reference to the joints in the human body has been defined in the literature under numerous contexts, including: static (passive) stability, dynamic (active) stability, functional stability, clinical stability, and mechanical stability. Numerous researchers have attempted to quantify and classify knee joint instability following ACL disruption based on many of these different definitions, with the exception of mechanical stability.

### 2.5.1 Knee Joint Stability

The passive stability of a joint (i.e. static stability or degree of laxity) is known to be contingent on joint geometry and the mechanical properties of connective tissues (ligament, joint capsule, and menisci) (Johansson *et al.*, 1991). The ACL has been classified as the primary restraint to passive anterior tibial displacement as, Noyes *et al.* (1980) determined that the ACL provided 85% of the total restraining force at a 5 mm anterior tibial displacement with the knee flexed at 90°.

Functional joint stability has typically been defined as the condition where the joint is stable and is absent of symptoms during physical activity (Johansson *et al.*, 1991). Inherent in the concept of functional stability is dynamic balance, where the forces acting externally on the knee are correctly balanced by passive and muscular forces acting internally. Functional stability “is determined by interaction of several factors, including: the passive restraints provided by the ligaments and other connective tissue structures, the joint geometry, the friction between the cartilage surfaces, and the load caused by the compression forces resulting from body weight and the muscles acting at the joint” (Johansson *et al.*, 1991). Additionally, another requirement of

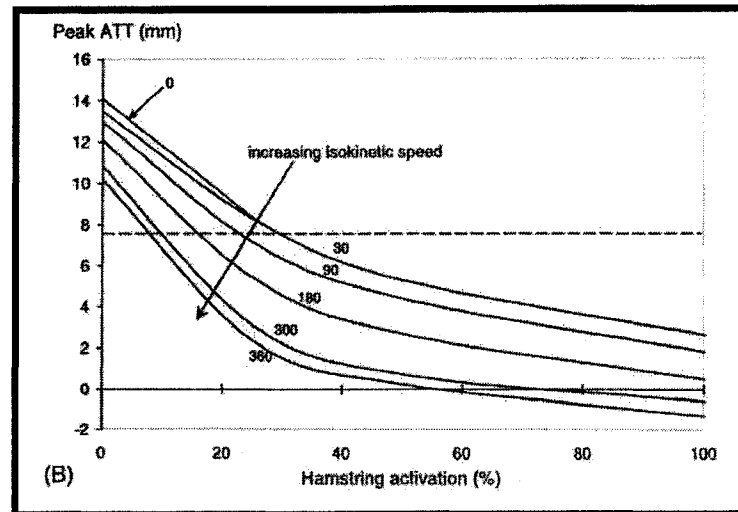
joint function is an absence of joint wear (Noyes *et al.*, 1980). The reestablishment of functional stability following ligamentous injury is dependent on the interaction of numerous factors as outlined in Figure 11.



**Figure 11.** A paradigm of functional stability, displaying the influence of mechanical instability and proprioceptive deficits on functional stability following ligamentous injury (Swanik *et al.*, 1997).

Following ACL injury, reflexive and preparatory muscular activation facilitated by the neuromuscular system is diminished. Dynamically, increased muscle activation is able to augment functional stability of the ACL-deficient knee through several mechanisms:

- 1) increased preparatory muscle stiffness reduces the electromechanical delay required to develop muscle tension (Solomonow *et al.*, 1987);
- 2) enhanced stretch sensitivity of the muscle spindle system through the gamma motorneuron system; and
- 3) increased joint load and subsequently decreased tibial translation in the anterior-posterior direction (Johansson *et al.*, 1991; Markholf *et al.*, 1978). Ultimately, the stability of the knee joint depends on the stiffness of muscles and ligaments within and surrounding the joint and may be considered the sum of the non-reflex and reflex mediated stiffness (Sinkjaer and Arendt-Nielson, 1991). The non-reflex stiffness is dependent on the number of engaged cross bridges in the active muscle and the orientation, cross-sectional area, and the tensile and shear modulus of each of the respective passive tissues (ligament, tendon, and other collagenous tissues).



**Figure 12.** Peak anterior tibial translation (ATT) is shown during isokinetic knee extension at 0, 30, 90, 180, and 300 deg/sec in the ACL deficient knee for different levels of hamstring activation. The horizontal line at 7.5 mm represents peak ATT with out hamstrings cocontraction and defines the limit of knee joint stability (Yanagawa *et al.*, 2002).

In an attempt to quantify the effects of hamstrings muscle activation on the stability of the ACL-deficient knee, Yanagawa *et al.* (2002) created a sagittal plane, mathematical, rigid linked model to simulate isokinetic contraction of the quadriceps. Hamstrings activation was preset at either 0%, 25%, 50%, or 100% and simulations were run at chosen isokinetic speeds of 30, 90, 180, or 360 deg/s. Anterior tibial translation was found to be inversely related to hamstrings co-contraction level and was also inversely related to extension speed in both the intact and ACL-deficient knee. During isometric extension (0 deg/s), the hamstrings must be activated to at least 35% of maximum in order for the model ACL deficient knee to remain stable. Conversely, when the model knee was extended at a rate of 360 deg/s, only 10% of maximum hamstrings activation was required to minimize the amount of anterior tibial translation. A horizontal line drawn at 7.5 mm (Figure 12) represents the peak anterior tibial translation calculated for isometric contractions of the quadriceps without hamstrings co-contraction and is used to define the limit of knee joint stability in the ACL deficient model. This model, although useful, overestimates anterior tibial translation and ACL force because the tibia

is constrained to move only in the sagittal plane. These limitations, along with the lack of true muscle activation patterns, make it difficult to apply these results to a clinical setting.

Doorenbosch and Harlaar (2003) quantified active stabilization of the ACL deficient knee through the use of a *co-contraction index*. This index provided an estimate of the agonist and antagonist muscle moments from EMG activity of the hamstrings and quadriceps in each direction during isokinetic flexion and extension. The co-contraction index (CCI) of the ACL deficient group of 5 subjects was calculated as 0.54, significantly higher than the CCI of 0.25 for the healthy group. The results of this study are suggested to be more clinically relevant, but do not differentiate between quadriceps inhibition/avoidance or increased hamstrings co-contraction as the primary mechanisms responsible for this increase in knee stability as assessed by the CCI.

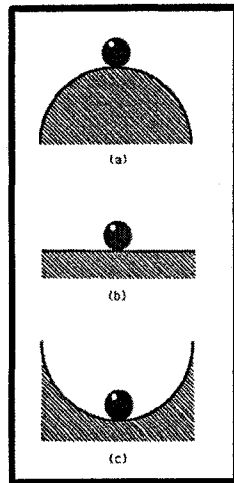
In examining the term stability it is necessary to differentiate between the meaning held by physicians and physicist. In medical practice, the physician associates the term stability with clinical stability, which is defined by White and Panjabi (1990) for the spine as:

The ability of the spine under physiological loads to limit patterns of displacement so as not to damage or irritate the spinal cord or nerve roots, and in addition, to prevent uncapacitating deformity or pain due to structural changes. Any disruption of the spinal components (ligaments, discs, facets) holding the spine together will decrease the clinical stability of the spine. When the spine loses enough of these components to prevent it from adequately providing the mechanical functions of protection, measures are taken to re-establish the stability.

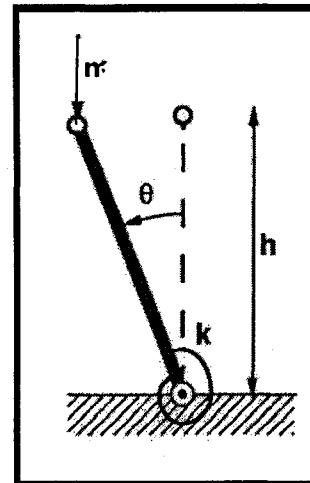
Mechanically, stability is defined as “the ability of a loaded structure to maintain static equilibrium even at small fluctuations around the equilibrium position. If stability does not prevail, an arbitrarily small change of the position is sufficient to cause *collapse*, i.e. the structure moves further away from equilibrium” (Bergmark, 1989). Unfortunately, no known investigation to date has attempted to formally define the mechanical stability of the knee in a manner similar to what has been done for the spine.

### 2.5.2 Joint Mechanical Stability

In examining mechanical stability, it is necessary to define the parameters under which stability may be measured. In 1989, Bergmark was the first to attempt to quantify and fully define the mechanical stability of the spine. “In elastic mechanical systems, equilibrium is necessary but not sufficient for stability” (Crisco III and Panjabi, 1990). In any mechanical system there exist two states of equilibrium: stable and unstable. A stable equilibrium infers that when a transient disturbing force is applied to the system, which afterwards is left to itself, the system will return to its original position” (Rozendaal, 1997). An unstable equilibrium, conversely, means that the effect of even very small disturbing forces increases in time, and even the slightest force on the system is sufficient to make it leave its equilibrium position. Bergmark (1989) maintained that a muscular system is considered to be in stable equilibrium when the potential energy ( $V$ ) of the system is at a relative minimum. In Figure 13, only the ball in c, the bottom sketch, is stable as it will return to its equilibrium position when perturbed. This simple ball analogy of stability is conceptualized in Figure 14, as a single degree of freedom column with rotational stiffness ( $k$ ) facilitating reestablishment of equilibrium when perturbed.



**Figure 13.** Simplified ball analogy of stability during static equilibrium. The ball in (c) returns to its equilibrium position when perturbed and is stable, while both (a) and (b) are unstable. (Crisco III and Panjabi, 1990).



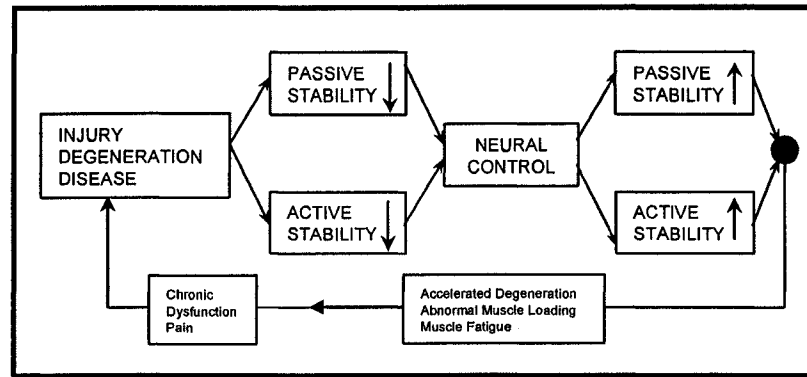
**Figure 14.** A single degree of freedom system to describe the components of stability: Mass ( $m$ ), height ( $h$ ), and rotational stiffness ( $k$ ). The mass applied to the system may cause a change in the potential energy of the system by altering the height of system and change the amount of energy stored in the spring, thus augmenting rotational stiffness. (Crisco III and Panjabi, 1990).

Mathematically, two conditions must be satisfied in order for stability to exist: 1) the system must be in mechanical equilibrium, meaning the first derivative of  $V$  with respect to angle ( $\theta$ ) (net moment) or the sum of all forces acting on the system must be zero; 2) the second derivative of  $V$  with respect to angle ( $\theta$ ) must be positive definite (greater than zero) indicating that  $V$  is at some minimum (Potvin and Brown, 2005). An additional requirement for calculating mechanical stability is that system being measured is in a quasi-static state. In a static mechanical system it is necessary for equations of equilibrium to always be fulfilled. In the quasi-static approach, static analysis of the moments, as determined from the body segment masses are only permitted during very slow dynamic motions (Cholewicki and McGill, 1996), where angular acceleration is zero. In the quasi-static approach, all dynamic properties, like momentum, are ignored and it is assumed that “the forces occur instantaneously, without any dynamic effect

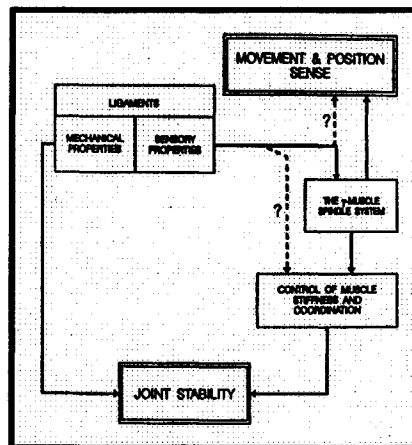


between the angle perturbation and the restoring force” (Rozendaal, 1997). A major caveat to the quasi-static approach exists: If quasi-static analysis indicates a system is unstable, the dynamic system may also be considered unstable; but when the quasi-static system predicts the system to be stable, there is no guarantee that the corresponding dynamic system is also stable (Rozendaal, 1997).

Panjabi (1992) conceptualized that the stability of the spine is contingent on the co-operative interaction of three independent subsystems: the passive (ligamentous) sub-system, the active (musculotendinous) sub-system, and the neural feedback sub-system. The ability of the spinal system to respond to dysfunction is one of the manifestations of its adaptability (Figure 15). However, this adaptability in response to injury or dysfunction is not unique to the spinal system. The knee, with traumatic injury to the ACL is able to enhance and regulate knee stability through modifications to one of these three subsystems. Johansson *et al.* (1991) conceptualized an organizational plan of joint stability and stiffness regulation in a similar manner to the spinal model developed by (Panjabi, 1992), as seen in Figure 16.



**Figure 15.** Dysfunction of the spinal stability system. Injury/disease may decrease the passive and or active stability. The neural control unit enhances the stabilizing function of the intact spinal components. This may lead to accelerated degeneration and may ultimately lead to chronic dysfunction and/or pain if the changes in neural control cannot sufficiently compensate for the loss of stability (Panjabi, 1992).



**Figure 16.** Regulation of stiffness of the muscles surrounding, the knee joint, and stability by the cruciate ligament joint afferents (Johansson *et al.*, 1991)

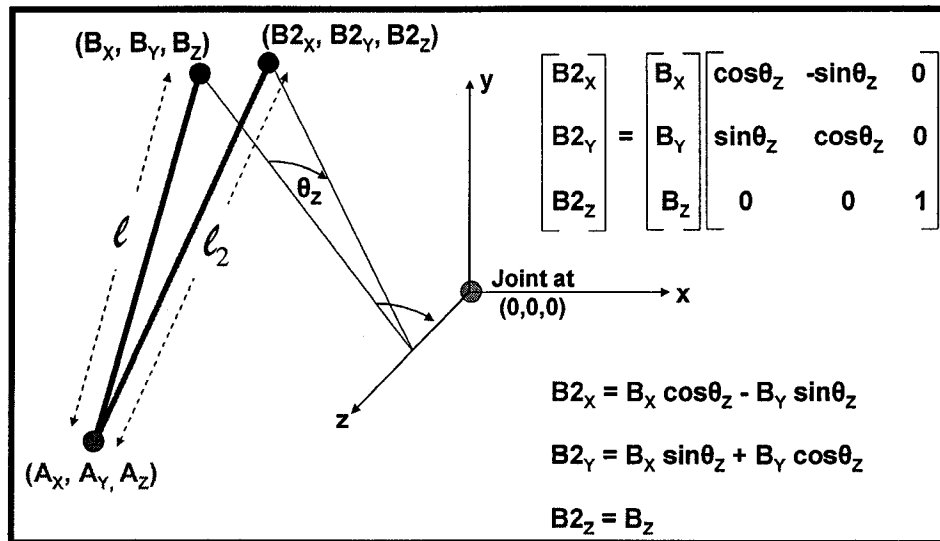
### 2.5.3 Calculation of Joint Stability

The concept of stability in the literature is usually only qualitatively discussed, but not quantitatively defined. This is particularly true for biomechanical models of the knee. Conversely, numerous studies to date have calculated the mechanical rotational stability of the spine. Two methods, the moment approach and the energy approach, have been used in the literature to calculate the overall contribution of any muscle to the

rotational stability of a specified joint. Unfortunately, the proposed models (Cholewicki *et al.*, 1999; Granata and Orishimo, 2001) appear to be limited to one single upright equivalent flexor and one single upright equivalent extensor. No indication is given regarding the ability to simultaneously analyze the contribution of multiple muscles to the stability of the spine. Bergmark (1989) and subsequent investigators (Cholewicki and McGill, 1996; Crisco and Panjabi, 1991; Granta and Wilson, 2001), using the energy approach, have examined the eigenvalues for each joint degree of freedom of the lumbar spine to quantify its stability through a stiffness and load matrix incorporating  $V$ . The criteria for the system to be stable is that, the overall determinant, plus each eigenvalue within the matrix, must be positive definite. The lowest eigenvalue, for which  $V$  is at a minimum, is considered the critical stiffness of the system. Any load that overcomes this stiffness will cause the system to become unstable and buckle. Therefore, systems with higher critical stiffness eigenvalues will be more stable. While, the results of this method provide a comprehensive examination of the stability of the lumbar spine, the intricacy in applying the required mathematical analyses can be quite significant and may be limiting to many researchers (Potvin and Brown, 2005). To this end, Potvin and Brown (2005) developed a simplified equation to quantify the stability contribution of individual muscles, as part of a multi-muscle system, about three axes of any specified joint.

Essentially, the stability of any specified joint may be determined with the coordinates of the individual muscles that cross the joint, the pre-activation level of the muscle, the functional moment arm, muscle length, muscle stiffness and the potential energy of the system. Using an anatomically based model, in which the insertion and origin coordinates are defined for each muscle relative to the joint being evaluated, the

functional moment arm and the length of the muscle may be determined. The force of any individual muscle may be predicted from measured surface EMG and used along with muscle length to calculate muscle stiffness. Bergmark (1989) calculated muscle stiffness as:  $k=qF/L$ , where  $L$  = total muscle length from origin to insertion,  $F$  = force, and  $q$  = dimensionless multiplier. In addition to the stiffness component of stability, pretension or pre-activation of a muscle will directly affect the work done by the muscle during a small rotation perturbation. This pre-activation component acts independently from its influence on muscle stiffness (Potvin and Brown, 2005). According to Potvin and Brown (2005), the stabilizing potential (stabilizing or destabilizing) of pretension is dependent on the orientation of the muscle about the axis of interest. They propose that the ideal stabilizer would have a short length and a long moment arm, with the moment arm being the more critical of the two. This method of calculating joint stability is briefly summarized in the methodological section of this thesis and may be seen in Figure 17.



**Figure 17.** Stability may be calculated for any muscular system given the origin coordinates ( $A_x$ ,  $A_y$ ,  $A_z$ ) and insertion coordinates ( $B_x$ ,  $B_y$ ,  $B_z$ ) relative to the axis of rotation of the joint of interest. The length of the muscle may be determined from this coordinate data. New insertion coordinates ( $B2_x$ ,  $B2_y$ ,  $B2_z$ ) may be determined using a transformation matrix and the new muscle length ( $\ell$ ) and change in muscle length ( $\Delta\ell$ ) may then be calculated (Potvin and Brown, 2005).

## 2.6 BIOMECHANICAL MODELLING OF THE KNEE

Due to technological restrictions and ethical considerations, the direct determination of forces transmitted by the structural members of the musculoskeletal system is very complex (Lu *et al.*, 1998). Estimation of the muscle forces to support or generate an external load applied to a joint may be achieved by one of two methods: Optimization or EMG in conjunction with an appropriate anatomical and muscle based model.

Optimization involves the prediction of muscle forces through the use of objective functions, such as: minimizing muscle stress, minimizing joint compression or shear force, or maximizing physiological efficiency. Optimization routines, typically, require anatomical simplification in order to overcome the high degree of mechanical redundancy of the locomotor system. In each leg, according to Crowninshield and Brand

(1981), there are 47 muscles which can influence locomotion, while the number of available equations to satisfy a dynamic or static equilibrium are far too few to simultaneously determine all the muscle forces acting at each joint. Static indeterminacy may be resolved by reducing the number of unknowns by using lumped muscle groups (Noyes *et al.*, 1992). However, by default, the use of an objective function cannot account for individual muscle recruitment patterns, co-contraction, non-uniform activation of synergistic muscles, and the selective muscle activation.

The use of EMG based anatomical models eliminates many of the constraints imposed by an objective function and allows for the estimation of muscle force by measuring individual muscle activity patterns. As well, EMG driven models take into account antagonistic activity and co-contraction, while facilitating the comparison of different muscle recruitment strategies used between subjects. For the purposes of this thesis, an EMG based approach will be used to determine the individual muscle forces and subsequently allow for the calculation of muscle stiffness to determine knee joint stability.

#### 2.6.1 EMG Based Lower Limb Musculoskeletal Model

The development of an EMG driven model to predict individual muscle forces requires the careful consideration of several parameters, including: an anatomical model, kinematics, kinetics and inverse dynamics, EMG to activation, Hill type muscle model, calibration and validation. Several researchers (Doorenbosch and Harlaar, 2003; Liu and Maitland, 2000; Lloyd and Besier, 2003; Wei, 2000) have successfully developed EMG based lower limb models by incorporating each of the above parameters.

### 2.6.1.1 Anatomical Model

The lower limb anatomical model developed by Delp *et al.* (1990), and further extended by Lloyd and Buchanan (1996), provides coordinates for the origin and insertion of muscles relative to a three-dimensional skeletal reference frame. An estimation of lines of action of the numerous muscles that span the knee joint may be determined and subsequently utilized, along with kinematic data ( $\Delta\ell$ ) and functional moment arm values, as inputs to calculate knee joint stability. The anatomical knee model developed by Lloyd and Buchanan (2003) included 13 musculotendon actuators, represented as line segments that wrap around bones or other muscles, including: semimembranosus, semitendinosus, biceps femoris long head and short head, sartorius, tensor fascia latae, gracilis, vastus lateralis, vastus medialis, vastus intermedius, rectus femoris, medial gastrocnemius, and lateral gastrocnemius. Two muscles, plantaris and popliteus, were omitted from the model due to their very small physiological cross-sectional area (PSCA) and negligible contribution to the total knee flexion and extension moment.

Differences in anthropometry between cadaveric specimens from which anatomical models are derived, and differences between these specimens and living subjects, is a potential source of error that may be eliminated through the use of osteometric scaling (White *et al.*, 1989). A scaling scheme based on measurable external landmarks (tibia and femur) allowed the estimation of all origins and insertions in living subjects with a significant reduction of inter-specimen variability (Brand *et al.*, 1982; White *et al.*, 1989).

### *2.6.1.2 Kinematics, Kinetics and Inverse Dynamics*

In order to quantify the change in length and determine the functional moment arm of a particular muscle, it is necessary to place markers over approximate joint locations, or alternatively, three marker triads are positioned in sets on each rigid body to be tracked (Yamaguchi, 2001). Simultaneous sampling of the markers, at each instant of time, is used to calculate segmental or joint angle generalized coordinates. Additionally, the tracking of markers facilitates the process of inverse dynamics. Inverse dynamics, ultimately may be used to determine net joint moments for validation of the EMG calibration (Doorenbosch and Harlaar, 2003).

### *2.6.1.3 EMG-Force Relationship*

Surface electromyography (sEMG) records the sum of motor unit action potentials and is useful in determining the force developed in an individual muscle and identifying the rate and degree muscular fatigue during prolonged contraction. Generally, the best relationship between EMG and force occurs under isometric conditions (Bigland-Ritchie, 1981), due to a lack of change in muscle length. Previously in the literature, significant controversy existed regarding the linearity of the EMG to force relationship (DeLuca, 1996). According to Bigland-Ritchie (1981), EMG-force relationships differ in their degree of linearity based on the physiological properties of the muscles involved. Muscles with a more homogeneous composition (Type 1) tend to produce more linear relationships, while muscles of mixed fibre distribution tend to have a non-linear relationship.



Estimation of force from EMG requires the consideration of both the passive and contractile elements of muscle tissue. The force created by the contractile element is assumed to be dependent on current length, velocity of contraction, and activation (Zajac, 1989). Additionally, the force of a particular muscle may be ascertained from the surface EMG signal using the product of normalized EMG, muscle cross-sectional area, a gain factor, and muscle parameter modulation factors including, muscle length and velocity (Marras and Sommerich, 1991).

#### *2.6.1.4 Calibration and Validation*

Inherent in the modelling of any biological system is error and in EMG driven models, potential sources of error or inaccuracies may include: the degree to which predicted joint centres matched with actual subjects; misalignment of joint centres when the subject stands may not reflect subjects anatomical variability and may not have been accurately represented in the original musculoskeletal database; inaccuracies inherent in locating and marking bony landmarks superficially; variation in optimal fibre length in humans; individual differences of moment arms defined by variable muscle origin, insertion and/or wrapping points. Typically, EMG based biomechanical models are validated by assuming that the summed moments derived from EMG is equal to the external moments estimated utilizing an inverse dynamics model (Lloyd and Besier, 2003). Doorenbosch and Harlaar (2003) used inverse dynamics to validate the results of their EMG based model of ACL deficient patients, but additionally calibrated the isokinetic EMG levels of the knee extensors and flexors to their respective exerted concentric moments. The concentric knee moment was measured on the dynamometer as a function of knee angle to obtain calibration factors for each muscle group. Lloyd and Buchanan (1996) used two global gain factors, one for the extensors and one for the flexors, to account for the variability of strength in these muscle groups between

individuals. Essentially, the parameters that define each component of the EMG model must be scaled as such to match the joint moments derived from inverse dynamics or use of an isokinetic dynamometer.

## CHAPTER 3

### METHODOLOGY

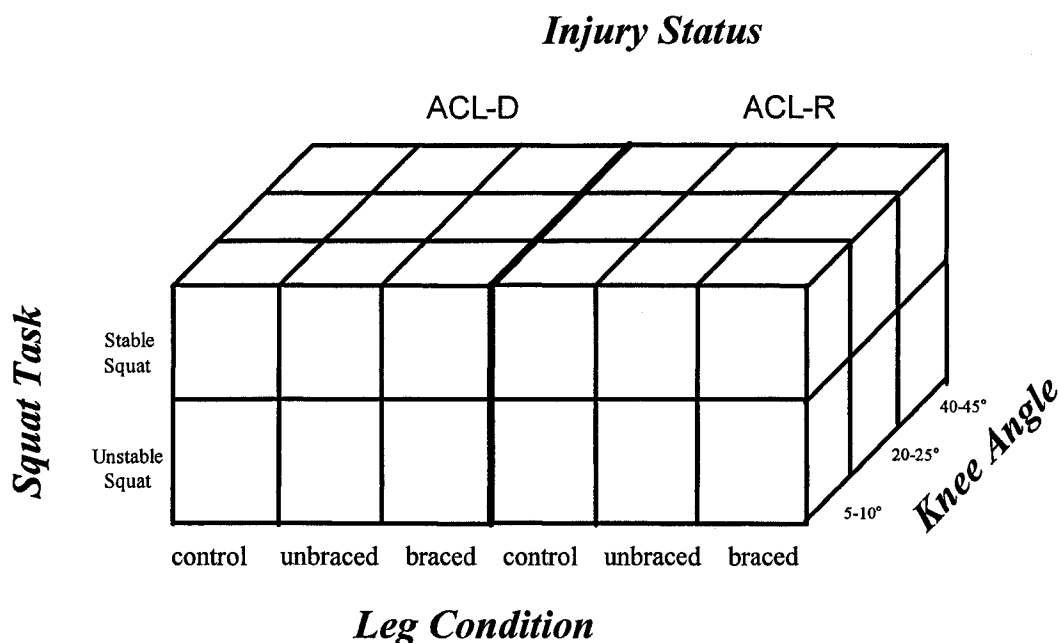
The scope of this project involved three components: 1) the recruitment and screening of ACL deficient and reconstructed subjects; 2) the development of an EMG-driven biomechanical model of the lower extremity with particular emphasis on tibiofemoral and patellofemoral kinematics; 3) the application of this model for the purposes of comparing the muscle recruitment and knee joint rotational stiffness of ACL deficient and reconstructed subjects, with and without functional knee brace use.

Prior to participating in the study, subjects read and signed a written consent form. The methodological aspects of this study received approval from the University of Windsor Research Ethics Board (Appendix A).

#### 3.1 STUDY DESIGN

This study examined the ability of functional knee braces to augment muscle contributions to flexion/extension rotational joint stiffness ( $JS_R$ ) of the knee joint in ACL deficient and reconstructed subjects. Subjects performed two different squat tasks, similar in nature to those found in rehabilitation and sport settings, for three different *Leg* conditions: *control* (uninvolved) leg, *unbraced* (involved) leg, and *braced* (involved) leg. An unstable single-leg squat and a stable single-leg squat were performed to determine the influence of functional/balance exercise on the muscular recruitment patterns required to stabilize the knee with and without brace use. The custom fabricated wobble board had a larger arc about which it pivoted, compared to the manufactured version. This arc had a diameter of 0.38 m, with the bottom of the arc 0.06 m from the flat platform, and was much more stable than the smaller radius on the Fitter wobble board (Figure 20). Pilot

testing confirmed that this diameter was adequate to provide an elevated neuromuscular response, while allowing subjects to balance without falling off the apparatus. The independent variables in this study, as described previously, were the repeated/within measures of *Squat Task* (stable and unstable), *Leg Condition* (control, unbraced, and braced), *Knee Angle* (5-10°, 20-25°, and 40-45°) and the between variable of *Injury Status* (deficient and reconstructed knee) (Figure 18).



**Figure 18.** The study design is represented as a 2 x 3 x 3 x 2 matrix of the independent variables for ACL-Deficient and Reconstructed groups. For within variables, there were two *Squat Tasks*, three *Leg Conditions* and three *Knee Angles*. The between variable of *Injury Status* had two levels.

### 3.2 SUBJECTS

Twenty male subjects, characterized as either ACL deficient or ACL reconstructed (Appendix C), were recruited from the university's athletic therapy clinic, the varsity sports' teams and a local orthopaedic surgeon. All study participants were

screened using a subject selection protocol. The exclusion criteria were quite specific and required all of, the following: 1) no known injury pathology or previous injury to the lower limb being tested; 2) contralateral lower limb injury; 3) documented neurological, metabolic bone, or connective tissue disease; 5) knee pain, effusion, range of motion limitation, or crepitus at time of study.

Subjects that satisfied the above criteria completed the following knee rating forms (prior to study participation): Tegner Activity, Lysholm Scales, Knee Outcome Survey (ADL and Sport), International Knee Documentation Committee (IKDC) Subjective and Knee History (Appendix A). These forms, along with the completion of a timed single leg hop test, were used to determine each subject's functional level. Subjects were allowed to warm-up prior to completing three single-leg hops for six metres. Subjects wore the knee brace during these trials. Each trial was timed using a Brower Wireless Sprint Timing System (Draper, Utah, USA). ACL deficient subjects were identified as either copers or non-copers based on the following criteria: non copers scored below the cut-off score identified for each of the utilized knee rating forms and scored less than 80% (injured vs. uninjured leg) on the single leg-hop test. Potential subjects wore an off the shelf knee brace for increased support during this dynamic test. Additionally, measurements were made of each subject's height and the bilateral circumferences of each thigh (obtained at six inches above the superior pole of the patella were taken). The thigh circumference measurements provided a reference (requirement for Lysholm Activity Scale) for potential differences in function between the involved and uninvolved legs.

### 3.3 FUNCTIONAL KNEE BRACE FITTING



**Figure 19.** FLEX Off the Shelf Functional Knee Brace from Innovation Sports.

Subjects selected for participation in the study were fit with the FLEX functional knee brace from Innovation Sports (Foothill Ranch, CA, USA). Innovation Sports supplied 5 different sized braces in left and right models. The thigh and calf shells of this brace are constructed from a heat mouldable carbon fibre polymer. Each subject was fit with a brace to their affected leg, following the manufacturer's directions. A heat gun was used, when necessary, to improve the fit of the brace by reshaping the thigh and/or calf shells to accommodate the anatomy of each subjects affected leg. Additionally, different thicknesses of condyle pads were used to ensure a snug fit in these areas. A best attempt was made to ensure that the brace fit each subject appropriately.

### 3.4 EXPERIMENTAL PROTOCOL

Subjects that met the inclusion criteria for the experiment were given a brief orientation to the experimental protocol. Specifically, subjects were given practice balancing on the wobble board to ensure they were comfortable performing this task throughout the course of the experiment.

Maximal voluntary contractions (MVCs) were performed by each muscle, for the purpose of normalization of experimental EMG data. The skin overlying and surrounding the electrode placement sites, specified for each muscle, was shaved and cleaned with sterile alcohol, prior to electrode placement. Subjects were appropriately positioned and asked to perform an isometric contraction for each muscle group according to each particular muscle's function. Manual resistance was applied in a direction opposite to that of each muscle's specific function. Each MVC trial was comprised of three maximal exertions against manual resistance. Each exertion consisted of a ramp up period, a maximum exertion period and a relaxation period over three to four seconds. After each MVC trial, subjects were asked if the effort they provided during the trial was their maximum. MVC trials were repeated for those trials rated by the subject as sub-maximal. Additionally, a noise trial was collected for each leg in order to determine the bias that needed to be removed from the EMG data obtained for the MVC's and trials for each of the three leg conditions.

More specifically, the MVC's for each muscle were achieved as follows: Gastrocnemius (lateral and medial heads), resistance to plantarflexion was provided by two dumbbell cable handles secured by a chain to a wooden platform. During this MVC trial, subjects held onto the cable handles while standing on the platform. The length of the chain was adjusted for each subject, so that approximately 50% of ankle plantarflexion was possible when subjects were asked to push maximally upwards (Flynn et al., 2004); Quadriceps, resistance to knee extension was achieved by providing manual resistance to the anterior aspect of the ankle and lower leg as subjects were requested to produce a maximum knee extension contraction. Knee angle was maintained between 90-

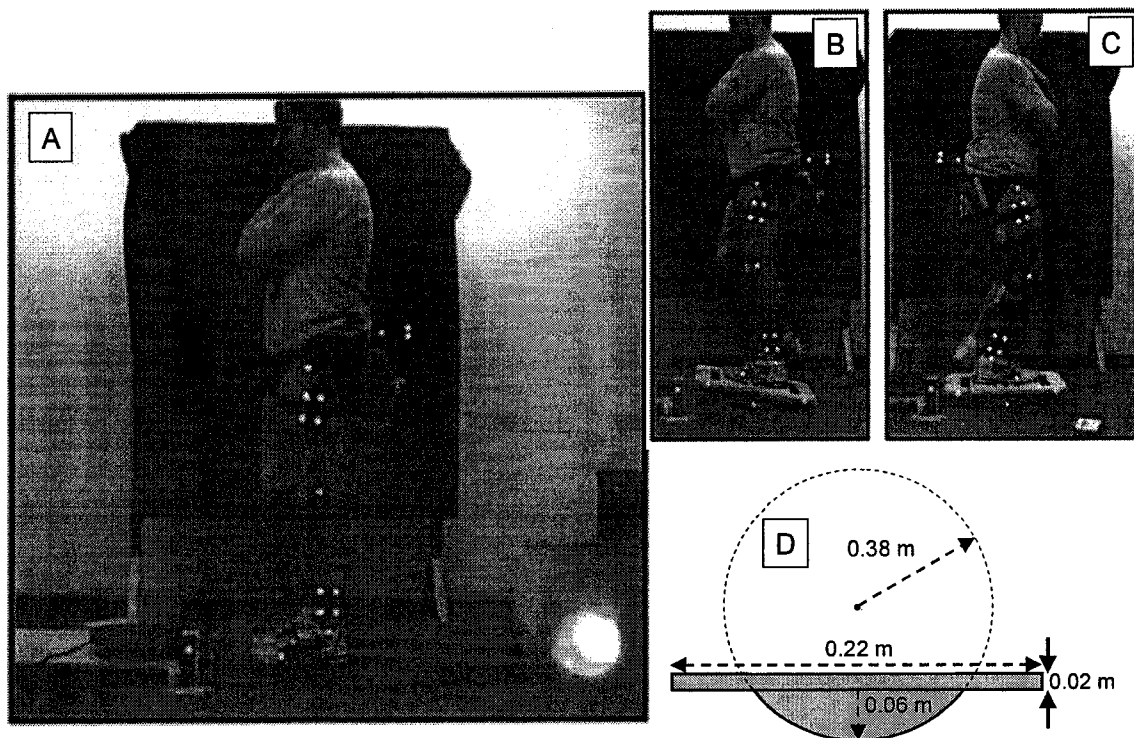
100 degrees, as Newman et al. (2003) found that the angle/torque relationship was not different between 50-70 degrees and 90-110 degrees, when percutaneously stimulated. Medial and Lateral hamstrings, manual resistance was applied to the posterior aspect of the lower leg with the knee in 90 degrees of flexion. Subjects were in an upright standing posture for these trials; Tensor fascia latae, resistance was applied to the lateral border of the foot by an inanimate object (wall). Subjects were standing and abducting their hip at the same time as attempting to internally rotate the thigh.

After the MVC and noise trials, reflective anatomical markers were placed on the subjects lower extremities and an electrogoniometer was secured to the posterior aspect of the thigh and lower leg using two-sided tape. A kinematic reference position was recorded with a video camera while the subjects were standing in an upright posture. The electrogoniometer was zeroed while the subject stood upright to verify that the knee was in full extension. Subjects were provided with continuous biofeedback of knee angle for the duration of each trial. A bias trial was then recorded from the force plate. This bias was subtracted from the analog signal in the data collection software to provide an accurate measurement of the shear and ground reaction force, and centre of pressure (COP).

Subjects then performed each of the single-leg squat tasks for all three leg conditions (Figure 20). Each leg condition was presented in a randomized order. Three trials were performed for each leg condition and squat task-knee angle posture combination blocked on leg condition. Eighteen trials were collected per leg condition, for a total of 54 trials per subject. The testing was done in this manner to minimize any chance of order effects from occurring during the course of the study. For both the stable and unstable single leg squat tasks, subjects were instructed to maintain their knee angle



at either 5-10, 20-25, or 40-45 degrees for the duration of the 10 second trials. Subjects were requested to cross their arms across their chest and to position themselves perpendicular to the field of view of the digital video recorder (parallel to the FY-plane of the force plate). The data collection program was manually triggered to start recording EMG and force plate data once subjects consistently displayed the desired knee angle. The trigger was activated by a switch that simultaneously controlled current flow to the A/D card and a LED. This LED was placed in the camera's field of view to provide a synchronization point for the forceplate, EMG and kinematic data. For the unstable squat tasks, the wobble board remained secured in a horizontal position until a switch was activated to disengage a pin (0.64 cm {1/4"} diameter) from a hole located in the edge of the board, perpendicular to the axis of wobble. This pin is pulled out of engagement by activating an electrically powered solenoid. The subject was first given a visual warning of the impending instability of the board by the switching on of a light bulb. The time interval between the visual indication and activation of the solenoid was preset to one second. For additional information on the solenoid, the wobble board and the control of its release, see Appendix B.



**Figure 20.** A) stable squat task, B) unstable squat task without brace, C) unstable squat with brace, D) schematic diagram of the unstable platform dimensions. In A) the LED is shown in the camera's field of view, indicating that EMG and force plate data collection has commenced. The anatomical markers, the pelvic fin with three markers, and the two cluster sets of thigh and lower leg markers are shown.

### 3.5 INSTRUMENTATION AND DATA ACQUISITION

A number of recording devices were utilized to collect the data required for input into the biomechanical model. The EMG and force plate data were collected with customized LabView software (National Instruments, Austin Tx.) using a PC compatible computer. Analog signals obtained from each instrument were converted to digital signals using a 12-bit A/D card (National Instruments, Austin Tx.). A twin-axis electrogoniometer (Biometrics Ltd., Model-SG110), secured onto the posterior aspect of the thigh and leg, provided subjects with a visual display of knee angle throughout each trial. A custom LabVIEW program (National Instruments, Austin Tx.) converted the

analog signal from the electrogoniometer into a digital display of knee angle. These data were used only for biofeedback and were not saved.

### 3.5.1 Electromyography

Surface EMG was measured for 8 of the 13 muscles that cross the knee joint. Preliminary biomechanical modelling of the lower extremity revealed that the gracilis and sartorius muscles make negligible contributions to knee joint stability and, therefore, were not included in the collection or subsequent biomechanical model. Based on Derouin & Potvin (2005), the gracilis provided only 0.65% of the overall JS<sub>R</sub> about the flexion-extension axis, while the sartorius contributed only 0.12%. Eight channels of EMG were used to record the activity of the remaining 11 muscles. The following muscles were grouped together due to the spatial limitations and accessibility of recording electromyography from surface electrodes: biceps femoris short head with biceps femoris long head and semitendinosus with semimembranosus. Although vastus intermedius cannot be measured using surface EMG, its activity was estimated from the electromyography values recorded for both the vastus lateralis and the vastus medialis. Akima et al. (2004) determined, using magnetic resonance imaging, a significant correlation between activity of these muscles, in healthy male subjects, after performing isokinetic knee extension at 120 degree/second. A strong correlation of the muscle activity of vastus intermedius and vastus lateralis ( $r = 0.719$ ,  $p < 0.0001$ ) and between vastus intermedius and vastus medialis ( $r = 0.783$ ,  $p < 0.0001$ ) was found. The semitendinosus and semimembranosus were lumped together and collected as the medial hamstrings channel. All textbooks on needle and surface electromyography technique referenced in this manuscript, group the semitendinosus and the semimembranosus

together as the medial hamstrings. As well, the long and short heads of the biceps femoris were recorded together from the lateral hamstrings channel.

Disposable bipolar Ag-AgCL surface electrodes (Medi-trace disposable electrodes, Graphic Controls) were positioned in an orientation parallel to the muscle's line of action, somewhere between the myotendonal junctions and innervation zones of each respective muscle. The inter-electrode distance was 2.5 cm. The placement of the electrodes over each muscle was as follows, according to standardized electrode placement (Basmajian, 1985; Cram, 1998; Geiringer, 1994; Perotto et al., 1994; Saitou et al., 2000): lateral head of gastrocnemius (one third length of the distance from the head of the fibula to the tuberosity of the calcaneus on the heel; 2 cm lateral to the midline); medial head of gastrocnemius (approximately 2 cm medial to the midline of the leg, over the bulge of the muscle belly); vastus lateralis (3 to 5 cm above the patella, on an oblique angle just lateral to midline); vastus medialis (one fifth of the distance from the medial tibial plateau to the anterior superior iliac spine, or 2 cm medial to the superior rim of the patella at an oblique angle of approximately 55 degrees, along the line of action of the muscle); rectus femoris (50% of the distance between the anterior superior iliac spine and the superior pole of the patella); tensor fasciae latae (2 cm inferior to the anterior superior iliac spine); biceps femoris long head and short head (two thirds of the distance from the greater trochanter to the back of the knee); semitendinosus and semimembranosus or medial hamstrings (half the distance between the gluteal fold and the posterior aspect of the knee, approximately 3 cm from the medial border of the thigh). Slight modifications to electrode placement (up to 1 cm) were made to accommodate the functional knee

brace, similar to the protocol used by Smith et al. (2003). For each trial, all surface EMG data were amplified 500 to 1000 times (CMRR > 80db) prior to sampling at 2100 Hz.

### 3.5.2 Force Plate Data

Force plate data (Advanced Mechanical Technology, Inc., Newton, Mass, USA) were measured and input into a four segment linked segment model to calculate the flexion/extension moments produced about the ankle and knee joint. The magnitude of the force acting on the foot and the locations of the center of pressure were recorded from the force plate during each of the single leg squat tasks, for each leg condition. The force plate data were sampled at a rate of 2100 Hz and calibrated to Newtons.

### 3.5.3 Single Leg Squat Kinematics

A Sony digital video camera with a 30 Hz sampling frequency was used along with spherical reflective anatomical markers to define the foot, lower leg, thigh and pelvis segments in the sagittal plane. Markers made of reflective tape were placed over the anatomical landmarks of interest (greater trochanter, lateral condyle of the knee, lateral malleolus, heel, and head of the fifth metatarsal). A rigid fin was fixated with two-sided tape over the sacrum. This fin had three reflective markers that were used to facilitate the measurement of pelvic tilt relative to the greater trochanter marker and ultimately assist in determining hip angle. Additionally, two sets of marker clusters were placed on the lateral aspect of the thigh and lower leg. Each cluster set was oriented parallel to the sagittal plane and was comprised of four spherical reflective markers that were mounted on a contoured polypropylene shell. These markers were used to verify the position of the knee joint for the braced leg condition.

The anatomical markers were digitized using the Ariel Video Analysis system (Ariel Performance Analysis Software, Ariel Dynamics, Inc, San, Diego, CA). Then, the digitized markers were filtered using a dual pass Butterworth filter with a cutoff of 3 Hz.

#### 3.5.4 Subject Data

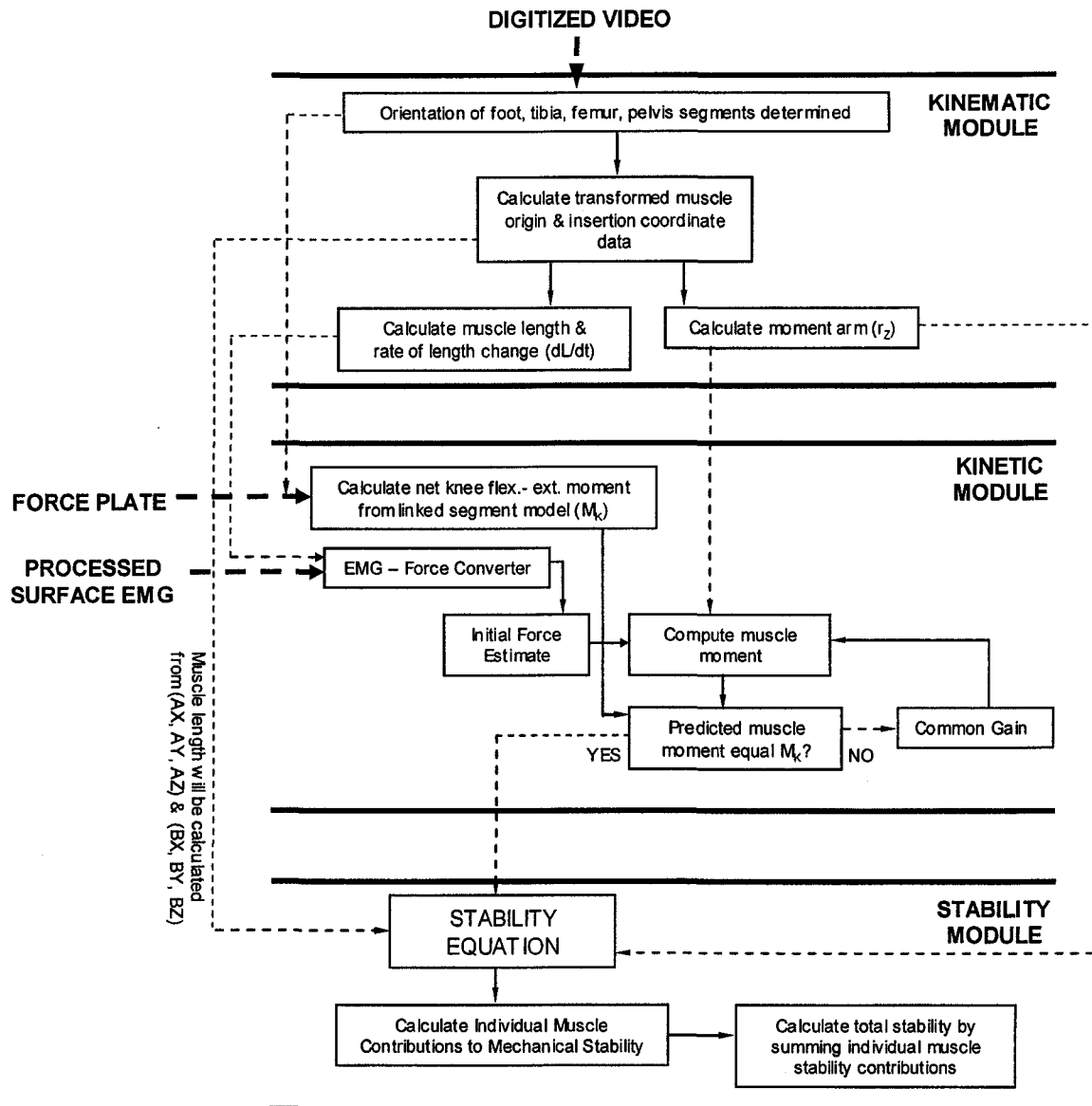
Scores from the knee rating scales, the single leg hop test, and other information regarding the subject's injury history and assessment were entered into a spreadsheet. Following the completion of the study by all the subjects, these data were grouped accordingly and stored on a PC for further analysis.

### **3.6 DATA ANALYSIS**

The determination of knee  $JS_R$ , about the flexion-extension axis of the knee, required an integration of the following parameters: anthropometric tables, a linked segment model, kinematic data, force plate data and surface EMG. Figure 21 outlines the sequential steps that were required to calculate the individual muscular contributions to knee joint stiffness.

#### 3.6.1 Anthropometric Data

Digitized subject anthropometry, and the calculated mass for each subject, were combined with published anthropometric data (Winter, 2005) to determine the mass and center of mass (COM) location for each segment. Each of these parameters was used to calculate the net moment acting at the knee from the measured force plate data.



**Figure 21.** Flowchart of the data analysis and model progression. Kinematic and kinetic data were calculated to ensure static equilibrium was achieved before  $JS_R$  was calculated. Average moments and force estimates were input into the stability model.

### 3.6.2 Muscle Orientations

The model of Delp et al. (1990) defines the lines of action of 43 musculotendon actuators, with their respective bony attachments to seven segments. The determination of muscle coordinate data was required for the three different knee angle ranges for each

leg and squat condition. For each trial, the linked segment model was manipulated to duplicate the instantaneous postures defined by the kinematic data which were determined by the joint markers and marker clusters. These kinematics were combined with estimates of subject anthropometry and force plate time-histories, to estimate the instantaneous ankle, then knee, joint reaction forces and moments. Averages of the knee flexion/extension moment were iteratively calculated with a sliding 3 second window. The interval with the lowest coefficient of variation in the knee moment represented the window where all averages of kinematic and kinetic data were taken for that trial. These averages were then input into the EMG-based biomechanical model to estimate individual muscle forces and contributions to knee flexion/extension JS<sub>R</sub>.

The manipulation of the model was achieved through the various kinematic functions described by Delp et al. (1990). Polynomial equations were fit to this kinematic data that describes the relative (x, y, z) translations of the tibial-femoral joints and the patello-femoral joints. Rotation matrices were used along with these kinematic functions to determine the precise coordinates of the origin, insertion and nodes (where applicable) for each muscle in each of the three knee angle ranges. These coordinate data were then used to determine each muscle's average length, average velocity and moment arm (rZ). These data were used as inputs in the biomechanical model described below.

### 3.6.3 EMG Processing and Initial Force Estimates

Surface EMG signals were digitally bandpass filtered (20-490 Hz), rectified, and low pass filtered using a first order Butterworth filter with a cutoff frequency of 2.0 Hz and normalized to the MVC amplitudes. The average EMG amplitude for each trial was calculated from a three second window where the net knee moment had the lowest



coefficient of variation. The processed EMG was then used to provide an estimate of force for each muscle. The biomechanical model developed by Delp et al. (1990) provides the maximal force capacity, active-passive force length data, and force-velocity data for each muscle of the lower extremity. This information was used to account for the force-length and force-velocity relationships in the biomechanical model that was developed. An initial estimate of the force produced by each muscle was calculated as follows:

$$\text{Force}(t)_j = \text{FMAX}_j * \text{EMG}_j(t)/\text{EMG}_{\text{max}j} * v(\text{Vel}_j) * l(\text{Len}_j) * \text{FPA}(\cos\alpha)$$

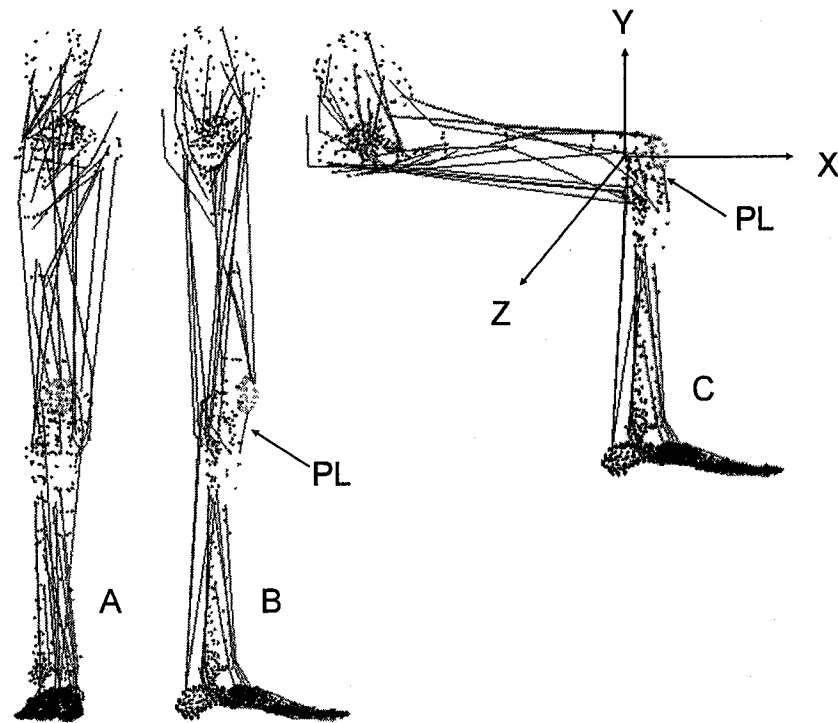
where,  $\text{Force}(t)_j$  is equal to the force of any given muscle,  $j$ , acting through its tendon at a given time ( $t$ );  $\text{FMAX}_j$  is the maximum force capacity of the specified muscle,  $j$ ;  $\text{EMG}_j(t)$  is the measured instantaneous EMG level ;  $\text{EMG}_{\text{max}j}$  is the maximal amplitude determined from the MVC contractions;  $v(\text{Vel}_j)$  and  $l(\text{Len}_j)$  are modulation factors that describe the EMG and force relationship as a function of muscle velocity and muscle length, respectively.  $\text{FPA}(\cos\alpha)$  is the force in the tendon after consideration of the pennation angle,  $\alpha$ .

#### 3.6.4 Biomechanical Model

In order to calculate rotational stiffness about the knee joint, an anatomically based biomechanical model of the lower extremity was integrated into a stability model developed using the QuickBASIC software program (Microsoft Corporation, 1985-1988, Redmond, Washington, USA.). A linked segment kinetic model incorporated the following skeletal structures for the lower extremity: pelvis, femur, shank, and foot.

An anatomical model of the lower extremity was developed based on the work of Delp et al. (1990). The muscle input file provides data on the following parameters: 1) a

dimensionless force-length curve for tendon, 2) a dimensionless force-length curve for muscle, 3) a dimensionless force-velocity curve for muscle, and 4) definitions for 43 of the lower limb muscles, including the muscles' three-dimensional line of action, nodes defined at specified joint angle ranges, tendon slack length, optimal fibre length, pennation angle and peak isometric force. The joint input file is defined as having four components: 1) body segments, 2) generalized coordinates, 3) joints, and 4) kinematic functions of the lower-limb model. Using the data provided in the muscle and joint input files, a biomechanical model of the lower extremity was developed (Figure 22) to determine the maximum stabilizing potential of 11 of the 13 muscles, crossing the knee, to flexion/extension knee joint stiffness. The model was defined in a Cartesian coordinate system, with the x-axis directed anteriorly from the tibial-femoral joint, the y-axis oriented vertically along the shaft of the tibia and the z-axis is represented as the axis about which knee flexion and extension occurs. The axis of knee flexion for the model is defined in the tibial reference frame as (0, 0, 0). For the purposes of calculating muscle contributions to joint stiffness, the origin coordinates ( $A_x$ ,  $A_y$ ,  $A_z$ ) and the insertion coordinates ( $B_x$ ,  $B_y$ ,  $B_z$ ) were defined relative to the knee joint. The following two sections explain how the knee joint kinematics and total flexor/extensor reaction moment were calculated at the knee for both squat tasks.



**Figure 22.** Sagittal and frontal view of an anatomical model of the lower extremity in two postures. A) Frontal view of model is shown in a neutral posture B) Sagittal view of model in a neutral posture C) the model is shown in a squat posture with the knee and hip flexed to 90 degrees. Note the change in orientation of the patellar ligament (PL) from the neutral posture to the squat posture (90 degrees of hip and knee flexion).

#### 3.6.4.1 Single Leg Squat Kinetics

Each segment of a link segment model can be considered to act independently and, therefore, was analyzed in a distal to proximal sequence. Derived kinematic data for each segment (angular acceleration  $\alpha$ , horizontal and vertical acceleration) were used, along with forces (gravitational, ground reaction, and muscle forces) acting on the link segment model, to calculate unknown reaction forces and the net muscle moment acting at first the ankle joint, then the knee joint. Determination of the instantaneous net muscle moment and reaction forces at the ankle was accomplished for each frame of interest as follows: 1) Input of segmental parameters and anthropometric variables for each subject,

2) calculation of the gravitational forces and COP acting at the foot, 3) calculation of the horizontal and vertical reaction forces at the ankle joint, 4) calculation of the moment of inertia ( $I_0$ ) of the foot, 5) calculation of the sum of the moments acting on the foot to determine the net ankle reaction moment, 6) the net ankle reaction moment was transferred to the lower leg segment and the net knee reaction moment ( $M_K$ ) was determined using the same sequence of steps. The average calculated knee flexion-extension moment with the lowest coefficient of variation, over a three second window throughout each trial, was input into the anatomical model. This window was determined after the solenoid triggered the instability of the wobble board. As noted previously, this same window was used to determine the average EMG and joint angles for each trial.

#### *3.6.4.2 Final Muscle Force Estimate*

The initial muscle force estimates (described in Section 3.6.3) were then multiplied by their respective flexor/extensor moment arms to estimate each muscle's contribution to the flexor/extensor moment acting at the knee. These individual muscle moments were then summed to determine the initial estimate of the net EMG-based muscle moment acting at the knee. This muscle moment was then compared to the net knee reaction moment determined with the methods described above. The ratio of the estimated EMG-based muscle moment, divided by the more accurate knee reaction moment determined with the linked segment kinetic model, was calculated for each trial. This most appropriate single common gain was determined for each subject through a linear regression analysis of the ratio between the EMG moment estimate and the measured net knee moment, for all 18 of the experimental conditions. An estimation of the net knee moment was determined from the following:  $y=mx + b$ , where  $y$  is the net

knee moment,  $m$ , is the slope of the best fit line,  $x$ , is the EMG measured moment and  $b$  is the y intercept of the line. The gain was defined as the slope ( $m$ ) of the best fit line and intercept was set to zero. This gain was multiplied by each muscle's initial force estimate. This ensured that the EMG/Force relationships were consistent across trials.

### 3.6.5 Calculation of Knee Joint Rotational Stiffness

Each muscle's contribution to knee  $JS_R$  was determined as the second derivative of potential energy ( $V$ ) of the system. The methodological development for calculating mechanical stability about any joint, using the energy approach, is outlined by Potvin & Brown (2005). For the purposes of this manuscript, it is necessary to briefly review the components of the equation that facilitate the calculation of stability, including: elastic potential energy of muscles, insertion coordinate transformation (rotation) matrix, functional moment arms, muscle stiffness ( $k$ ), three-dimensional muscle length, node length, and geometric stability.

The ability of any muscle to perform work on a joint or system of joints is dependent on force, displacement and stiffness as outlined below:

$$U_m = F_m \Delta \ell + \frac{1}{2} k_m \Delta \ell^2 \quad (1)$$

where,  $\Delta \ell$  for a small rotation is defined by the muscle's initial length  $\ell$  subtracted from its new rotated length ( $\ell_2$ ),  $F_m$  the preactivation tension developed in the muscle (N), and  $k$  the muscle stiffness (N/m). The work done by a muscle is defined by the term,  $F_m \Delta \ell$ , while the energy stored in a muscle, which acts in the same manner as a spring, is defined as  $\frac{1}{2} k_m \Delta \ell^2$ .

The stability contribution of each individual muscle acting about the knee's z-axis (flexion-extension) is calculated as the second derivative of stored potential energy, with respect to a small rotation about z:

$$S(m)_z = \frac{d^2U(m)}{d\theta^2_z} \quad (2)$$

Such that:

$$S(m)_z = F \left[ \frac{A_x B_x + A_y B_y - r^2_z}{\ell} \right] + kr^2_z \quad (3)$$

Bergmark (1989) determined muscle stiffness as

$$k = \frac{qF}{L}, \quad (4)$$

where, L is the total muscle length from origin to insertion, q the proportionality constant characterized by the relation of muscle force and length to stiffness. Substituting equation 4 into 3 yields:

$$S(m)_z = F \left[ \frac{A_x B_x + A_y B_y - r^2_z}{\ell} + \frac{qr^2_z}{L} \right] \quad (5)$$

The corrected individual muscle forces for each specified knee angle were input into the stability equation (equation 5 above) to calculate each individual muscle's instantaneous  $S(m)_z$  or  $JS_R$ . For each muscle, the moment arm ( $r_z$ ), and the origin ( $A_x$ ,  $A_y$ ,  $A_z$ ) and insertion ( $B_x$ ,  $B_y$ ,  $B_z$ ) coordinates were input into the stability equation. These data were derived through the manipulation of the biomechanical model as described in section 3.6.2. For the calculation of mechanical stability contributions about the flexion-extension axis, only  $A_x$ ,  $A_y$  and  $B_x$ ,  $B_y$  were required. The three-dimensional length of

the muscle was also calculated and input into the equation. In most cases where a node is not specified for a particular muscle at a specified knee angle, the overall muscle length ( $L$ ) will be equivalent to ( $\ell$ ). In order to determine the total knee  $JS_R$  the instantaneous individual  $JS_R$  contributions of each muscle defined in the biomechanical model were summed together.

### 3.7 STATISTICAL ANALYSIS

For each of the 18 conditions, means and standard deviations were calculated for each dependent variable across the three repeat trials. These mean values were used to represent each subjects response to that condition and were used in the subsequent statistical analysis. A  $2 \times 3 \times 3 \times 2$  mixed model Repeated Measures Analysis of Variance (ANOVA) was used to determine the influence of the within independent variables: Squat Task (stable and unstable), Leg Condition (control, unbraced, and braced), and Knee Angle ( $5-10^\circ$ ,  $20-25^\circ$ , and  $40-45^\circ$ ) and the between variable of *Injury Status* (deficient and reconstructed knee). The significance level for each ANOVA was set at  $p < 0.05$ . The dependent variables in this study included: total muscle contributions to  $JS_R$ , individual muscle contributions to  $JS_R$ , and average force/torque for each individual muscle within each specified knee range. For the significant main and interaction effects of the four independent variables, a Post Hoc analysis was performed using a Tukey's significant post hoc analysis test. Post Hoc analyses were conducted to determine where the variance in the data occurred. If no significant interactions were found between the independent variables, then any significant main effects that existed were identified and analyzed to determine the cause of the variance in the data. In each ANOVA performed,

there were three levels of interactions and four levels of main effects examined. In total ten interactions effects are possible, six 2 way interactions, three 3 way interactions and one 4 way interaction.

The standard deviations calculated across the three repeat trials within each subject were then averaged across subjects for each of the 18 conditions. These values represented the within-subject reliability for each dependent variable within each condition.

In the literature, it is widely regarded that co-activation of the hamstrings and quadriceps is the primary mechanism that is used by the neuromuscular system to enhance stability about the knee joint in ACL injured patients. According to Delp et al. (1990), the vastus lateralis and the semimembranosus have the highest maximal force capacity of the extensor and flexor muscles, respectively. Additionally, preliminary biomechanical modelling (Derouin and Potvin, 2005) has shown that the vastus lateralis and semimembranosus have the greatest potential to stabilize the knee joint in a squat posture. Thus, a qualitative comparison of hamstrings-quadriceps co-contraction was made to determine the relationship, if any, between the ratio of semimembranosus to vastus lateralis activity. This was calculated by dividing the semimembranosus %MVC by the vastus lateralis %MVC and multiplying this ratio by 100.



## CHAPTER 4

### RESULTS

The results of this study are divided into two sections. The first section describes the effect of knee angle and knee moment data on two of the four independent variables examined in this thesis, Squat Task and Leg Condition. Additionally, the effect of the *Injury Status* independent variable is discussed. The second section describes the dependent variables examined in this project, including: overall knee joint rotational stiffness, individual muscle contributions to stiffness, electromyographic and moment arm data for each of the muscles that were studied.

#### 4.1 INDEPENDENT VARIABLES

In this project the independent variable, *Injury Status* had no significant interaction or main effects for any of the dependent variables examined, that is there were no significant differences between ACL deficient subjects from ACL reconstructed subjects. Therefore, the data for each group of subjects were pooled.

##### 4.1.1 Subject Data

The two injury groups (n=17), ACL deficient and ACL reconstructed, had very similar demographic and anthropometric characteristics. The ACL deficient subjects (n=9) mean (standard deviation) anthropometric data were as follows: age 27.80 years (8.87), height 182.46 cm (3.76), mass 83.28 kg (8.14). Mean and standard deviation data for the ACL reconstructed group (n=8) was summarized as: age 25.82 years (8.76), 181.85 cm (9.71), and mass 86.75 kg (12.40). Injury and functional injury status data for each subject is presented in Appendix C.

#### 4.1.2 Knee Angle and Knee Moment Data

Table 4.1 presents the mean knee angle data for each of the 18 conditions that each subject completed. The measured mean knee angle and (standard deviation) data for the “5-10°”, “20-25°” and “40-45°” degree knee angle was 8.64° (3.68°), 20.6° (4.31°), and 33.88° (5.54°), respectively.

**Table 4.1.** Average knee angles and standard deviations. Leg ( $p<0.001$ ) and knee angle ( $p<0.0001$ ) both had significant main effects ( $n=17$ ).

Surface	Knee Angle	Control		No Brace		Brace	
		Mean	St. Dev	Mean	St. Dev	Mean	St. Dev
Stable	5-10	7.17	3.52	8.48	3.85	10.15	2.84
	20-25	18.95	3.84	20.25	3.29	22.73	4.46
	40-45	31.94	4.92	33.41	4.36	35.92	5.99
Unstable	5-10	7.65	3.84	7.92	3.62	10.21	3.65
	20-25	19.36	5.11	20.16	3.34	22.06	4.73
	40-45	32.46	5.39	33.35	5.40	36.17	6.28

A significant main effect of leg condition ( $p<0.05$ ) on knee angle was found. The mean knee angle of the brace leg condition was 3.3 degrees greater than that of the control leg. The unstable (board) conditions had more variability compared to the stable (flat) conditions. This was reflected in somewhat larger standard deviations for knee angle ranges, within each leg condition, for the board vs. flat trials. Table 4.2 displays the significant main effects of knee angle and leg condition on the measured knee moment about the flexion-extension axis. The control leg had 33.4% lower knee moments compared to the brace condition. The control leg, during the 5-10 degree knee angle range for both the stable and unstable trials, had negative (knee extensor) moments, while both the nobrace and brace conditions had positive (knee flexor) moments in the same knee flexion range.

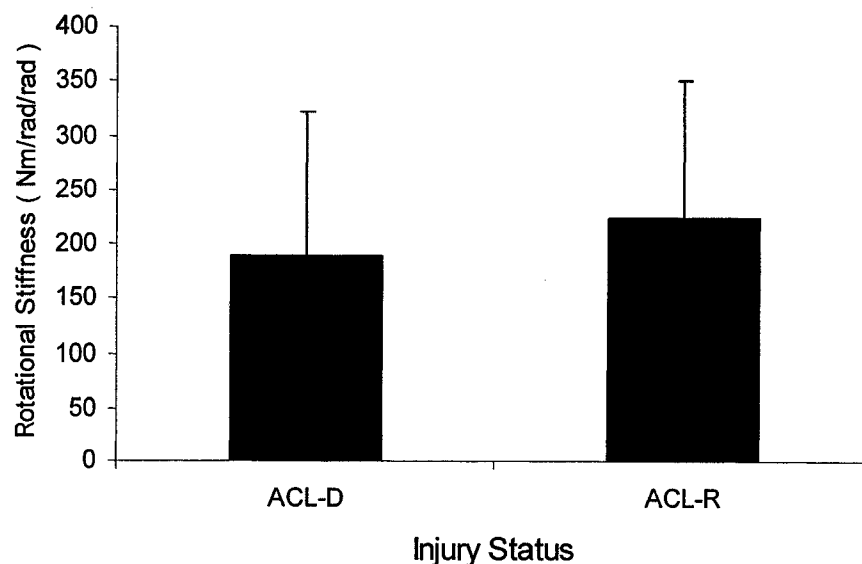
**Table 4.2.** Average knee moment (flexion-extension) and standard deviations (n=17). Leg ( $p<0.05$ ) and knee angle ( $p<0.0001$ ) both had significant main effects on knee moment.

Surface	Knee Angle	Control		No Brace		Brace	
		Mean	St. Dev	Mean	St. Dev	Mean	St. Dev
Stable	5-10	-3.62	13.63	6.94	19.51	6.96	20.13
	20-25	24.90	17.29	34.37	19.65	37.89	22.18
	40-45	53.11	23.15	67.70	23.88	70.93	28.46
Unstable	5-10	-6.21	18.34	0.63	18.69	2.86	18.75
	20-25	24.53	22.52	32.42	19.95	35.53	24.73
	40-45	58.18	23.41	66.78	25.99	72.31	29.03

## 4.2 DEPENDENT VARIABLES

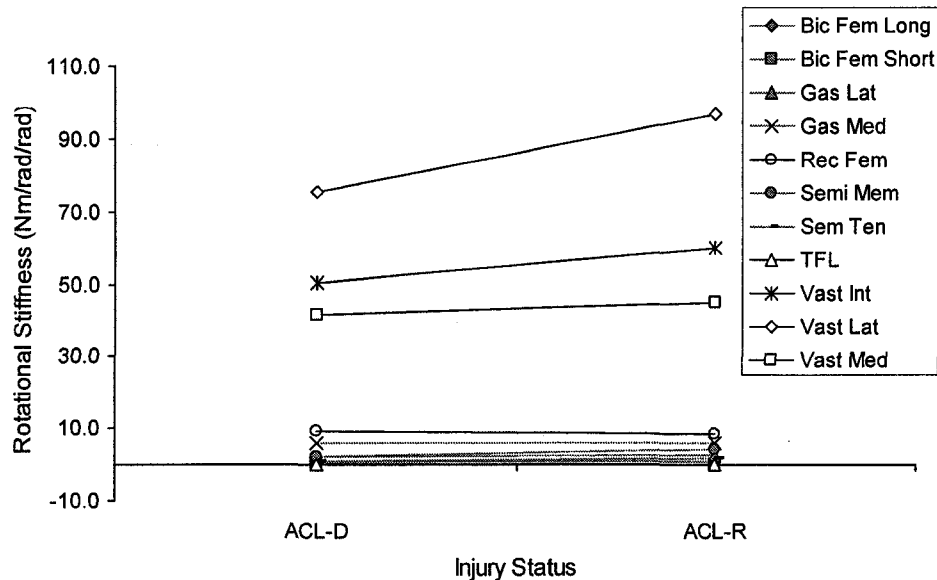
### 4.2.1 Rotational Stiffness Data - Overall and Individual Muscular Contributions

A non-significant, 18%, increase in overall  $JS_R$  was found in ACL reconstructed subjects compared to ACL deficient subjects (Figure 23). Figure 24 displays the trend of injury status on the individual muscle contributions to total  $JS_R$ . The semimembranosus and the vastus lateralis had 42% and 22.2% higher stiffness contributions, respectively, in the ACL-R subjects compared to the ACL-D subjects.



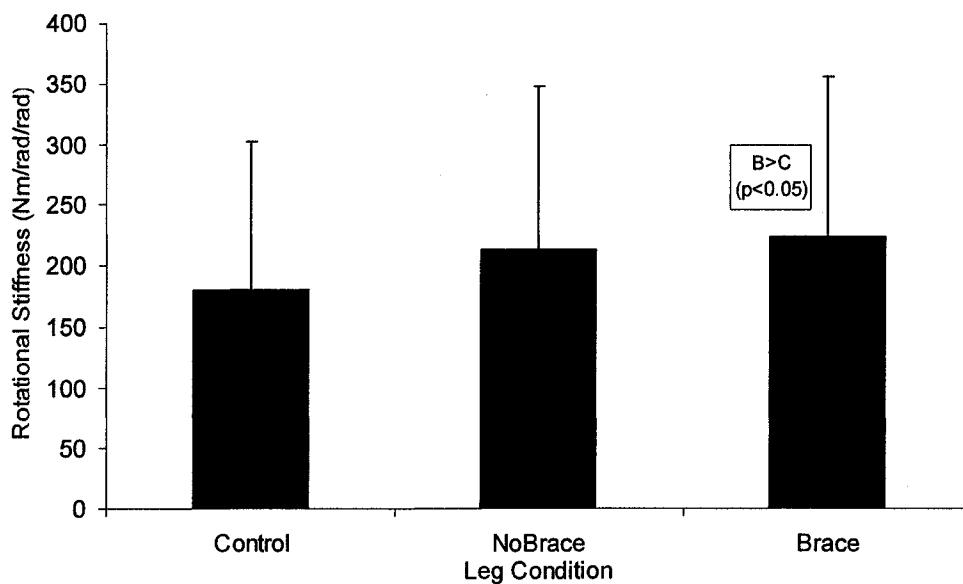
**Figure 23.** Trend of injury status on overall  $JS_R$  ( $p=0.3881$ ). ACL Deficient (n=9) and ACL Reconstructed (n=8).

A repeated measures ANOVA (Figure 25) indicated a significant main effect for leg condition ( $p < 0.05$ ) on overall  $JS_R$ . Post hoc analysis revealed that the brace condition had a significant, 19.3 %, increase in total  $JS_R$  compared to the control leg. The no brace condition demonstrated a trend towards a 15.1% (non-significant) increase in stiffness compared to the control leg.

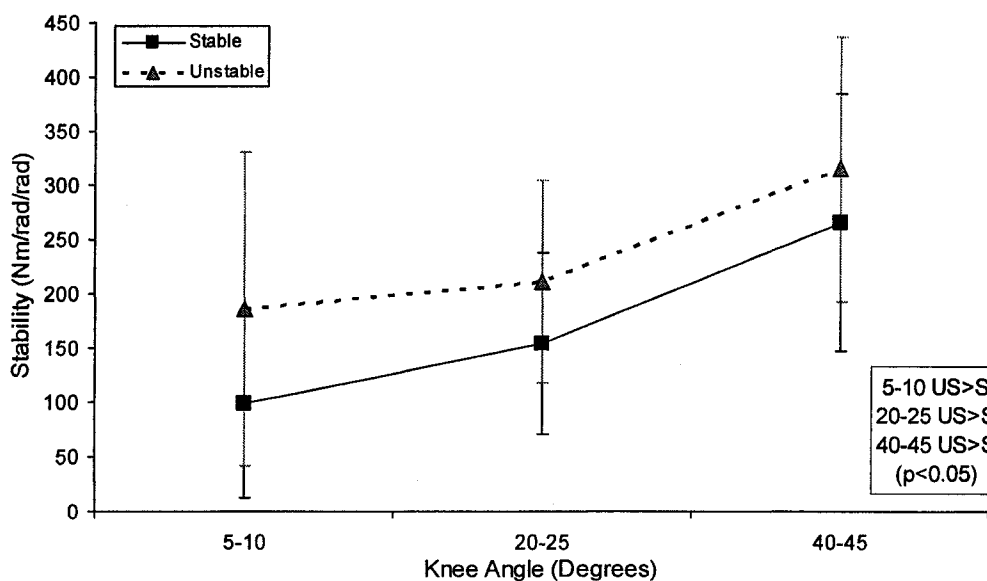


**Figure 24.** Trend of injury status on individual muscular contributions to overall stiffness ( $p > 0.05$ ). ACL Deficient ( $n=9$ ) and ACL Reconstructed ( $n=8$ ). Vastus lateralis ( $p=0.2183$ ) and semimembranosus ( $p=0.1271$ ) displayed the most pronounced trends.

Figure 26 shows the significant interaction effect of knee angle and squat task ( $p < 0.05$ ) on total  $JS_R$ . The unstable trials, collapsed across leg condition for each of the three knee angle ranges, demonstrated significantly higher overall stiffness values compared to the stable trials. Knee joint stiffness increased 40.3%, 27.1% and 15.6% in the 5-10, 20-25, 40-45 degree knee angle ranges, respectively.



**Figure 25.** Main effect of leg condition on total knee joint stiffness (n=102). Standard deviation bars are displayed. Post hoc results are presented.



**Figure 26.** Interaction effect between squat task (stable vs. unstable) and knee angle (n=51). Standard deviation bars are displayed. Post hoc results are presented.

A summary of the significant interaction and main effect data is provided in Table 4.1 for the overall knee stiffness and the individual muscle contributions to knee stiffness.

Significant interactions between knee angle and squat task were found for overall knee stiffness and all the muscles that cross the knee joint, with the exception of vastus medialis and tensor fascia latae. A main effect for leg condition was found for overall knee stiffness and some of the flexor muscles of the knee: gastrocnemius medialis and lateralis, and both heads of the biceps femoris. Rectus femoris, a two joint muscle, was the only muscle that did not demonstrate a significant main effect of knee angle on knee joint stiffness. The biceps femoris long and short heads were the only group of muscles that did not show a main effect between the stable and unstable trials. Post hoc analysis data are presented in Table 4.2. The biceps femoris long and short heads had negative (destabilizing) contributions to knee joint stiffness for both the unstable and stable trials for the 5-10 degree knee angle range. The unstable conditions had significantly more negative, destabilizing, knee stiffness values. The control leg condition of the gastrocnemius medialis had significantly higher knee stiffness contributions than either the brace or no brace conditions.

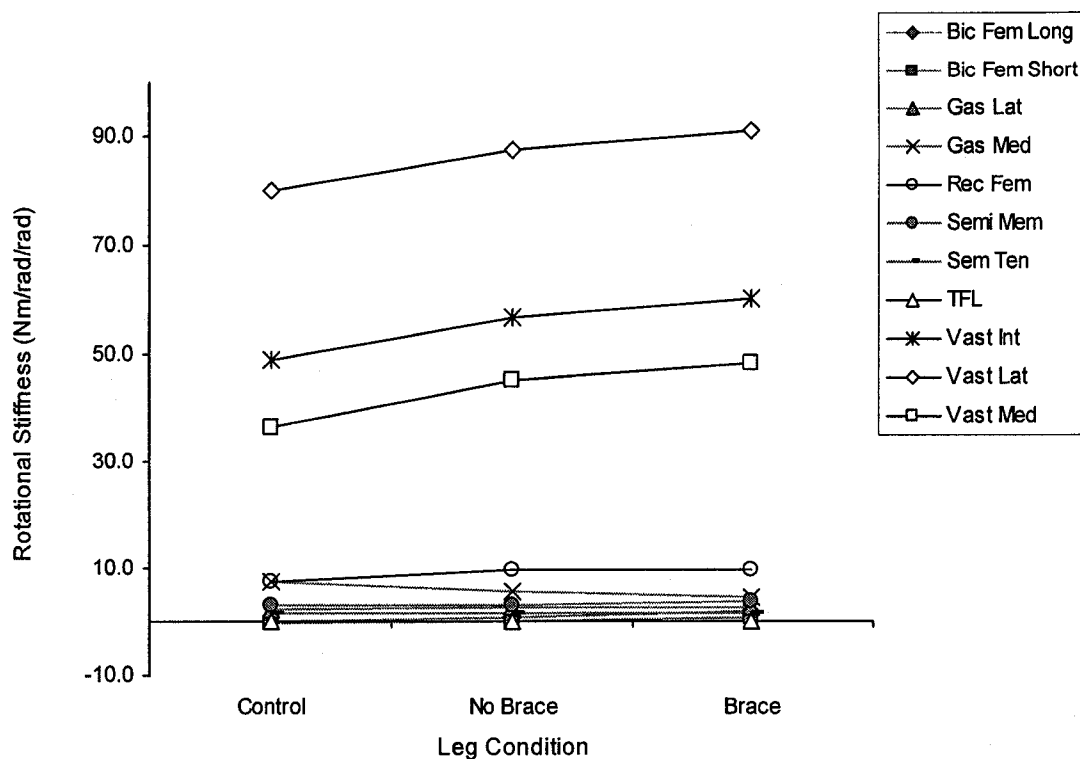
**Table 4.3.** Significant interaction and main effects ( $p < 0.05$ ) presented for overall stiffness and the individual muscular contributions to stiffness

	Interaction Effects	Main Effects		
	( $p < 0.05$ )	( $p < 0.05$ )		
	ST*KA	Leg (L)	Angle (KA)	Squat Task (ST)
<b>Overall JS<sub>R</sub></b>	0.0459	0.0449	<.0001	<.0001
Vastus Lateralis	0.0064		0.0004	<.0001
Vastus Intermedius	0.0329		<.0001	<.0001
Vastus Medialis			<.0001	0.0001
Rectus Femoris	0.043			0.012
Gastrocnemius Medialis	0.0257	<.0001	<.0001	<.0001
Gastrocnemius Lateralis			0.0145	0.0206
Semimembranosus	<.0001		<.0001	<.0001
Semitendinosus	<.0001		<.0001	<.0001
Biceps Fem L	<.0001	0.0061	<.0001	
Biceps Fem S	<.0001	0.0073	<.0001	
Tensor Fascia Latae			<.0001	0.0483
Post-hocs performed on highlighted cells				

Figure 27 shows the effect of leg condition on the individual muscle contributions to knee stiffness. The vasti muscles displayed a small and non-significant increase in stiffness between the brace-no brace conditions and the no brace-control conditions. The vastus lateralis stiffness contribution was 12.7 % greater in the brace conditions compared to the control leg conditions. Only the biceps femoris had significantly higher stiffness values for the brace condition, compared to the control leg.

**Table 4.4.** Post hoc results for the significant interaction and main effects for overall stiffness and individual muscular contributions to knee joint stiffness.

	Interaction Effects			Main Effects		
	Squat Task-Stable vs.Unstable			(p<0.05)		
	5-10	20-25	40-45	Leg (L)	Angle (KA)	Squat Task (ST)
<b>Overall JS<sub>R</sub></b>	US>S	US>S	US>S	B>C		
Vastus Lateralis	US>S	US>S	US>S			
Vastus Intermedius	US>S	US>S	US>S			
Vastus Medialis					40>5;40>20	US>S
Rectus Femoris	US>S					
Gastrocnemius Medialis	US>S	US>S	US>S	C>B;C>NB		
Gastrocnemius Lateralis						US>S
Semimembranous		US>S	US>S			
Semitendinosous	US>S	US>S	US>S			
Biceps Fem L	S>US		US>S	B>C		
Biceps Fem S	S>US		US>S	B>C		
Tensor Fascia Latae					5>20;5>40	Post hoc ns
US = Unstable; S = Stable / B = Brace; C = Control; NB = No Brace						

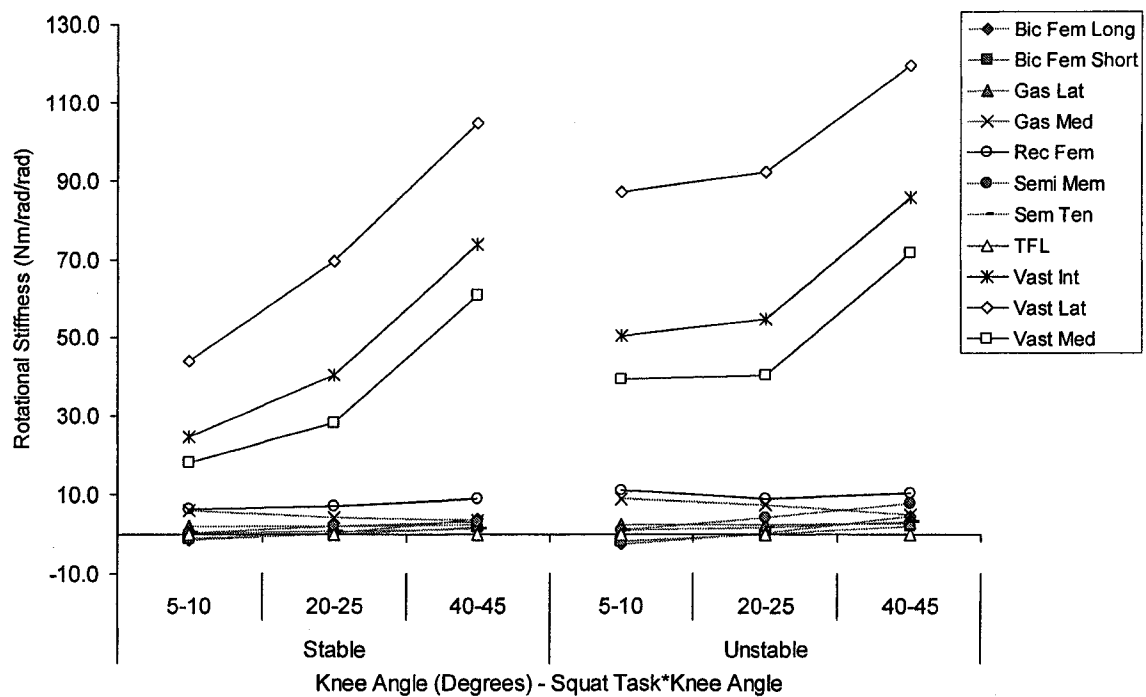


**Figure 27.** The effect of leg condition on individual muscle contributions to overall knee joint stiffness (n=102).

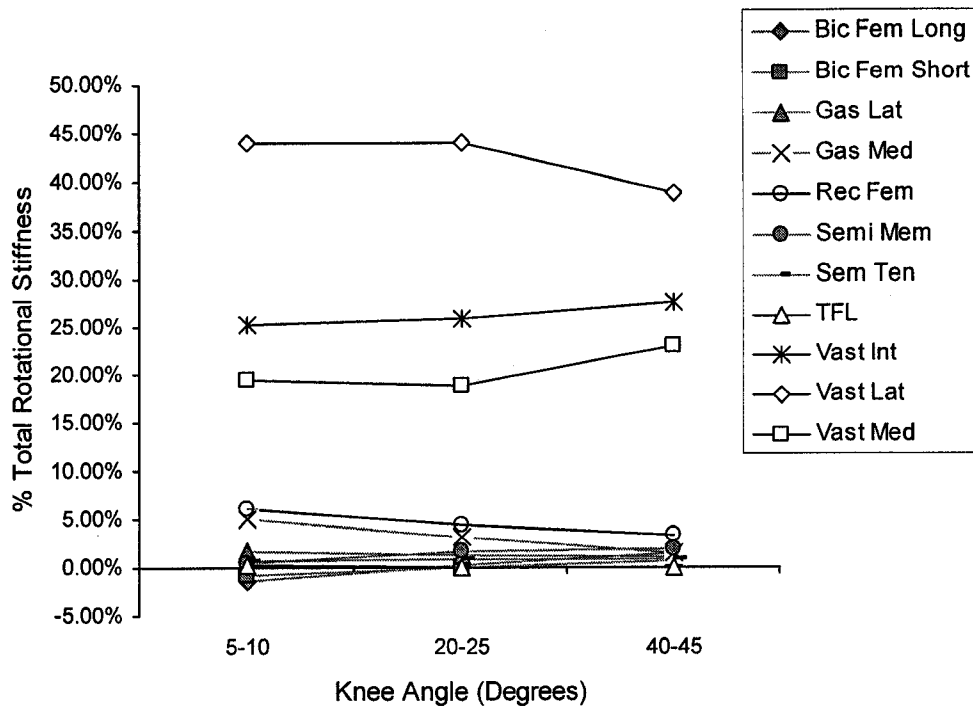
Figure 28 shows the significant interaction effects, between knee angle and squat task for the 11 muscles included in this analysis of knee joint stiffness. Again, the post hoc results are summarized in Table 4.2. It is apparent that the vasti muscles play the most significant role in maintaining joint stiffness. Additionally, these muscles appear to be the most sensitive to changes in joint angle and squat task. For the vastus lateralis, the 5-10 degree knee angle range represented the largest increase in stiffness between the stable and unstable trials (49.5%), while in the 20-25, and 40-45 degree knee angle range stiffness increased 24.5% and 12.3%, respectively. The relative contribution to stiffness by each muscle is depicted in Figure 29, as a percentage of total stiffness. The most important knee stabilizers, presented in ranked order as an average percentage of overall stiffness were: 1) vastus lateralis: 42.2%, 2) vastus intermedius: 26.2%, 3) vastus medialis: 20.3%, 4) rectus femoris: 4.6%, 5) gastrocnemius medialis: 3.3%, 6)



semimembranosus: 1.4%. The vastus lateralis and vastus medialis appeared to be the most sensitive to changes in knee angle. The vastus lateralis contribution decreased from the 20-25 degree to the 40-45 degree knee angle range by 5.6%, while the vastus medialis increased 3.9%. The gastrocnemius medialis stiffness contribution decreased 3.5% from the 5-10 degree to the 40-45 degree knee angle ranges. The semimembranosus contribution increased 1.5%, from 0.5% in the 5-10 degree range to 2.0% in the 40-45 degree knee angle range.

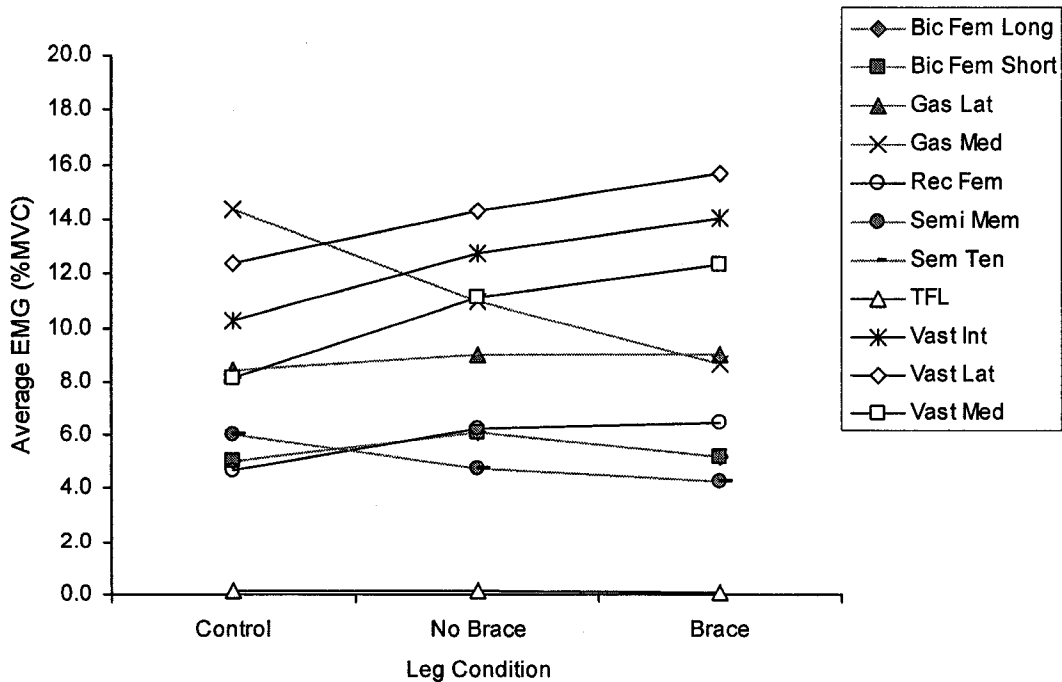


**Figure 28.** Interaction effect of knee angle and squat task on the absolute individual muscle contributions to knee joint stiffness (n = 51).



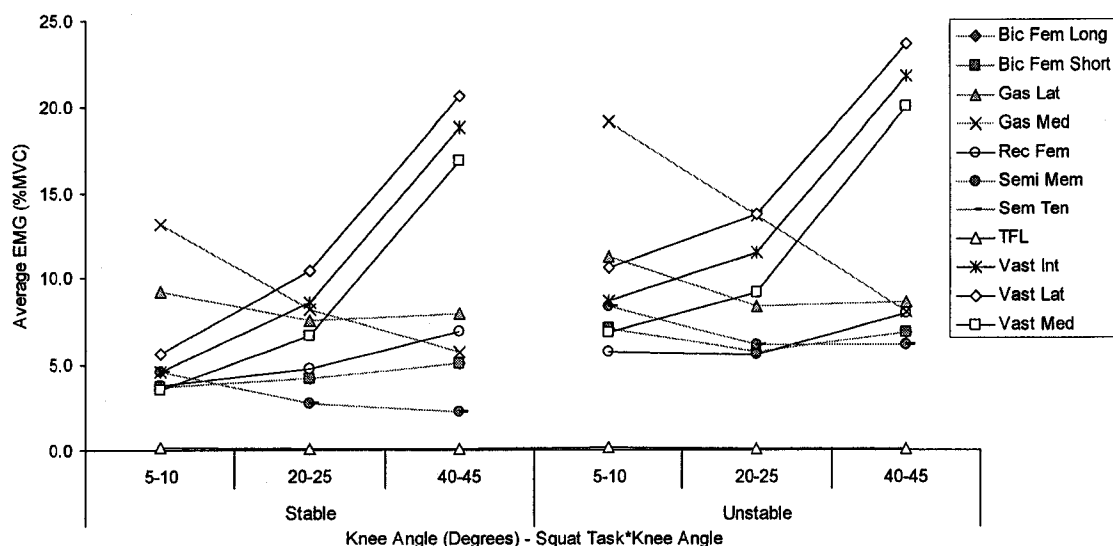
**Figure 29.** The effect of knee angle on the individual muscle contributions to knee joint stiffness, presented as a percentage of overall stiffness (n=102).

4.2.2 Rotational Stiffness Inputs- EMG and Moment Arm Data



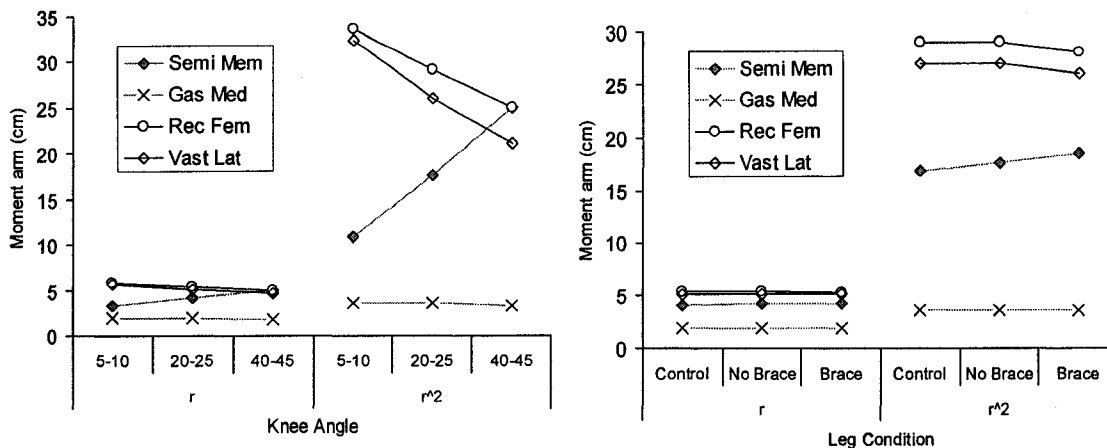
**Figure 30.** Significant main effect of leg condition on average EMG of the individual muscles that were used as inputs to calculate knee stiffness (n=102).

In Figure 30, the effect of leg condition on mean average EMG is presented for each of the 11 muscles included in the stiffness model. A significant main effect of muscle activation on leg condition was found for the gastrocnemius medialis ( $p < 0.0001$ ), vastus medialis ( $p < 0.05$ ) and intermedius ( $p < 0.05$ ). The gastrocnemius medialis displayed the largest changes in activation as a percentage of maximal voluntary contraction, between leg conditions. Post hoc analysis on gastrocnemius medialis muscle activation, revealed the following: the control leg had higher EMG than both the nobrace and brace conditions, and the nobrace condition had greater activation than the brace condition. The vastus medialis EMG for the brace condition was significantly higher than the control leg. As well, vastus intermedius activation for the brace condition was significantly higher for the brace condition compared to the control leg. Figure 31 shows the interaction of knee angle and squat task on muscular activation levels for each muscle. The gastrocnemius medialis had a significant interaction between knee angle and squat task ( $p = 0.0049$ ). The unstable trials, for each knee angle range, had significantly higher EMG than the stable trials. Significant main effects for knee angle and squat task were found for all muscles. These results are summarized in Appendix D.

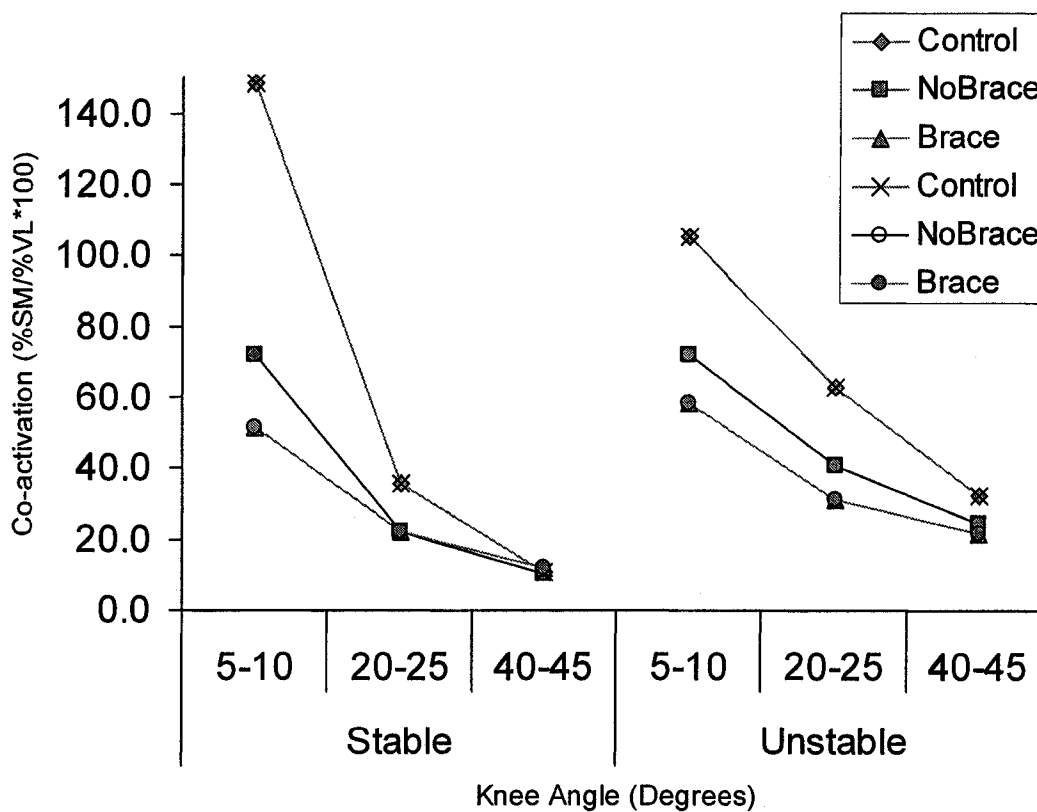


**Figure 31.** Interaction effect of knee angle and squat task on the average EMG of individual muscles that were used as inputs to calculate knee stiffness (n=51).

Figure 32 shows the effect of knee angle and leg condition on the flexion-extension functional moment arm and the squared moment arm values. The squared functional moment arm values were used as inputs into the stability equation to determine each muscle's individual contribution to stiffness. The effect of knee angle on the squared moment arm value for the vasti, rectus femoris and semimembranosus is much more pronounced compared to the effect of leg condition. These results are summarized in Appendix D.



**Figure 32.** A) Mean moment arm data showing effect of knee angle on moment arm and  $r^2$  value used to calculate knee stiffness (n=102) B) Effect of leg condition on moment arm and  $r^2$  value used to calculate knee stiffness (n=102).



**Figure 33.** Mean effect of leg condition, squat task, and knee angle on coactivation of the semimembranosus relative to the vastus lateralis (n=17).

Figure 33 depicts the co-activation ratio of the semimembranosus EMG activity relative to vastus lateralis activation. The 5-10 degree knee angle range, for both the stable and unstable trials, had the highest overall co-activation ratios for each leg condition. At this knee angle range the activity of the vastus lateralis was at its lowest level, while the EMG activation of the semimembranosus was at its highest level. This pattern of activation reversed as the knee flexion range was increased to 40-45 degrees.

## CHAPTER 5

### DISCUSSION

The objective of this study was to ascertain the influence of functional knee braces on the knee joint rotational stiffness of ACL deficient and reconstructed subjects. Seventeen unilaterally ACL injured subjects (9 ACL-D, 8 ACL-R) attempted to maintain static postures during stable and unstable squat trials at three knee angle ranges for each leg condition. Average kinematic, force plate and EMG data were integrated into a linked-segment biomechanical model of the leg, to ultimately determine the individual contribution to joint stiffness of each muscle that spans the knee joint. The sum of these individual muscle stiffness values, for each condition, was used to determine the influence of several independent variables on overall joint rotational stiffness ( $JS_R$ ) about the flexion/extension axis of the knee. The independent variables examined in this project, included: *Squat Task*, *Leg Condition*, *Knee Angle*, and the between variable of *Injury Status*.

The most important significant findings from this study are summarized as follows: 1) In the braced leg conditions, subjects displayed significantly higher overall  $JS_R$  compared to the control leg, despite a significant decrease in muscle activation of medial head of the gastrocnemius, the most important contributor to  $JS_R$  of all the knee flexor muscles. 2) A significant interaction was found between knee angle and squat task. For each knee angle range, the unstable squat trials had significantly higher  $JS_R$  values than the stable squat trials. However, the difference in the  $JS_R$  values between the stable and unstable squat tasks at the 5-10° was 1.5 greater than at 20-25° and 2.8 times greater than at 40-45°. This larger increase in stiffness at the most extended knee posture

suggest that the knee is more vulnerable to buckling at 5-10°, particularly if an unexpected perturbation is applied to the lower extremity during a squat performed on a stable surface. 3) The influence of *Squat task* (unstable vs. stable) on JS<sub>R</sub> was significant for nine of the 11 muscles included in this joint stiffness analysis. This indicated that a functional training exercise, like that performed on a wobble board, challenges the neuromuscular system sufficiently to augment muscle activity of the knee joint musculature, thus, increasing overall stiffness of the knee.

### 5.1 KNEE ANGLE EFFECTS

Prior to each trial, subjects were verbally informed to maintain knee joint angle for the duration of the trial within one of the three ranges, 5-10°, 20-25°, or 40-45°. The average measured knee angle for the most extended knee position ranged from 7.2° to 10.2°, well within 5-10°. For the 20-25° knee joint range, the measured angles ranged from 19.0° to 22.7°, slightly biased towards an extended knee position. In the most flexed knee posture, the calculated knee joint angles from the kinematic data differed substantially from the 40-45° angle range (Table 4.1). In the 40-45° knee angle range, the calculated average knee angles ranged from 31.9° to 36.2°. The measured values for the 40-45° angle range were consistently biased towards an extended knee posture. The electrogoniometer provided invaluable biofeedback to the subjects throughout each trial. Although, the measured knee angle in this range was substantially smaller by 8-9°, the biofeedback allowed subjects to consistently maintain joint angles within a repeatable range of 32 to 36 degrees. This consistency is evident from the comparable variability (standard deviations) for all of the knee joint angle ranges. The manufacturer of the



electrogoniometer (Biometrics, Ltd.) reports a measurement accuracy of  $\pm 1^\circ$  over a range of  $90^\circ$ . The disparity from the  $40\text{-}45^\circ$  knee angle range may be attributed to the placement of the electrogoniometer and the roll-glide characteristics of the tibiofemoral joint. The positioning of the electrogoniometer on the posterior aspect of the knee was required due to the shell construction (Figure 9A) of the FLEX functional knee brace (Figure 19). Unanticipated soft tissue deflection of the hamstring and gastrocnemius muscles, associated with increasing knee flexion angles, most likely contributed to the systematically smaller knee flexion values in the  $40\text{-}45^\circ$  range. The forward gliding and backward rolling motion of the femur that occurs as the knee is flexed may have also influenced the accuracy of the measured knee joint angles. Placement of a reflective anatomical marker on the lateral condyle of the femur does not account for the changing axis of rotation of the tibiofemoral joint that occurs with knee flexion and extension. Two additional estimates of the knee joints instantaneous center of rotation were provided by two cluster markers sets, positioned on the leg and thigh. An average x and y coordinate was determined throughout the entire 10 second trial for each condition. This was done to facilitate a more accurate representation of the subjects' knee angle throughout the course of the experiment. Overall, the unstable trials performed on the wobble board appeared to have slightly more variability compared to the stable trials. This reflects the increased neuromuscular demand placed on the knee joint as subjects attempted to maintain knee joint angle throughout these unstable trials. The larger measured knee angle for the brace condition compared to the control leg reflects the difficulties in determining knee joint position from the methods described above.

As expected, the average net knee moment increased as the knee was flexed from an almost fully extended position to approximately 34°. In the unbraced and braced leg conditions, for both the stable and unstable trials completed at 5-10°, subjects maintained positive (flexor) knee moments (Table 4.2). However, subjects maintained negative (extension) knee moments for both squat tasks completed with the control leg in the same angle range. This finding does not support the quadriceps avoidance strategy described by Berchuck *et al.*, (1990) and Devita *et al.*, (1998).

The study completed by Berchuck *et al.* (1990) found that unilateral ACL deficient subjects, compared to normal subjects, had significantly different kinematic and kinetic patterns in the knee and hip joints during level walking and jogging. During level walking, injured subjects demonstrated significantly greater maximum flexion moments about the hip in the early part of stance phase compared to the control subjects. A *quadriceps avoidance pattern* was adopted by 75% of the deficient subjects. Normally, at midstance there is an external knee flexion moment, which is counteracted by a net internal extension moment generated by the quadriceps. The subjects in this study did not display this *characteristic* external knee flexion moment, but instead exhibited an external extension moment produced by hamstrings contraction during midstance. Devita *et al.* (1998) reported similar findings in ACL reconstructed patients compared to normal subjects during level walking. After the completion of an accelerated rehabilitation program, patients continued to show a marked reduction in the knee extensor torque and excessive torques at the hip joint during the first half of stance phase, compared to the healthy subjects. Although gait was not analyzed in the current study, investigation of the relationship between the external moments and the opposing net muscle moments

discussed in these studies may provide insight into the mechanisms used by subjects in the current experiment to stabilize the knee. In the present study, during both the unbraced and braced leg trials completed at near full knee extension, subjects maintained an internally generated quadriceps extensor torque to overcome the small external flexor moment acting at the knee. Subjects maintained this external knee flexor moment despite the increased neuromuscular demand placed on the lower extremity during the unstable squat task. Perhaps the neuromuscular system attempts to minimize the vulnerability to subluxation when the knee joint is in its most extended position. Avoidance of an external knee extensor moment in the injured leg trials performed at 5-10° may eliminate the need for additional and perhaps unexpected quadriceps activity required to overcome internally generated knee flexor torques by the gastrocnemius and hamstrings muscles (Figure 28). Invariably, the neuromuscular system chooses to minimize the potential shear force generated by the gastrocnemius muscles rather than utilize a quadriceps avoidance strategy.

A surprising finding was the non-significant differences in the average knee moments generated in the stable squat trials compared to the unstable trials. Additionally, subjects demonstrated very similar fluctuations (standard deviations) around the same knee moment values measured for each of the three knee angle ranges.

## **5.2 INJURY STATUS (ACL DEFICIENT VS. RECONSTRUCTED) EFFECTS**

The effect of *Injury Status* on overall JS<sub>R</sub> was primarily due to the increased activation of the vastus muscles in the ACL reconstructed subjects, compared to the deficient subjects. Although the semimembranosus EMG amplitude had a relative

increase of 42% in the surgically treated group, compared to deficient group, this represented a non-significant increase. The ACL-D group had a mean EMG of 2.3% MVC while the ACL-R group was 4.4% MVC. This finding of increased quadriceps activity in the ACL reconstructed subjects is in line with the results presented by Suter *et al.* (2001). In that study, an interpolated twitch technique revealed a significant difference in quadriceps inhibition during a series of isometric knee extension exercises completed at 30° of knee flexion. The ACL deficient group had significantly greater muscle inhibition compared to the reconstructed group. In the current study, the highest average muscle activity was recorded from the vastus lateralis in the unstable squat at 40-45°. The average mean EMG recorded from the vastus lateralis did not exceed 30 %MVC. The lack of significant main effect on overall JS<sub>R</sub> between the ACL-D and ACL-R groups may be due to the relatively low muscle activity elicited during the two isometric squat tasks. During more dynamic exercises, in healthy subjects with uninjured knees, Graham *et al.* (1993) reported much higher coactivation of the quadriceps and hamstrings, compared to the present study. For example, slideboard exercises elicited quadriceps and hamstrings activities of 55.8 and 41.3 %MVC, respectively, while step-up exercises produced 40.9 and 25.0 %MVC. The quarter squat exercise performed in that study produced quadriceps activation of 25.9% MVC, which is similar to that generated during the unstable squat completed in the current study, at 40-45°. However, Graham *et al.* (1993) found substantially higher hamstrings activity of 15.9 %MVC during this exercise compared to the current 8.3% from the unstable trials performed at 5-10° of knee flexion.

### 5.3 LEG CONDITION (BRACED, UNBRACED, CONTROL) EFFECTS

The significant increase in overall  $JS_R$  in the braced *Leg Condition*, compared to the control leg (Figure 23), indicates that functional knee braces may provide a protective effect against buckling, by facilitating and augmentation of the activation patterns of the muscles that cross the knee joint. This finding illustrates the importance of considering the entire compliment of muscles that span the knee joint when discussing rotational joint stiffness. Despite the fact that nearly all of the major knee stabilizers did not show a significant main effect of *Leg Condition*, the overall stiffness contribution of the muscles, taken as a group, increased significantly. Additionally, the medial head of the gastrocnemius, the most important knee stabilizer of the flexor muscles, had a significantly lower stiffness contribution in the braced condition compared to the control leg. The lack of significant differences, between braced versus unbraced and control versus unbraced, is most likely due to the quasi-static nature of the both of the squat tasks completed during this investigation. In the literature, several studies (Smith *et al.*, 2003; Branch *et al.*, 1989; Diaz *et al.*, 1997; Nemeth *et al.*, 1997; Acierno *et al.*, 1995) have examined the influence of functional knee braces on stabilizing the knee during dynamic activities, such as skiing, walking, running, and isokinetic exercise. In each of these studies, bracing the ACL injured knee was shown to significantly alter the electromyographic activity of one, or more, of muscles that cross the knee joint. However, none of these studies, or any other in the literature to date, have defined and determined the stiffness of the knee joint from a mechanical perspective. Therefore, one may only speculate about the potential stability that functional knee braces might provide during more demanding and dynamic activities that elicit near maximal muscle activity.

Despite the methodological limitations and challenges of having subjects perform a quasi-static task, functional knee braces were shown to be associated with a significant enhancement of the total rotational stiffness of the knee joint of the injured leg, compared to the control leg. Additionally, a few studies (Kirkendall *et al.* 2002; Grabiner *et al.* (1993).; Limbird *et al.*, 1998) have shown significant differences in muscle activity of either ACL injured versus 1) the non-involved leg or, 2) compared to a control group of normal subjects. Limbird *et al.* (1988) showed that injured subjects had significantly different levels and patterns of muscle activity during free and fast walking speeds compared to normal subjects. Grabiner *et al.* (1993) found that ACL injured subjects had significantly higher levels of hamstrings co-contraction when compared to normals during isometric knee extension contractions performed at maximum and pre-selected percentages of maximum. Again, these findings indicate that the isometric squat tasks completed during this study may not have sufficiently challenged the neuromuscular system to produce significant differences in  $JS_R$  between the unbraced condition and both the braced and control leg conditions.

#### **5.4 SQUAT TASK (STABLE VERSUS UNSTABLE) EFFECTS**

As expected, the trials that were completed on the custom-made wobble board had significantly higher total knee  $JS_R$  values than those trials performed on the flat surface. In fact, eight of the eleven muscles examined in this thesis (Table 4.3) had significant main effects for *Squat Task* (stable vs. unstable). This finding demonstrates the increased neuromuscular demand required to maintain pre-selected knee angles during an unstable isometric single leg squat, compared to a stable single leg squat. In

the most extended position, the knee appears to be most vulnerable to buckling (Figure 26 and 28). During simulated isolated contraction of the quadriceps, Grabiner *et al.* (1989) determined that maximum anterior drawer force occurs at 5° of knee flexion and was calculated to be 14% of the tension developed in the quadriceps. In the present study, the total stiffness increase in the stable squat task compared to the unstable squat task in the 5-10° range, was nearly 1.5 and 2.8 times greater compared to the 20-25° and 40-45° knee angle ranges, respectively. In the stable trials at 5-10°, without the threat a potential destabilizing perturbation, the neuromuscular system invariably relies on the intact passive structures (articular cartilage, joint and skeletal geometry) to preserve joint stability and, therefore, does not require substantial JS<sub>R</sub> to maintain stability of the knee. However, during the unstable trials at the same knee angle range, the body systematically increases electromyographic activity of the knee musculature to enhance stiffness and minimize the risk of buckling. It is postulated that joint mechanoreceptors that affect muscle stiffness mediate tuning of the muscle spindle system to enhance joint stiffness (Johansson *et al.*, 1991). The unstable single leg squat tasks, performed on the wobble board, required the neuromuscular system to act in a coordinated fashion in order to maintain the knee joint at the pre-selected angle. Chmielewski *et al.* (2002) found that ACL injured subjects, that completed a progressive perturbation training program, developed coordinated and compensatory muscle activation patterns during self-paced walking trials. Subjects demonstrated coupled hamstring and soleus activation, along with vastus lateralis activity, in the post training trials. Additionally, the vastus lateralis had a significantly longer integral of muscle activity following the perturbation training. In the present study, it was hypothesized that the trials performed on the unstable surface

simultaneously evoked reflex activation and inhibition of the antagonistic muscle pairs that cross the knee joint. Nichols (1994) found, in experiments performed on decerebrate cats, that muscles simultaneously react to perturbations applied to the joint. Muscles that oppose the perturbation are stretched and their activation levels are reflexively increased, while muscles that act in the same direction of the perturbation are inhibited to minimize unwanted motion. Chmielewski *et al.* (2002) suggests that the net result of this coordinated co-activation is to stiffen the joint and maintain dynamic stability.

## **5.5 INDIVIDUAL MUSCLE CONTRIBUTIONS TO OVERALL JOINT STIFFNESS**

As presented previously, a muscle's stabilizing potential is determined by the following parameters: muscle force (related to activation level), functional moment arm, muscle length, muscle stiffness and the geometric orientation of the muscle relative to the knee joint. A discussion of these parameters, for each of the major stabilizers of the extensor and flexor compartments, is necessary to understand how their stiffness contributions are influenced by the four independent variables.

### **5.5.1 Knee Extensor Compartment**

The quadriceps femoris muscles provided, on average, over 90% of the  $JS_R$  calculated for each of the three knee angle ranges (Figure 29). The vastus lateralis' contribution to overall joint stiffness was the largest among all the quadriceps muscles, primarily due to the fact that it has the highest maximal force capacity and demonstrated the highest activation (Table AD-1 Appendix D). Additionally, it has a somewhat smaller resting length and slightly larger functional moment arm compared to the other vasti



muscles. As noted in equation 5, muscles with shorter lengths and larger functional moment arms have a greater potential to stabilize the knee. In general, the quadriceps muscles had significantly higher stiffness contributions during the unstable trials, compared to the stable trials, in each of the knee angle ranges (Table 4.4).

The increase in muscle activity between 5-10° and 20-25° appears to be much smaller in the unstable, compared to the stable squat trials (Figure 31). This, again, reflects the increased neuromuscular demand required to stabilize the knee joint in an extended posture. Numerous studies (Bodor, M., (2001); Ciccoti *et al.*, 1994; and Suter *et al.*, 2001) in the literature summarize the importance of the quadriceps in stabilizing the knee joint. Ciccoti *et al.* (1994) evaluated rehabilitated ACL deficient, ACL reconstructed, and normal subjects during several diverse tasks, such as walking on flat and ramped surfaces, ascending and descending stairs, running and cross-cutting, to determine differences in the electromyographic activity of the thigh and lower leg muscles. The significantly increased activity in the vastus lateralis of the rehabilitated patients accounted for sixty seven percent of the differences in the quadriceps EMG. They suggest that the force vector of the vastus lateralis may facilitate a protective mechanism, by acting to resist internal tibial rotation and lessen, or perhaps prevent, the pivoting action that may be evoked with ACL deficiency.

In the present study, the vastus lateralis had the largest contributions to JS<sub>R</sub> and the highest average EMG across all conditions. However, no significant differences were found in the knee joint rotational stiffness and average EMG values between the ACL deficient and reconstructed groups. The essential knee stabilizing role of the quadriceps, specifically the vastus lateralis, may be explained by the difference in the orientation of

its vector in closed versus open kinetic chain exercises. Bodor (2001), in a very simple experiment, demonstrated that the quadriceps vector is directed superiorly during open kinetic chain, and inferiorly during closed kinetic chain, movements performed near 35° of knee flexion. An ultrasound transducer placed over the vastus lateralis revealed that, during open kinetic chain extension (subject seated), the vastus lateralis muscle fibres moved superiorly with respect to the femur, indicating a superior orientation of the quadriceps. During closed kinetic chain knee extension (subject standing), the vastus lateralis muscle fibers appear to move neither superiorly or inferiorly, indicating that the muscle fibers are moving together with the femur.

Bodor (2001) suggest that, in order for the femur to move anteriorly on its axis of rotation, an anterior quadriceps vector has to exist and this may only be satisfied if the quadriceps vector is directed in an inferior direction. “Therefore the quadriceps vector  $Q$  for closed kinetic chain knee extension originates from the femur, is directed inferiorly, has anterior  $Q_a$  and inferior  $Q_b$  components.” The shear component of the quadriceps vector in the closed kinetic chain movement leads to reduced anterior tibial femoral shear force. Bodor (2001) concluded that the quadriceps protect the ACL during closed kinetic chain activities, regardless of the activity of the hamstrings. Both of the closed kinetic chain single leg squat tasks, performed during the present study, support the potential protective function and stabilizing role of the quadriceps. A trend of higher muscle activation for all the quadriceps muscles was found (Figure 30). The unbraced leg had greater quadriceps activity compared to the control leg condition, and the braced legs had higher quadriceps activity compared to the unbraced legs. Additionally, the vastus medialis had significantly higher activity in the braced versus control condition. This

suggests that functional knee braces possibly elicit greater quadriceps activity. Greater quadriceps activation may be required in order to maintain knee joint position at the specified knee angle range and overcome the posterior tibial femoral shear force ( $R_5$  – Figure 10) generated by the 4 point pressure system of the brace.

### 5.5.2 Knee Flexor Compartment

Overall, the contribution of the hamstrings and gastrocnemius muscles to total  $JS_R$  was nearly insignificant compared to the stiffness contributions provided by the quadriceps. The medial head of the gastrocnemius provided the greatest contribution to  $JS_R$  (3% of overall stiffness) of all the muscles of the knee flexor compartment, while the semimembranosus had the highest stiffness contribution (1.3 % of overall stiffness) of all the hamstring muscles (Figure 27). The low level of hamstring activation was primarily responsible for the very small stiffness values for each of the four hamstring muscles. Additionally, the maximum force capacity of the semimembranosus is approximately 45% lower, and its functional moment arm values are much smaller, than that of the vastus lateralis (Appendix D). Several investigators (Isear *et al.*, 1996; Kvist *et al.*, 2001; and Grabiner *et al.*, 1989) have reported low levels of hamstring activity during various activities, including: unloaded squats, isometric knee extension and active knee extension. Isear *et al.* (1996) reported hamstring activity of 4-12% of maximum voluntary isometric contraction during a controlled dynamic unloaded squat performed by normal subjects from full knee extension to 90° of knee flexion. This was similar to the low semimembranosus activity (2.0-10.5 MVC%, Appendix D) found in the present study. Kvist *et al.* (2001) also found low levels of hamstring electromyography (3-9 %MVC)

during several variations of closed kinetic chain squats and an open kinetic chain exercise in ACL deficient and normal subjects. Additionally, Grabiner *et al.* (1989) revealed minimal biceps femoris activation (9.5-13.6 %MVC) in ACL deficient patients during isometric knee extensions performed at a knee angle between 10-20°. Subjects were evaluated in 10% intervals of maximum isometric knee extension from 10-100% MVC. Surprisingly, muscle activity in the biceps femoris only increased 3.4 %MVC compared to a 71-83 %MVC greater increase in quadriceps EMG and a 90 %MVC increase in knee extensor torque.

Those studies, along with the findings in the present investigation, contradict the role of hamstrings co-contraction in stabilizing the knee joint. (Figure 33). Knee joint stiffness appears to have an almost inverse relationship with co-activation of the hamstrings relative to the quadriceps, if one compares the patterns between the rotational stiffness (Figure 26) and co-contraction graphs (Figure 31). The quasi-static nature of the squats tasks, and the relatively upright trunk position maintained by subjects in the present experiment, may not have required significant hamstrings recruitment. As discussed previously, the hamstrings muscles function to decelerate the thigh and shank at terminal swing phase during walking, running and cutting movements. Bodor (2001) argues that the role of the hamstrings in stabilizing the knee, and counteracting anterior tibial-femoral shear during closed kinetic chain scenarios, is secondary to its primary function as a static and dynamic stabilizer of the trunk and pelvis. In the current study, the gastrocnemius muscles played a more significant role than the hamstrings in stabilizing the knee. Kvist *et al.*, (2001) concluded that the “simultaneous activation of the quadriceps and the gastrocnemius muscles seems to represent an important

mechanism for stabilization of the unstable knee in contrast to hamstring muscle activation”.

The medial head of the gastrocnemius had the highest contributions of all the flexor muscles and appeared to be the most sensitive to changes in leg condition and joint angle (Figure 28 and 29). The gastrocnemius muscle activation patterns had a nearly inverse relationship to that of the quadriceps. Fleming *et al.* (2001) examined the relationship between ACL strain and isolated contraction of the gastrocnemius, quadriceps, and hamstrings. They found that the amount of ACL strain produced was significantly greater when both the gastrocnemius and the quadriceps were simultaneously stimulated, compared to isolated stimulation of either muscle. ACL strain from gastrocnemius stimulation was found to be greater as knee flexion angle increased up to 30°, with 2.8% and 3.5% strain produced at 5° and 15°, respectively. The results of the present study, along with the findings of Fleming *et al.* (2001), indicate that the neuromuscular system acts in a coordinated fashion to maintain JS<sub>R</sub> and minimize anterior tibial femoral shear force.

## 5.6 HYPOTHESES REVISITED

1a. *Total muscle contributions to knee flexion-extension JS<sub>R</sub> will show a statistically significant main effect ( $p < 0.05$ ) between leg conditions. Post hoc analysis will reveal that the brace involved leg condition will have significantly lower total knee JS<sub>R</sub> compared to the unbraced involved leg.*

The hypothesis was accepted as a significant main effect for leg condition was found.

Post-hoc analysis revealed that the braced involved leg condition had significantly

higher overall  $JS_R$  compared to the uninjured leg. The braced involved leg condition actually showed a trend of higher overall knee rotational stiffness values compared to the unbraced involved leg. This finding was attributed to the increased activation of the quadriceps muscles while wearing the functional knee brace. The design of the functional knee brace worn by subjects in this study, incorporates a four point pressure system. As part of this four point pressure system, a posteriorly directed tibial femoral shear force is applied to the proximal-anterior aspect of the tibia. It is postulated that the quadriceps were activated to a greater degree to overcome this posteriorly directed shear force while attempting to maintain knee joint angle at the specified range.

*1b. Total muscle contributions to knee flexion-extension  $JS_R$  will show a statistically significant interaction ( $p < 0.05$ ) between leg condition and squat task. Further Post hoc analysis will reveal that the highest total  $JS_R$  will be displayed for the unbraced involved leg performing the unstable single leg squat, compared to any other leg x squat task combination.*

It was not possible to reject the null hypothesis as no significant interaction effects between leg condition and squat task were found for any one of the eleven muscles analyzed.

*2. The vastus lateralis will exhibit the greatest contribution to knee flexion-extension  $JS_R$  of all the extensor muscles, while the semimembranosus will display the highest  $JS_R$  of all the knee flexors. The ratio of vastus lateralis and semimembranosus EMG will be*

*used to quantify hamstring co-activation and to determine if any qualitative relationship exist between co-contraction and rotational joint stiffness.*

The hypothesis was accepted for the knee extensors, but it was not possible to reject the null hypothesis for the knee flexors. The vastus lateralis had the largest contribution to JS<sub>R</sub> of all the muscles across all conditions, while the medial head of the gastrocnemius, not the semimembranosus, had the largest contribution to JS<sub>R</sub> of all the knee flexor muscles. The co-activation patterns between the vastus lateralis and semimembranosus displayed a somewhat inverse relationship compared to the pattern of JS<sub>R</sub>. The co-activation values decreased as knee joint angle increased from the most extended knee position to 40-45°, while the JS<sub>R</sub> values increased as knee joint angle increased from the most extended knee position to 40-45°.

## **5.7 LIMITATIONS**

The potential limitations of this study were the diverse range of ages (21-47 years) of the subjects recruited and tested and the variability in functional status of the subjects in either the ACL deficient or reconstructed group. (Appendix C). Despite this large range in ages between subjects, several significant interaction and main effects were found for several of the dependent variables examined. The relatively small sample size, and the substantial differences in functional status within each ACL injury group, may have contributed to the lack of significant findings between the two groups. The small sample sizes for the ACL deficient (n=9) and reconstructed (n=8) subjects in the current study had insufficient power to potentially reject the null hypothesis. Due to the difficulty

in recruiting unilaterally injured subjects for this project, the screening process and exclusion criteria had to be more accommodative. Therefore, both subject groups differed substantially in the following areas: functional injury status, elapsed time since injury, degree of ACL-rupture, level of sports participation, injury status (rehabilitation and surgical reconstruction techniques), previous functional knee brace wear (length of use and type of brace), concomitant injury (meniscal, accessory structures, collateral ligaments).

One of the major constraints in the determination of rotational stability is that mechanical equilibrium must exist (i.e. the first derivative of potential energy must be zero). However, the purpose of this investigation was to determine the overall and individual muscle contributions to knee joint rotational stiffness, and therefore the potential energy of the trunk, pelvis and thigh did not have to be determined. As well, static equilibrium was not as large a constraint for the type of activity that could be performed during the study. Quasi-static squat tasks were chosen so that subjects could maintain similar knee joint angle ranges with the same level of consistency. The stable and unstable single leg squats performed were thought to generate low angular accelerations about the knee joint. In spite of this, several significant findings were found for each of the dependent variables. Due to the overwhelming amount of data collected during this study, average kinematic and kinetic data were input into the stability equation. Unfortunately, this method of processing did not allow us to consider the variability in joint angles, muscle length, knee moment, functional moment arm, and muscle activity throughout each trial and its influence on knee joint stiffness. An analysis of the neuromuscular response of the lower extremity and knee to the perturbation that



occurred just after the solenoid was disengaged from the wobble board, may have provided valuable insight into the mechanisms that the body uses to deal with instability.

The stiffness analysis performed in this thesis only took into account the contributions of the muscles about the flexion-extension axis of the knee, and did not consider the passive contributions provided by ligaments, articular cartilage, the joint capsule and compressive joint loading. Additionally, it is difficult to specifically determine how the functional knee brace influenced the individual muscle contributions to joint stiffness, as functional knee braces have been shown to augment both muscle activity and the passive stiffness of muscles and tissues surrounding the knee joint, respectively. The underlying assumption, in only considering joint stiffness about the flexion-extension axis of the knee, is that moments about the valgus-varus and transverse axes of the knee were negligible. A complete stability analysis of the knee would require the quantification of the potential energy of the segments (thigh, pelvis and trunk) proximal to the knee joint and the calculation of a 3 X 3 Hessian matrix to account for the rotations about the flexion-extension, transverse and valgus-varus axes of the knee, hip, and ankle. Therefore, the gain factor was determined solely for the net flexion-extension moment of the knee joint. A complete stability analysis would require a much more sophisticated method to collect the kinematic data and achieve a solution with simultaneous static equilibrium about 9 degrees of freedom.

Several studies (Lloyd and Besier, 2003; Doorenbosch and Harlaar, 2003; Liu and Maitland, 2000; and Lu *et al.*, 1998) have attempted to quantify the forces acting at the knee joint by utilizing a simplified, sagittal plane, 2 dimensional biomechanical model of the lower extremity to represent the three dimensional anatomical structure of the lower

extremity. Lu *et al.* (1998) determined that a sagittal plane model underestimates the maximal axial force in the femur during walking by approximately 30%. They suggest that 70% of this force was attributed to the action of the extensor and flexors. In the present study, the estimates of the knee flexion-extension moments from muscle activity were determined between 5°-40°. According to Kapandji (1987), the knee achieves 72% of its maximal lateral rotation about the transverse axis at 30° of knee flexion compared to that achieved at 90°. Therefore, neglecting the muscle forces that act to stabilize the transverse motions of tibial-femoral joint (Figure 9) may not allow a comprehensive understanding of the neuromuscular response to ACL injury status (ACL deficient vs. reconstructed), surface stability index (stable vs. unstable), knee angle, and functional knee braces.

Additional limitations to this thesis, that may have influenced the calculation of the knee joint moments derived from both inverse dynamics and electromyography, include: the assumed flexion-extension axis of the knee; the linear EMG-to-force relationship used to calculate muscle force (Alkner *et al.*, 2000); and the lack of osteometric scaling (Brand *et al.*, 1982; White *et al.* 1989) along the Y-axis to account for differences in subjects' height compared to the model defined by Delp *et al.* (1990).

## **5.8 FUTURE DIRECTIONS**

Future studies should be directed at investigating more complex three dimensional knee loading patterns that occur in rotational sports such as: soccer, basketball, and football. The current study only challenged subjects in the sagittal plane, as the wobble board was stable in each of the other 2 anatomical planes. A larger, younger, more

functionally similar set of ACL deficient and reconstructed subjects should be used to determine the influence of surgery on  $JS_R$ . An additional direction that needs to be examined is the influence of gender on knee joint stiffness. The development of a MRI based biomechanical model of the female lower extremity anatomy is critical to understand the role that physical structure plays when considering the stiffness provided by both active and passive structures acting at the knee. This should be accomplished with the tibiofemoral joint flexed in three knee angle ranges ( $0^\circ$ ,  $15^\circ$ , and  $30^\circ$ , Aalbersberg, 2006) at neutral and valgus knee postures under a physiological load. Additionally, the role of foot and ankle structure should be evaluated to determine its influence in rotational and valgus-varus kinematics, and thus  $JS_R$  of the knee joint.

A more precise method, such as high speed biplane radiography and computer tomography (Tashman and Anderst, 2003) for determining knee three dimensional kinematics, needs to be incorporated into the analysis of  $JS_R$  of the knee joint. Tashman and Anderst (2003) reported an average precision of 0.064 mm and  $0.31^\circ$  in tracking the distance and the angles between tantalum markers, respectively. This method effectively removes the skin motion artifacts and improves tracking precision during the complex three dimensional movements of the tibiofemoral joint that occurs with knee flexion. The use of this precise method to determine tibiofemoral joint position would provide extremely valuable data that could be used to calculate *in-vivo* ACL ligament strain, and ultimately determine the relationship between shear force musculature and rotational joint stiffness.

## 5.9 CONCLUSIONS

The joint rotational stiffness analysis completed in this project defines a new paradigm in the evaluation of ACL injury, functional knee bracing, and rehabilitative exercise. To date, this is the first study to incorporate linked segment biomechanical and anatomical models of the lower extremity to determine knee joint rotational stiffness ( $JS_R$ ). This stability based model of the lower extremity provided valuable information regarding the specific individual muscle contributions to overall knee joint rotational stiffness, especially with respect to ACL injury. The most significant findings from this analysis were as follows:

- 1) In the braced leg conditions, subjects displayed significantly higher overall  $JS_R$  compared to the control leg despite a significant decrease in muscle activation of medial head of the gastrocnemius, the most important contributor to  $JS_R$  of all the knee flexor muscles. The lack of significant differences between *Leg* conditions, for each of the individual major knee stabilizers, revealed that consideration of the sum of these non-significant increases in  $JS_R$  is imperative to understanding the role of all musculature crossing the knee, related to overall knee joint rotational stiffness. The increase in  $JS_R$  facilitated by the FLEX functional knee brace may act in conjunction with the passive stabilizing capacity that the brace offers. The functional knee brace may provide both enhanced dynamic and passive knee joint stiffness to the ACL injured subject. Although a dynamic increase in joint stiffness was demonstrated with acute brace wear, premature muscle fatigue and other

negative side effects may occur with prolonged brace use during sporting events and recreational activities.

- 2) A significant interaction was found between knee angle and squat task. For each knee angle range, the unstable squat trials had significantly higher  $JS_R$  values than the stable squat trials. However, the difference in the  $JS_R$  values between the stable and unstable squat tasks at the 5-10° was 1.5 greater than at 20-25° and 2.8 times greater than at 40-45°. This larger increase in stiffness at the most extended knee posture suggest that the knee is more vulnerable to buckling at 5-10°, particularly if an unexpected perturbation is applied to the lower extremity during a squat performed on a stable surface.
  
- 3) The influence of *Squat Task* (unstable vs. stable) on  $JS_R$  was significant for nine of the 11 muscles studied. This indicated that functional training exercises, such as that performed on a wobble board, challenges the neuromuscular system sufficiently to require the augmentation of muscle activity of the knee joint musculature, thus increasing overall stiffness about the flexion/extension axis of the knee.

**APPENDICES**

### **Explanation of Revision/Amendment to Ethics Application**

Recently, I attended the International Society of Biomechanics Conference in Cleveland, Ohio. Several of the podium presentations on ACL deficient subjects and knee mechanics have provided the impetus for revisions to the methodology of this project.

The single leg squat task will be modified to a static task. Subjects will be requested to maintain a specified knee joint angle range while standing on a force plate. Three selected knee joint angle ranges (10-15, 20-25, and 40-45 degrees) will be tested in a randomized order for each brace condition. The isokinetic knee extension task will be eliminated from the experimental protocol due to technological difficulties associated with the brace, EMG system and quantification of lower body segment angles in a dynamometer and because it is not as appropriate for our purpose as the task we are now proposing. The isokinetic task will be replaced with a static unstable single leg squat on a wobble-board placed over the force plate. The wobble board will be unstable only in the flexion-extension axis of the knee or sagittal plane. Subjects will be required to maintain specified knee joint angles for each of the three brace conditions, exactly as specified for the stable single leg squat task. Unstable surfaces are typically used in physical therapy or sport training facilities to allow injured patients to learn specific movement patterns to compensate for injury deficits and return function to normal. Examples of unstable surfaces used in these settings include: BOSU balls, Swiss/Resistance Balls, Extreme Balance Boards, Sit-Fit Balls, Reebok Core Board, and Fitter Wobble Boards.

#### ***Known/Anticipated Risk***

The unstable static single leg squat task may be a task that is novel to the subject. However, each subject will be given adequate time to learn to develop adaptations to balancing on the wobble board prior to testing. This will be done such that subjects will have stabilizing platforms to contact if they feel the need to do so. A balance board will be selected that has a very large radius of curvature to minimize the anterior-posterior displacement velocity of the subject's center of pressure (as measured from the force plate).

## SUBJECT CONSENT AND FUNCTIONAL STATUS FORMS



### CONSENT TO PARTICIPATE IN RESEARCH

**Title of Study:** Stability Based Biomechanical Model of the ACL Injured Knee with and without Functional Knee Brace Use

You are asked to participate in a research study conducted by Dr. Jim Potvin, Associate Professor, Department of Human Kinetics and Aaron Derouin, Masters Candidate, Department of Human Kinetics at the University of Windsor. The results of this study will contribute towards a thesis, in partial fulfilment of the requirements for the degree of Masters of Human Kinetics.

If you have any questions or concerns about the research, please feel free to contact Dr. Jim Potvin, Research Advisor @ 253-3000 (ext. 2461) or Aaron Derouin, Graduate Student @ 253-3000 (ext. 2468).

#### PURPOSE OF THE STUDY

The purpose of the study is to quantitatively assess the effectiveness of functional knee bracing in providing mechanical stability to the knee joint of anterior cruciate ligament deficient or surgically reconstructed subjects.

#### PROCEDURES

If you volunteer to participate in this study, we would ask you to do the following things:

Initially, you will be asked to fill out a series of questionnaires to assess the degree of your ACL injury, other lower extremity injuries, the treatment you received following the injury, and your current functional status. Additionally, you will be requested to perform a timed single leg hop as quickly as you can on your non-injured leg; then you will be asked to perform the same task on your injured leg. This screening process will be used to identify your functional status prior to and post ACL injury.

Subjects selected for the study will have a negative plaster impression or measurements taken of the injured leg from which a custom or off the shelf functional knee brace will be designed, fabricated and/or fit. Following measurement of the leg and sizing of the brace, subjects will be don the brace to ensure proper fit and comfort. Participation in this study will occur over two time periods, an orientation/training session and a data collection session. The total length of time for subject involvement will be approximately 4 hours; 3 hours for training and data collection and 1 hour for brace casting/measurement and fitting.

Subjects will perform 2 tasks with 3 conditions per task. During the first task you will perform a stable static single leg squat on a force platform. For the second task subjects will complete a static unstable single leg squat on a Fitter-like Wobble board. In each of the tasks subjects will be required to maintain a knee joint angle over 3 different joint angle ranges: 5-10, 20-25, and 40-45 degrees. Subjects will perform



each task under 3 conditions: unbraced, braced, and the uninjured leg. Each condition for each task will be performed three times to ensure consistent muscle activation patterns are achieved.

Surface electrodes will be positioned over specific areas of the thigh and leg to assess muscle activity of several muscle groups during the first two tasks. The surface electrodes will be placed over each of the following muscles: gastrocnemius, hamstrings, quadriceps, tensor fascia latae, and sartorius. Surface electrodes only record muscle activity, whereas stimulating electrodes provide a brief pulse of electrical current to further activate the muscle.

## **POTENTIAL RISKS AND DISCOMFORTS**

During the single leg hop test you may choose not to participate if you feel that these tasks will your cause your knee to become unstable or give way. Timed single leg hopping test have been used by numerous investigators (Fitzgerald, G.K., Axe, M.J., Synder-Mackler, L, 2000; Eastlack, M.E., Axe, M.J., Snyder-Mackler, L., 1999; Snyder-Mackler, L., Fitzgerald, G.K., Bartolozzi, A.R., Ciccotti, M.G., 1997; Rudolph, K.S., Axe, M.J., Snyder-Mackler, L., 2000) as a method to screen non-coper ACL-deficient subjects from copers.

Muscle stiffness may result after the collection, but this should be no more than would be experienced after any unaccustomed physical activity. In addition, the self-adhesive, surface electrodes may cause a slight skin irritation. This discoloration will not last longer than a day or two and poses no health risk to the participant.

## **POTENTIAL BENEFITS TO SUBJECTS AND/OR TO SOCIETY**

Subjects will benefit by developing a further understanding of ACL-injury and the effectiveness of functional knee bracing in providing dynamic stability to the knee.

Health care practitioners that treat ACL-injured patients (orthopaedic surgeons, physical therapist, athletic therapist and orthotist) will gain further insight into the mechanisms by which functional knee braces modify muscle activity to stabilize the knee joint following injury to the ACL. Additionally, biomechanist, brace manufacuters/designers, and orthotist will have a new method in which to evaluate the efficacy of functional knee bracing and knee instability, as a result of ACL disruption.

## **PAYMENT FOR PARTICIPATION**

Participation in this study is on a volunteer basis and no remuneration will be provided to you for the time you spend as a subject in this study.

## **CONFIDENTIALITY**

All data will be kept confidential. Only the researchers mentioned above will know your identity and personal information. This information will be stored in a secure computer in the biomechanics/ergonomics laboratory and will not be discussed or displayed in any form that would provide an indication of your identity.

## PARTICIPATION AND WITHDRAWAL

You can choose whether to be in this study or not. If you volunteer to be in this study, you may withdraw at any time without consequences of any kind. You may also refuse to answer any questions you do not want to answer and still remain in the study. The investigator may withdraw you from this research if circumstances arise which warrant doing so.

## FEEDBACK OF THE RESULTS OF THIS STUDY TO THE SUBJECTS

Procedures and methods pertaining to this study will be communicated both verbally and in written format prior to the onset of the study. If there has been anything neglected or something you wish to be further clarified feel free to ask questions at any time before, during or after the study. If desired, subjects will be given verbal and/or written feedback once all the data has been analyzed.

## RIGHTS OF RESEARCH SUBJECTS

You may withdraw your consent at any time and discontinue participation without penalty. This study has been reviewed and received ethics clearance through the University of Windsor Research Ethics Board. If you have questions regarding your rights as a research subject, contact:

Research Ethics Coordinator  
University of Windsor  
Windsor, Ontario  
N9B 3P4

Telephone: 519-253-3000, ext. 3916  
E-mail: lbunn@uwindsor.ca

## SIGNATURE OF RESEARCH SUBJECT/LEGAL REPRESENTATIVE

I understand the information provided for the study, titled, "Stability Based Biomechanical Model of the ACL Injured Knee with and without Functional Knee Brace Use", as described herein. My questions have been answered to my satisfaction, and I agree to participate in this study. I have been given a copy of this form.

\_\_\_\_\_  
Name of Subject

\_\_\_\_\_  
Signature of Subject

\_\_\_\_\_  
Date

## SIGNATURE OF INVESTIGATOR

These are the terms under which I will conduct research.

\_\_\_\_\_  
Signature of Investigator

\_\_\_\_\_  
Date

## TEGNER ACTIVITY LEVEL SCALE

Please indicate in the spaces below the **HIGHEST** level of activity that you participated in **BEFORE YOUR INJURY** and the highest level you are able to participate in **CURRENTLY**.

**BEFORE INJURY:** Level \_\_\_\_\_ **CURRENT:** Level \_\_\_\_\_

Level 10	Competitive sports- soccer, football, rugby (national elite)
Level 9	Competitive sports- soccer, football, rugby (lower divisions), ice hockey, wrestling, gymnastics, basketball
Level 8	Competitive sports- racquetball or bandy, squash or badminton, track and field athletics (jumping, etc.), down-hill skiing
Level 7	Competitive sports- tennis, running, motorcars speedway, handball Recreational sports- soccer, football, rugby, bandy, ice hockey, basketball, squash, racquetball, running
Level 6	Recreational sports- tennis and badminton, handball, racquetball, down-hill skiing, jogging at least 5 times per week
Level 5	Work- heavy labor (construction, etc.) Competitive sports- cycling, cross-country skiing, Recreational sports- jogging on uneven ground at least twice weekly
Level 4	Work- moderately heavy labor (e.g. truck driving, etc.)
Level 3	Work- light labor (nursing, etc.)
Level 2	Work- light labor Walking on uneven ground possible, but impossible to back pack or hike
Level 1	Work- sedentary (secretarial, etc.)
Level 0	Sick leave or disability pension because of knee problems

Y Tegner and J Lysohm. *Rating Systems in the Evaluation of Knee Ligament Injuries*. Clinical Orthopedics and Related Research. Vol. 198: 43-49, 1985.

### **SURGICAL HISTORY**

Have you had any additional surgeries to your knee other than those performed by Dr. Stone?

Yes / No

If Yes:

What procedure(s) were performed? \_\_\_\_\_

When was the surgery performed? \_\_\_\_\_

Who performed the surgery? \_\_\_\_\_

## Lysholm Knee Scale

### Limp (5 Points)

None 5 \_\_\_\_\_  
 Slight or periodic 3 \_\_\_\_\_  
 Severe and constant 0 \_\_\_\_\_

### Support (5 Points)

Full Support 5 \_\_\_\_\_  
 Cane or crutch 3 \_\_\_\_\_  
 Weight Bearing impossible 0 \_\_\_\_\_

### Stair Climbing (5 points)

No problems 5 \_\_\_\_\_  
 Slightly impaired 3 \_\_\_\_\_  
 One step at a time 2 \_\_\_\_\_  
 Unable 0 \_\_\_\_\_

### Squatting (5 Points)

No problem 5 \_\_\_\_\_  
 Lightly impaired 3 \_\_\_\_\_  
 Not past 90 degrees 2 \_\_\_\_\_  
 Unable 0 \_\_\_\_\_

### TOTAL

\_\_\_\_\_

### Walking, Running and Jumping

#### Instability (30 Points)

Never giving way 30 \_\_\_\_\_  
 Rarely gives way except  
 for athletic or other  
 severe exertion 25 \_\_\_\_\_  
 Gives way frequently  
 during athletic events  
 or severe exertion 0 \_\_\_\_\_  
 Occasionally in daily activities 10 \_\_\_\_\_  
 Often in daily activities 5 \_\_\_\_\_  
 Every step 0 \_\_\_\_\_

#### Swelling (10 Points)

None 10 \_\_\_\_\_  
 With giving way 7 \_\_\_\_\_  
 On severe exertion 5 \_\_\_\_\_  
 On ordinary exertion 2 \_\_\_\_\_  
 Constant 0 \_\_\_\_\_

#### Pain (30 Points)

None 30 \_\_\_\_\_  
 Inconstant and slight  
 during severe exertion 25 \_\_\_\_\_  
 Marked on giving way 20 \_\_\_\_\_  
 Marked during severe exertion 15 \_\_\_\_\_  
 Marked on or after walking  
 more than 1 ¼ miles 10 \_\_\_\_\_  
 Marked on or after walking  
 less than 1 ¼ miles 5 \_\_\_\_\_  
 Constant and severe 0 \_\_\_\_\_

#### Atrophy of thigh (5 Points)

None 5 \_\_\_\_\_  
 1-2 cm 3 \_\_\_\_\_  
 > 2 cm 0 \_\_\_\_\_

### TOTAL

\_\_\_\_\_

**Knee Outcome Survey  
Activities of Daily Living Scale**

**Instructions:**

The following questionnaire is designed to determine the symptoms and limitations that you experience because of your knee while you perform your usual daily activities. Please answer each question by checking the one statement that best describes you over the last 1 to 2 days. For a given question, more than one of the statements may describe you, but please mark only the statement which best describes you during your usual daily activities.

**Symptoms**

To what degree does each of the following symptoms affect your level of daily activity? (check one answer on each line)

	<b>I Do Not Have the Symptom</b>	<b>I Have the Symptom But It Does Not Affect My Activity</b>	<b>The Symptom Affects My Activity Slightly</b>	<b>The Symptom Affects My Activity Moderately</b>	<b>The Symptom Affects My Activity Severely</b>	<b>The Symptom Prevents Me From All Daily Activities</b>
Pain	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Stiffness	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Swelling	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Giving Way, Buckling or Shifting of Knee	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Weakness	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Limping	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

### Functional Limitations with Activities of Daily Living

How does your knee affect your ability to... (check one answer on each line)

	Activity Is Not Difficult	Activity is Minimally Difficult	Activity is Somewhat Difficult	Activity is Fairly Difficult	Activity is Very Difficult	I am Unable to Do the Activity
Walk?	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Go up stairs?	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Go down stairs?	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Stand?	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Kneel on the front of your knee?	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Squat?	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Sit with your knee bent?	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Rise from a chair?	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

How would you rate the current function of your knee during your usual daily activities on a scale from 0 to 100 with 100 being your level of knee function prior to your injury and 0 being the inability to perform any of your usual daily activities?

\_\_\_\_\_

How would you rate the overall function of your knee during your usual daily activities? (please check the one response that best describes you)

- normal
- nearly normal
- abnormal
- severely abnormal

As a result of your knee injury, how would you rate your current level of daily activity? (please check the one response that best describes you)

- normal
- nearly normal
- abnormal
- severely abnormal

**Knee Outcome Survey  
Sports Activities Scale**

**Instructions:**

The following questionnaire is designed to determine the symptoms and limitations that you experience because of your knee while you perform sports activities. Please answer each question by **checking the one statement that best describes you over the last 1 to 2 days**. For a given question, more than one of the statements may describe you, but please mark only the statement which best describes you when you participate in sports activities.

**Symptoms**

To what degree does each of the following symptoms affect your level of sports activity? (check one answer on each line)

	Never Have	Have, But Does Not Affect Sports Activity	Affects Sports Activity Slightly	Affects Sports Activity Moderately	Affects Sports Activity Severely	Prevents Me From All Sports Activity
Pain	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Grinding or Grating	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Stiffness	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Swelling	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Slipping or Partial Giving Way of Knee	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Buckling or Full Giving Way of Knee	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Weakness	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

### Functional Limitations with Sports Activities

How does your knee affect your ability to... (check one answer on each line)

	Not Difficult at All	Minimally Difficult	Somewhat Difficult	Fairly Difficult	Very Difficult	Unable to Do
Run straight ahead?	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Jump and land on your involved leg?	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Stop and start quickly?	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Cut and pivot on your involved leg?	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

How would you rate the current function of your knee during sports activities on a scale from 0 to 100 with 100 being your level of knee function prior to your injury and 0 being the inability to perform any sports activities?

\_\_\_\_\_

How would you rate the overall function of your knee during sports activities? (please check the one response that best describes you)

- normal
- nearly normal
- abnormal
- severely abnormal

As a result of your knee problem, how would you rate your current level of activity during sports? (please check the one response that best describes you)

- normal
- nearly normal
- abnormal
- severely abnormal



### Changes in Sports Activity

Describe your highest level of sports activity at each of the following points in time. (check one answer on each line)

	<b>Strenuous Sports (ex. football, soccer, basketball)</b>	<b>Moderate Sports (ex. tennis, skiing)</b>	<b>Light Sports (ex. cycling, swimming, golf)</b>	<b>No Sports Activities Possible</b>
Prior to your knee injury	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Prior to treatment of your knee injury	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Currently	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

Describe the frequency that you participated in sports activity at each of the following points in time. (check one answer on each line)

	<b>4 to 7 Times per Week</b>	<b>1 to 3 Times per Week</b>	<b>1 to 3 Times per Month</b>	<b>Less Than 1 Time per Month</b>
Prior to your knee injury	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Prior to treatment of your knee injury	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
Currently	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

## 2000 IKDC SUBJECTIVE KNEE EVALUATION FORM

Your Full Name \_\_\_\_\_

Today's Date: \_\_\_\_\_ / \_\_\_\_\_ / \_\_\_\_\_  
Day Month Year

Date of Injury: \_\_\_\_\_ / \_\_\_\_\_ / \_\_\_\_\_  
Day Month Year

### **SYMPTOMS\***:

\*Grade symptoms at the highest activity level at which you think you could function without significant symptoms, even if you are not actually performing activities at this level.

1. What is the highest level of activity that you can perform without significant knee pain?

- Very strenuous activities like jumping or pivoting as in basketball or soccer
- Strenuous activities like heavy physical work, skiing or tennis
- Moderate activities like moderate physical work, running or jogging
- Light activities like walking, housework or yard work
- Unable to perform any of the above activities due to knee pain

2. During the past 4 weeks, or since your injury, how often have you had pain?

	0	1	2	3	4	5	6	7	8	9	10	
Never	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	Constant

3. If you have pain, how severe is it?

	0	1	2	3	4	5	6	7	8	9	10	
No pain	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	Worst pain imaginable

4. During the past 4 weeks, or since your injury, how stiff or swollen was your knee?

- Not at all
- Mildly
- Moderately
- Very
- Extremely

5. What is the highest level of activity you can perform without significant swelling in your knee?

- Very strenuous activities like jumping or pivoting as in basketball or soccer
- Strenuous activities like heavy physical work, skiing or tennis
- Moderate activities like moderate physical work, running or jogging
- Light activities like walking, housework, or yard work
- Unable to perform any of the above activities due to knee swelling

6. During the past 4 weeks, or since your injury, did your knee lock or catch?

- Yes     No

7. What is the highest level of activity you can perform without significant giving way in your knee?

- Very strenuous activities like jumping or pivoting as in basketball or soccer
- Strenuous activities like heavy physical work, skiing or tennis
- Moderate activities like moderate physical work, running or jogging
- Light activities like walking, housework or yard work
- Unable to perform any of the above activities due to giving way of the knee

Page 2 – 2000 IKDC SUBJECTIVE KNEE EVALUATION FORM

**SPORTS ACTIVITIES:**

8. What is the highest level of activity you can participate in on a regular basis?

- Very strenuous activities like jumping or pivoting as in basketball or soccer
- Strenuous activities like heavy physical work, skiing or tennis
- Moderate activities like moderate physical work, running or jogging
- Light activities like walking, housework or yard work
- Unable to perform any of the above activities due to knee

9. How does your knee affect your ability to:

		Not difficult at all	Minimally difficult	Moderately Difficult	Extremely difficult	Unable to do
a.	Go up stairs	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
b.	Go down stairs	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
c.	Kneel on the front of your knee	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
d.	Squat	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
e.	Sit with your knee bent	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
f.	Rise from a chair	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
g.	Run straight ahead	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
h.	Jump and land on your involved leg	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>
i.	Stop and start quickly	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>	<input type="checkbox"/>

**FUNCTION:**

10. How would you rate the function of your knee on a scale of 0 to 10 with 10 being normal, excellent function and 0 being the inability to perform any of your usual daily activities which may include sports?

FUNCTION PRIOR TO YOUR KNEE INJURY:

Cannot perform daily activities      0   1   2   3   4   5   6   7   8   9   10      No limitation in daily activities

CURRENT FUNCTION OF YOUR KNEE:

Cannot perform daily activities      0   1   2   3   4   5   6   7   8   9   10      No limitation in daily activities

## 2000 IKDC KNEE HISTORY FORM

Patient Name \_\_\_\_\_ Birthdate \_\_\_\_\_ / \_\_\_\_\_ / \_\_\_\_\_  
Day Month Year

Date of Injury \_\_\_\_\_ / \_\_\_\_\_ / \_\_\_\_\_ Date of Initial Exam \_\_\_\_\_ / \_\_\_\_\_ / \_\_\_\_\_ Today's Date \_\_\_\_\_ / \_\_\_\_\_ / \_\_\_\_\_  
Day Month Year Day Month Year Day Month Year

Involved Knee:  Right  Left

Contralateral:  Normal  Nearly Normal  Abnormal  Severely abnormal

Onset of Symptoms: (date) \_\_\_\_\_ / \_\_\_\_\_ / \_\_\_\_\_  
Day Month Year

Chief Complaint: \_\_\_\_\_

Activity at Injury:  ADL  Sports  Traffic  Work

Mechanism of Injury:

- Non-traumatic gradual onset  Traumatic non-contact onset  
 Non-traumatic sudden onset  Traumatic contact onset

### Previous Surgery:

Type of Surgery: (check all that apply)

#### Meniscal Surgery

- Medial meniscectomy  Lateral meniscectomy  
 Medial meniscal repair  Lateral meniscal repair  
 Medial meniscal transplant  Lateral meniscal transplant

#### Ligament Surgery

- ACL Repair  Intraarticular ACL reconstruction  Extraarticular ACL reconstruction  
 PCL Repair  Intraarticular PCL reconstruction  Posterolateral corner reconstruction  
 Medial collateral ligament repair/reconstruction  
 Lateral collateral ligament repair/reconstruction

#### Type of Graft

- Patella tendon graft  Ipsilateral  Contralateral  
 Single hamstring graft  
 2 Bundle hamstring graft  
 4 Bundle hamstring graft  
 Quadriceps tendon graft  
 Allograft  
 Other

## Page 2 – 2000 IKDC KNEE HISTORY FORM

## Extensor Mechanism Surgery

- Patella tendon repair       Quadriceps tendon repair

## Patellofemoral Surgery

- Extensor Mechanism Realignment

## Soft Tissue Realignment

- Medial imbrication       Lateral release

## Bone Realignment

## Movement of the tibial tubercle

- Proximal       Distal       Medial       Lateral       Anterior

- Trochleoplasty

- Patellectomy

## Osteoarthritis Surgery

- Osteotomy

- Articular Surface Surgery       Shaving       Abrasion       Drilling       Microfracture  
 Cell therapy       Osteochondral autograft transfer/mosaic-plasty       Other

Total number of previous surgeries \_\_\_\_\_

## Imaging Studies:

- Structural       MRI       CT       Arthrogram

- Metabolic (Bone Scan)

## Findings:

Ligament \_\_\_\_\_

Meniscus \_\_\_\_\_

Articular Cartilage \_\_\_\_\_

Bone \_\_\_\_\_

**POTENTIAL ACL DEFICIENT NON-COPER SUBJECTS**

Name: \_\_\_\_\_ Date: \_\_\_\_\_

Phone: \_\_\_\_\_ Weight (lbs): \_\_\_\_\_

Age: \_\_\_\_\_ D.O.B.: \_\_\_\_\_ Height (ft/in): \_\_\_\_\_

Date of Injury: \_\_\_\_\_ Side Injured: \_\_\_\_\_

Name of Physician Diagnosing ACL Injury: \_\_\_\_\_

Injury Diagnosed by MRI or Arthroscopy: \_\_\_\_\_

Other Structural Damage to Involved Leg?  
check yes or no

Meniscus	yes <input type="checkbox"/>	no <input type="checkbox"/>
MCL	yes <input type="checkbox"/>	no <input type="checkbox"/>
LCL	yes <input type="checkbox"/>	no <input type="checkbox"/>
PCL	yes <input type="checkbox"/>	no <input type="checkbox"/>

Please describe any structural deficits found at the time your ACL injury was evaluated or any other major injuries such as fractures, dislocations, or severe sprains.  
\_\_\_\_\_  
\_\_\_\_\_Structural Damage to Non-ACL injured Leg? yes  no   
Select Yes or No. Please Describe Below if You Answered YesWas your ACL surgically repaired? yes  no 

Date of Surgery: \_\_\_\_\_ Name of Surgeon: \_\_\_\_\_

Did you complete a physiotherapy program? yes  no   
For How Long? \_\_\_\_\_Do you wear a functional knee brace? yes  no   
What type? \_\_\_\_\_Since injuring your knee, please indicate the number of times your knee has *given way* or experienced an episode of *instability* 1  2  3  4+   
check one only

## APPENDIX B

### WOBBLE BOARD CONTROL CIRCUIT DESCRIPTION

The wobble board (Figure 34) is kept from moving by the restraining action of a pin in a hole located in the edge of the board, perpendicular to the axis of wobble. This pin is pulled out of engagement by activating an electrically powered solenoid (Figure 35). The subject is first given a visual indication of the impending release of the board by the lighting of a light bulb, and after a preset but variable delay, the solenoid is actuated. The computer system is supplied with a 1Volt DC signal that indicates that the subject has been notified, and ends when the solenoid is actuated.

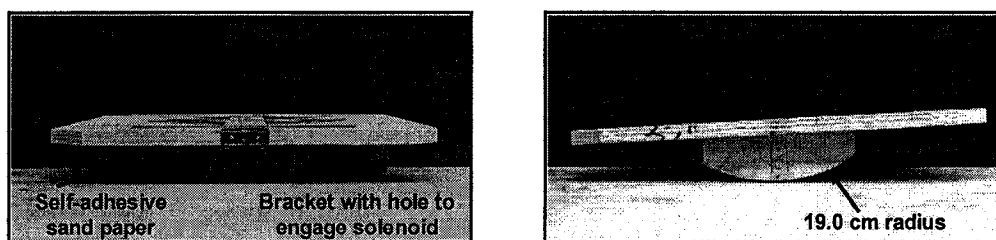
The solenoid used for this study was a McMaster-Carr, part # 7723K6. This is a 2.5" long, by 2.1" high, by 1.8" wide solenoid, with a 1" long stroke. It has a pull force rating of 150 oz at a 1/8" stroke. It operates continuously on 120 VAC at a power of 50 watts.

The circuit that controls this system has a start button that is activated by the researcher. The start button is depressed after data collection on the video camera and computer data collection systems has been initiated.

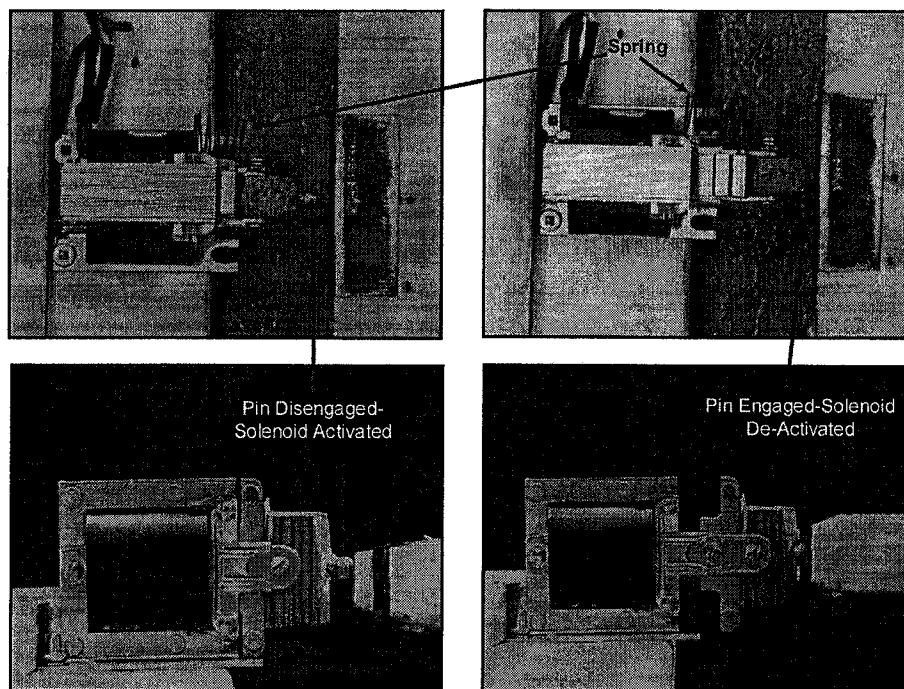
This applies power to an AC adaptor style, 12 Volt DC power supply. This powers the lamp that alerts the subject to the timed delay release of the wobble board by solenoid activation. It also supplies power to light a couple of small LEDs, one of which is located in the corner of the field of view of the video camera, and the other is located on the control box.

Activated simultaneously by the start button, another section of the circuit supplies the 1 Volt DC timing signal to the computer data collection system. This signal is removed by the opening of the N.C. contact of the timing relay, whose timing period is also initiated by the start button (Figure 36 & 37).

Finally the wobble board is set free to move as the solenoid is activated at the end of this subject notification period, by closing the N.O. contact of the On-Delay style relay. This relay is an Allen Bradley type number 700-HTM12JA1, with its basic timing range set to 0 to 5 seconds.

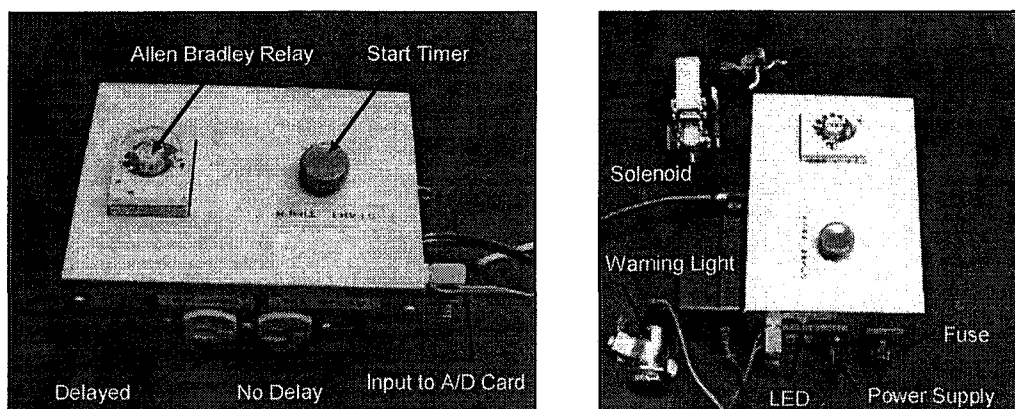


**Figure 34.** Custom fabricated Fitter-Like wobble board. The platform of the wobble board measured 22 x 20 cm, designed larger than the surface area of the force plate. A ¼" rubberized mat (17¾" x 19 ¼" cm) was mounted onto the force plate to minimize slip from shear forces created while subjects performed unstable squat trials. Additionally, self-adhesive sand paper was secured onto the rockered surface of the wobble board to further minimize slip due to shear forces.

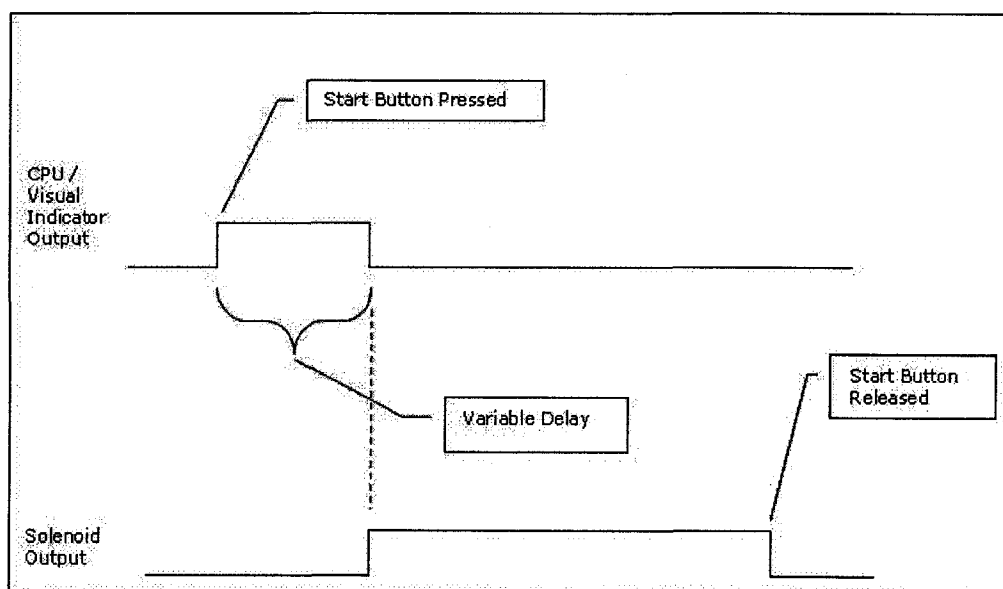


**Figure 35.** Solenoid engagement with wobble board. Pin is retained in the bracket by the spring. Activation of solenoid disengages pin from the hole in the bracket, which is secured to the board.





**Figure 36.** Wobble board control module. The Allen Bradley timer was set to one second delay.



**Figure 37.** Timing and recording of the solenoid activation.

## KNEE BRACE SPECIFICATIONS

PRODUCTS &gt; KNEE &gt; OTS

FLEX



Combines Flex™ with the TheraCool Cold Therapy System for hours of soothing comfort (TheraCool sold separately)

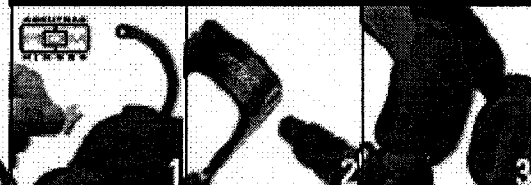
LOW MED HIGH  
CONTACT/IMPACT LEVEL

## FLEX &gt;&gt;

Flex™ might just be the world's most versatile functional knee brace. Whether for functional support, post-op/rehab support, or both, Flex is equipped for the task. Flex's new Accutra™ ROM hinges control flexion and extension and its carbon fiber frame provides long-term durability while being heat-moldable to accommodate changes in leg shape during recovery from injury. GelFit™ condylar pads contour to bony prominences and provide soothing comfort to tender, post-operative joints.

## Indications &gt;&gt;

- ACL, MCL, LCL, PCL, rotary and combined instabilities
- Low, medium and high contact/impact activities
- Post-op, rehab and functional support with ROM control for patients requiring long-term support (>4 weeks after injury or surgery) not provided by disposable post-operative braces or immobilizers



## Product Highlights &gt;&gt;

1. New Accutra™ ROM hinges with easy-to-use flexion/extension stops
  - Lockout: 0° Flexion: 45° 60° 75° 90° // Extension: 10° 15° 30° 40°
  - Optional lockout 0° // Lockout 15° 30° 50° // Flexion: 10° 25° 30°
2. Heat-moldable, carbon composite frame accommodates edema and/or changes in leg shape during recovery (call for heating instructions)
3. GelFit™ condylar pads for improved suspension and comfort

## Other Features &amp; Benefits &gt;&gt;

- No-Slip Guarantee when used with Anti-Migration System™ wrap (AMS™ wrap is free at time of order)
- Total Support System™ for ACL, MCL, LCL, PCL, rotary and combined instabilities
- Rigid frame helps control flexion/extension, insures accurate application and is durable for both sports and everyday activities
- Anterior tibial strap for precise hinge depth location and tibial control
- Large "open window" frame design accommodates post-op dressing and edema
- Non-corrosive materials, ideal for water sports
- Pivoting buckles enable proper strap placement above gastrocnemius
- Simple, 4-strap design allows for quick application and removal
- Easy to stock and cost-effective for today's healthcare environment
- Lightweight, streamlined, non-bulky frame improves compliance



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PART #	DESCRIPTION	SIDE	SIZE	MEASUREMENTS	SHIPS	COLORS	SUGGESTED HCPCS CODE
FLEX-SR	Flex	Right	Small	Caliper: 3 $\frac{1}{2}$ " - 4" (8.9 - 10.2cm)	Same Day	Clear Carbon	L185
FLEX-SL		Left		Circumference: 11 - 12 $\frac{1}{2}$ " (27.9 - 32.1 cm)			
FLEX-MR		Right	Medium	Caliper: 4 - 4 $\frac{1}{2}$ " (10.2 - 11.7cm)			
FLEX-ML		Left		Circumference: 12 $\frac{1}{2}$ - 14 $\frac{1}{2}$ " (32.1 - 36.8cm)			
FLEX-LR		Right	Large	Caliper: 4 $\frac{1}{2}$ - 5 $\frac{1}{2}$ " (11.7 - 13cm)			
FLEX-LL		Left		Circumference: 14 $\frac{1}{2}$ - 16" (36.8 - 40.6cm)			
FLEX-LR		Right	XLarge	Caliper: 5 $\frac{1}{2}$ - 6 $\frac{1}{2}$ " (13 - 14.3cm)			
FLEX-XL		Left		Circumference: 16 - 17 $\frac{1}{2}$ " (40.6 - 44.8cm)			

CALLIPER MEASUREMENTS (M/L at joint width) should be taken in a standard, weight bearing position.  
CIRCUMFERENCE MEASUREMENTS should be taken just below the distal border of the patella.

**OPTIONS >>**

Patella Cap Kit, Gear Guards, XCL Kit, Motorcrose Kit (includes patella cup, gear guards, Softleave & sticks)



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## APPENDIX C

## SUBJECT DEMOGRAPHIC AND FUNCTIONAL STATUS DATA

Table AC-1A. Subject demographic and anthropometric data.

Subject Code	Injury Status D or R	Age	Height	Mass	BMI
		(years)	(cm)	(Kg)	
S02	ACL-D	24.92	189.23	77.34	21.60
S04	ACL-D	34.42	176.53	79.88	25.63
S05	ACL-D	20.75	179.71	72.57	22.47
S07	ACL-D	23.17	183.52	87.77	26.06
S08	ACL-D	20.33	181.61	92.32	27.99
S11	ACL-D	21.08	179.07	82.23	25.64
S12	ACL-D	33.33	185.42	92.82	27.00
S14	ACL-D	25.17	183.52	91.60	27.20
S16	ACL-D	47.00	183.52	73.01	21.68
S03	ACL-R	23.25	170.18	75.92	26.21
S06	ACL-R	46.92	182.25	90.18	27.15
S09	ACL-R	25.50	193.04	88.69	23.80
S10	ACL-R	24.42	171.45	75.33	25.63
S15	ACL-R	19.33	184.15	80.99	23.88
S17	ACL-R	21.00	195.58	103.21	26.98
S18	ACL-R	21.83	172.72	74.21	24.88
S20	ACL-R	24.33	185.42	105.49	30.68

Table AC-1B. Subject functional injury status scores.

Subject Code	Injury Diagnosis	No. Years Injured	Instability Episodes	Lysholm Score Transformed	Tegner		KOS (Transformed)		IKDC Subjective	Knee Brace Use
					Pre-Injury	Post Injury	ADL	Sports		
S02	MRI	7.92	4	30.53	9	5	58.57	52.05	39.14	Y
S04	MRI	0.17	6	62.11	10	4	70.00	61.64	50.17	Y
S05	MRI	0.17	0	92.63	9	6	91.43	80.82	72.21	Y
S07	MRI	0.50		85.26	10	8	84.29	58.90	47.02	YNN
S08	Manual	0.25								Y
S11	MRI	0.33	0	92.63	9	9	100.00	86.30	78.51	N
S12	MRI	0.08	4	69.47	6	2	81.43	45.21	31.27	Y
S14	Manual	7.33	4	44.21	9	4	87.14	57.53	45.44	N
S16	Arthrogram	29.17	0	94.74	9	7	100.00	90.41	83.23	N
S03	Arthrogram	10.25	0	100.00	10	10	100.00	100.00		YNN
S06	MRI	6.50	2	95.79	6	6	92.86	82.19	73.78	YNN
S09	MRI	3.00	4	95.79	9	8	91.43	82.19	73.78	Y
S10	MRI	1.42	0	97.89	9	9	100.00	95.89	89.53	N
S15	MRI	4.50	0	100.00	9	9	100.00	97.26	91.10	YNN
S17	MRI	0.42	0	86.32	9	4	84.29	57.53	45.44	N
S18	MRI/CT	2.08	0	89.47	9	9	95.71	94.52	87.95	Y
S20	MRI	8.08	0	94.74	9	9	97.14	94.52	87.95	N

## APPENDIX D

## FUNCTIONAL MOMENT ARM AND EMG DATA

**Table AD-1A.** Functional moment arm and average surface EMG data presented for the knee extensors. Maximum force capacity (Fmax) and optimal length (rlen) of each muscle is shown. Significant interaction and main effects ( $p < 0.05$ ) are displayed for the EMG (%MVC) of each muscle.

		Functional Moment Arm (rz)						Average Surface EMG		
		Fmax (N) = 1871, rlen (cm) = 8.40						stability, knee angle		
		Control		No Brace		Brace				
Surface	Knee Angle	r	r <sup>2</sup>	r	r <sup>2</sup>	r	r <sup>2</sup>	Control	No Brace	Brace
Flat	5-10	-5.80	33.64	-5.70	32.49	-5.60	31.36	3.79	6.39	6.49
	20-25	-5.20	27.04	-5.20	27.04	-5.10	26.01	8.43	10.48	12.21
	40-45	-4.70	22.09	-4.70	22.09	-4.60	21.16	18.02	20.22	23.39
Unstable	5-10	-5.80	33.64	-5.70	32.49	-5.60	31.36	9.95	11.03	10.76
	20-25	-5.20	27.04	-5.20	27.04	-5.10	26.01	12.46	14.17	14.49
	40-45	-4.70	22.09	-4.70	22.09	-4.60	21.16	20.88	23.31	26.32
		Fmax (N) = 1365, rlen (cm) = 8.70						leg, stability, knee angle		
		Control		No Brace		Brace				
Surface	Knee Angle	r	r <sup>2</sup>	r	r <sup>2</sup>	r	r <sup>2</sup>	Control	No Brace	Brace
Flat	5-10	-5.30	28.09	-5.30	28.09	-5.20	27.04	2.984	5.285	5.437
	20-25	-4.90	24.01	-4.80	23.04	-4.80	23.04	6.61	8.334	10.627
	40-45	-4.60	21.16	-4.50	20.25	-4.50	20.25	15.192	19.478	21.554
Unstable	5-10	-5.30	28.09	-5.30	28.09	-5.20	27.04	7.703	9.264	9.097
	20-25	-4.90	24.01	-4.80	23.04	-4.80	23.04	10.042	11.54	12.699
	40-45	-4.60	21.16	-4.50	20.25	-4.50	20.25	18.358	22.366	24.618
		Fmax (N) = 1294, rlen (cm) = 8.90						leg, stability, knee angle		
		Control		No Brace		Brace				
Surface	Knee Angle	r	r <sup>2</sup>	r	r <sup>2</sup>	r	r <sup>2</sup>	Control	No Brace	Brace
Flat	5-10	-5.40	29.16	-5.30	28.09	-5.20	27.04	2.18	4.18	4.38
	20-25	-4.90	24.01	-4.80	23.04	-4.80	23.04	4.79	6.19	9.05
	40-45	-4.50	20.25	-4.40	19.36	-4.40	19.36	12.36	18.73	19.72
Unstable	5-10	-5.40	29.16	-5.30	28.09	-5.20	27.04	5.45	7.50	7.43
	20-25	-4.90	24.01	-4.80	23.04	-4.80	23.04	7.63	8.91	10.91
	40-45	-4.50	20.25	-4.40	19.36	-4.40	19.36	15.84	21.42	22.91
		Fmax (N) = 779, rlen (cm) = 8.40						stability, knee angle		
		Control		No Brace		Brace				
Surface	Knee Angle	r	r <sup>2</sup>	r	r <sup>2</sup>	r	r <sup>2</sup>	Control	No Brace	Brace
Flat	5-10	-5.90	34.81	-5.80	33.64	-5.70	32.49	2.86	4.38	4.42
	20-25	-5.40	29.16	-5.40	29.16	-5.30	28.09	3.59	4.98	5.86
	40-45	-5.00	25.00	-5.00	25.00	-4.90	24.01	5.27	7.67	7.87
Unstable	5-10	-5.80	33.64	-5.80	33.64	-5.70	32.49	4.58	6.63	6.02
	20-25	-5.40	29.16	-5.40	29.16	-5.30	28.09	4.83	5.81	6.12
	40-45	-5.00	25.00	-5.00	25.00	-4.90	24.01	6.90	8.45	8.51
		Fmax (N) = 155, rlen (cm) = 9.50						stability		
		Control		No Brace		Brace				
Surface	Knee Angle	r	r <sup>2</sup>	r	r <sup>2</sup>	r	r <sup>2</sup>	Control	No Brace	Brace
Flat	5-10	-1.10	1.21	-1.10	1.21	-1.00	1.00	5.82	6.56	5.20
	20-25	-0.90	0.81	-0.90	0.81	-0.80	0.64	4.92	5.58	5.25
	40-45	-0.70	0.49	-0.60	0.36	-0.60	0.36	5.46	5.64	5.70
Unstable	5-10	-1.10	1.21	-1.10	1.21	-1.00	1.00	7.63	7.92	6.41
	20-25	-0.90	0.81	-0.90	0.81	-0.80	0.64	7.41	6.48	5.03
	40-45	-0.70	0.49	-0.60	0.36	-0.60	0.36	6.55	6.71	6.72

**Table AD-1B.** Functional moment arm and average surface EMG data presented for the knee flexors. Maximum force capacity (Fmax) and optimal length (rlen) of each muscle is shown. Significant interaction and main effects ( $p < 0.05$ ) are displayed for the EMG (%MVC) of each muscle.

		Functional Moment Arm (rz)						Average Surface EMG		
		Fmax (N) = 1113, rlen (cm) = 4.50						leg, stability, knee angle		
		Control		No Brace		Brace		stability*knee angle		
Surface	Knee Angle	r	r <sup>2</sup>	r	r <sup>2</sup>	r	r <sup>2</sup>	Control	No Brace	Brace
Flat	5-10	1.90	3.61	1.90	3.61	1.90	3.61	16.41	12.77	10.55
	20-25	1.90	3.61	1.90	3.61	1.90	3.61	10.76	7.97	5.90
	40-45	1.80	3.24	1.80	3.24	1.80	3.24	7.48	5.38	4.42
Unstable	5-10	1.90	3.61	1.90	3.61	1.90	3.61	23.39	19.14	15.10
	20-25	1.90	3.61	1.90	3.61	1.90	3.61	18.08	13.48	9.47
	40-45	1.80	3.24	1.80	3.24	1.80	3.24	10.33	7.21	6.35
		Fmax (N) = 488, rlen (cm) = 6.40						knee angle, leg*knee angle		
		Control		No Brace		Brace		Control	No Brace	Brace
Surface	Knee Angle	r	r <sup>2</sup>	r	r <sup>2</sup>	r	r <sup>2</sup>	Control	No Brace	Brace
Flat	5-10	2.00	4.00	2.00	4.00	2.00	4.00	10.41	8.93	8.46
	20-25	2.00	4.00	2.00	4.00	2.00	4.00	6.30	7.90	8.20
	40-45	1.90	3.61	1.90	3.61	1.90	3.61	7.00	7.83	8.94
Unstable	5-10	2.00	4.00	2.00	4.00	2.00	4.00	11.46	11.77	10.40
	20-25	2.00	4.00	2.00	4.00	2.00	4.00	8.19	8.60	8.33
	40-45	1.90	3.61	1.90	3.61	1.90	3.61	7.46	8.78	9.43
		Fmax (N) = 1030, rlen (cm) = 8.00						stability, knee angle		
		Control		No Brace		Brace		leg*knee angle		
Surface	Knee Angle	r	r <sup>2</sup>	r	r <sup>2</sup>	r	r <sup>2</sup>	Control	No Brace	Brace
Flat	5-10	3.20	10.24	3.30	10.89	3.50	12.25	5.63	4.59	3.36
	20-25	4.10	16.81	4.20	17.64	4.40	19.36	3.02	2.34	2.72
	40-45	4.90	24.01	5.00	25.00	5.10	26.01	1.96	2.04	2.72
Unstable	5-10	3.30	10.89	3.30	10.89	3.50	12.25	10.48	7.91	6.30
	20-25	4.20	17.64	4.20	17.64	4.30	18.49	7.84	5.75	4.52
	40-45	5.00	25.00	5.00	25.00	5.10	26.01	6.73	5.67	5.69
		Fmax (N) = 328, rlen (cm) = 20.10						stability, knee angle		
		Control		No Brace		Brace		leg*knee angle		
Surface	Knee Angle	r	r <sup>2</sup>	r	r <sup>2</sup>	r	r <sup>2</sup>	Control	No Brace	Brace
Flat	5-10	3.90	15.21	4.00	16.00	4.10	16.81	5.63	4.59	3.36
	20-25	4.80	23.04	4.90	24.01	5.00	25.00	3.02	2.34	2.72
	40-45	5.60	31.36	5.60	31.36	5.70	32.49	1.96	2.04	2.72
Unstable	5-10	4.00	16.00	4.00	16.00	4.20	17.64	10.48	7.91	6.30
	20-25	4.80	23.04	4.90	24.01	5.00	25.00	7.84	5.75	4.52
	40-45	5.60	31.36	5.60	31.36	5.80	33.64	6.73	5.67	5.69
		Fmax (N) = 717, rlen (cm) = 10.90						stability, stability*knee angle		
		Control		No Brace		Brace		Control	No Brace	Brace
Surface	Knee Angle	r	r <sup>2</sup>	r	r <sup>2</sup>	r	r <sup>2</sup>	Control	No Brace	Brace
Flat	5-10	2.20	4.84	2.40	5.76	2.50	6.25	2.97	4.95	3.09
	20-25	3.50	12.25	3.60	12.96	3.80	14.44	3.09	4.18	3.96
	40-45	4.70	22.09	4.80	23.04	5.10	26.01	4.13	6.04	5.80
Unstable	5-10	2.30	5.29	2.30	5.29	2.60	6.76	6.67	8.13	6.00
	20-25	3.50	12.25	3.60	12.96	3.80	14.44	6.12	5.67	5.13
	40-45	4.80	23.04	4.90	24.01	5.10	26.01	6.64	6.89	6.60
		Fmax (N) = 402, rlen (cm) = 17.30						stability, stability*knee angle		
		Control		No Brace		Brace		Control	No Brace	Brace
Surface	Knee Angle	r	r <sup>2</sup>	r	r <sup>2</sup>	r	r <sup>2</sup>	Control	No Brace	Brace
Flat	5-10	0.50	0.25	0.60	0.36	0.60	0.36	2.97	4.95	3.09
	20-25	1.00	1.00	1.10	1.21	1.20	1.44	3.09	4.18	3.96
	40-45	1.60	2.56	1.70	2.89	1.80	3.24	4.13	6.04	5.80
Unstable	5-10	0.50	0.25	0.50	0.25	0.60	0.36	6.67	8.13	6.00
	20-25	1.10	1.21	1.10	1.21	1.20	1.44	6.12	5.67	5.13
	40-45	1.60	2.56	1.70	2.89	1.80	3.24	6.64	6.89	6.60

## REFERENCES

- Aalbersberg-van Berkel, S. (2006) *Knees in Need: Neuromuscular control of the ACL deficient knee*. PhD Dissertation, Amsterdam.
- Acierno, S.P., D'Ambrosia, C., Solomonow, M., Baratta, R.V., and D'Ambrosia, R. (1995) Electromyographic and biomechanics of a dynamic knee brace for anterior cruciate ligament deficiency. *Orthopedics* **18**, 1101-1107.
- Ageberg, E., Zatterstrom, R., Moritz, U., and Frieden, T. (2001) Influence of supervised and nonsupervised training on postural control after an acute anterior cruciate ligament rupture: A three-year longitudinal prospective study. *Journal of Orthopaedic and Sports Physical Therapy*. **31**, 632-644.
- Akima, H., Takahashi, H., Kuno, S., and Katsuta, S. (2004) Coactivation pattern in human quadriceps during isokinetic knee-extension by muscle functional MRI. *European Journal of Applied Physiology* **91**, 7-14.
- Alkner, B.A., Tesch, P.A., and Berg, H.E. (2000) Quadriceps and EMG/force relationship in knee extension and leg press. *Medicine and Science in Sports & Exercise* **32**, 459-463.
- Allen, C.R., Wong, E.K., Livesay, G.A., Sakane, M., Fu, F.H., and Woo, S.L. (2000) Importance of the medial meniscus in the anterior cruciate ligament in the anterior cruciate ligament-deficient knee. *Journal of Orthopaedic Research* **18**, 115.
- Amis, A.A. and Dawkins, P.C. (1991) Functional anatomy of the anterior cruciate ligament: Fibre bundle actions related to ligament replacements and injuries. *The Journal of Bone and Joint Surgery*. **73B**, 260-267.
- Arnockzky, S.P. (1983) Anatomy of the anterior cruciate ligament. *Clinical Orthopaedics and Related Research*. **172**, 19-25.
- Baratta, R., Solomonow, M., Zhou, B.H., Letson, D., Chuinard, R., and D'Ambrosia, R. (1988) The role of the antagonist musculature in maintaining knee stability. *The American Journal of Sports Medicine*. **16**, 113-122.
- Barrack, R.L., Skinner, H.B., and Buckley, S.L. (1989) Proprioception in the anterior cruciate deficient knee. *The American Journal of Sports Medicine*. **17**, 1-6.
- Basmajian, J.V. (1985) *Muscles alive, their functions revealed by electromyography*. Williams & Wilkins, Baltimore, MD.
- Beard, D.J., Kyberd, P.J., Fergusson, C.M., and Dodd, C.A.F. (1993) Proprioception after rupture of the anterior cruciate ligament. *Journal of Bone and Joint Surgery (British Edition)*. **75B**, 311-315.

- Beard,D.J., Kyberd,P.J., O'Connor,J.J., Fergusson,C.M., and Dodd,C.A. (1994) Reflex hamstring contraction latency in anterior cruciate ligament deficiency. *Journal of Orthopaedic Research*. **12**, 219-228.
- Beck,C., Drez,D., Young,J., Cannon,W.D., and Stone,M.L. (1986) Instrumented testing of functional knee braces. *The American Journal of Sports Medicine*. **14**, 253-256.
- Berchuck,M., Andriacchi,T.P., Bach,B.R., and Reider,B. (1990) Gait adaptations by patients who have a deficient anterior cruciate ligament. *The Journal of Bone and Joint Surgery*. **72A**, 871-877.
- Bergmark,A. (1989) Stability of the lumbar spine: A study in mechanical engineering. *Acta Orthopaedica Scandinavica* **60**, 1-54.
- Beynnon,B.D., Good,L., and Risberg,M.A. (2002) The effect of bracing on proprioception of knee with anterior cruciate ligament injury. *Journal of Orthopaedic and Sports Physical Therapy* **32**, 11-15.
- Beynnon,B.D., Peura,G.D., Johnson,R.J., Renstrom,P.A., Nichols,C.E., and Pope,M.H. An in-vivo investigation of anterior cruciate ligament strain: The effect of functional knee bracing and attachment strap tension.  
Ref Type: Internet Communication; [http://www.donjoy.com/rd/reference/recent\\_publications/bio\\_labinvivo.html](http://www.donjoy.com/rd/reference/recent_publications/bio_labinvivo.html), Retrieved June 2004.
- Beynnon,B.D., Pope,M.H., Wertheimer,C.M., Johnson,R.J., Fleming,B.C., Nichols,C.E., and Howe,J.G. (1992) The effect of functional knee-braces on strain on the anterior cruciate ligament in vivo. *The Journal of Bone and Joint Surgery*. **74A**, 1298-1312.
- Bigland-Ritchie,B. (1981) EMG/force relations and fatigue of human voluntary contractions. *Exercise Sport Science Review* **9**, 75-117.
- Bledsoe,G.R. Dynamic bracing and its effect on the knee.  
Ref Type: Internet Communication; [http://bledsoebrace.com/pdf/CP030014\\_A.pdf](http://bledsoebrace.com/pdf/CP030014_A.pdf), Retrieved: June 2004
- Bodor, M. (2001) Quadriceps protects the anterior cruciate ligament. *Journal of Orthopaedic Research*, **19**, 629-633.
- Boyd,I.A. (1954) The histological structure of the receptors in the knee-joint of the cat correlated with their physiological response. *Journal of Physiology* **124**, 476-488.
- Branch,T., Hunter,R., and Reynolds,P. (1988) Controlling anterior tibial displacement under static load: A comparison of two braces. *Orthopedics* **11**, 1249-1252.
- Branch,T.P., Hunter,R., and Donath,M. (1989) Dynamic EMG analysis of anterior cruciate ligament legs with and without bracing during cutting. *The American Journal of Sports Medicine* **17**, 35-41.



- Brand,R.A., Crowninshield,R.D., Wittstock,C.E., Pedersen,D.R., and Clark,C.R. (1982) A model of lower extremity muscular anatomy. *Journal of Biomechanical Engineering* **104**, 304-310.
- Brantigan,O.C. and Voshell,A.F. (1941) The mechanics of the ligaments and menisci of the knee joint. *The Journal of Bone and Joint Surgery* **23**, 44-66.
- Butler,D.L., Noyes,F.R., and Grood,E.S. (1980) Ligamentous restraints to anterior-posterior drawer in the human knee. *The Journal of Bone and Joint Surgery* **62A**, 259-270.
- Carter,N.D., Jenkinson,T.R., Wilson,D., Jones,D.W., and Torode,A.S. (1997) Joint position sense and rehabilitation in the anterior cruciate ligament deficient knee. *British Journal of Sports Medicine*. **31**, 209-212.
- Cawley,P.W., France,E.P., and Paulos,L.E. (1991) The current state of functional knee bracing research: A review of literature. *The American Journal of Sports Medicine*. **19**, 226-233.
- Chmielewski,T.L., Rudolph,K.S., and Snyder-Mackler,L. (2002) Development of dynamic knee stability after acute ACL injury. *Journal of Electromyography and Kinesiology* **12**, 267-274.
- Cholewicki,J., Juluru,K., and McGill,S.M. (1999) Intra-abdominal pressure mechanism for stabilizing the lumbar spine. *Journal of Biomechanics* **32**, 13-17.
- Cholewicki,J. and McGill,S.M. (1996) Mechanical stability of the in vivo lumbar spine: implications for injury and chronic low back pain. *Clinical Biomechanics* **11**, 1-15.
- Ciccotti,M., Kerlan,R.K., and Perry,J. (1994) An electromyographic analysis of the knee during functional activities. *The American Journal of Sports Medicine* **22**, 651-658.
- Colville,M.R., Lee,C.L., and Ciullo,J.V. (1986) The Lenox Hill brace: An evaluation of effectiveness in treating knee instability. *The American Journal of Sports Medicine* **14**.
- Cram,J.R. (1998) *Introduction to surface electromyography*. Aspen Publishers, Gaithersburg, Maryland.
- Crisco III,J.J. and Panjabi,M.M. (1990) Postural biomechanical stability and gross muscular architecture in the spine. In *Multiple muscle systems: Biomechanics and movement organization* (Edited by Winters,J.M. and Woo,S.L.) Springer-Verlag, New York.
- Crisco,J.J. and Panjabi,M.M. (1991) The intersegmental and multisegmental muscles of the lumbar spine: A biomechanical model comparing lateral stabilizing potential. *Spine* **16**, 793-799.

- Crowninshield,R.D. and Brand,R.A. (1981) A physiologically based criterion of muscle force prediction in locomotion. *Journal of Biomechanics* **14**, 793-801.
- Daniel,D.M., Akeson,W.H., and O'Connor,J.J. (1990) *Knee Ligaments: Structure, function, injury and repair*. Raven Press, New York.
- Daniel,D.M., Stone,M.L., Dobson,B.E., Fithian,D.C., Rossman,D.J., and Kaufman,K.R. (1994) Fate of the Acl-Injured Patient - A Prospective Outcome Study. *The American Journal of sports medicine* **22**, 632-644.
- Derouin, A.J. and Potvin, J.R. (2005) Knee stability: Mechanical contributions of individual muscles. *International Society of Biomechanics World Congress XXth Conference Proceedings*.
- Delp,S.L., Loan,J.P., Hoy,M.G., Zajac,F.E., Topp,E.L., and Rosen,J.M. (1990) An Interactive Graphics-Based Model of the Lower-Extremity to Study Orthopedic Surgical-Procedures. *IEEE Transactions on Biomedical Engineering* **37**, 757-767.
- DeLuca,C.J. Surface electromyography: Detection and recording. 1996.  
Ref Type: Internet Communication;  
[http://www.delsys.com/Attachments\\_pdf/WP\\_SEMGintro.pdf](http://www.delsys.com/Attachments_pdf/WP_SEMGintro.pdf), Retrieved: June 2004
- Devita, P., Hortobagyi, T. and Barrier, J. (1998) Gait biomechanics are not normal after anterior cruciate ligament reconstruction and accelerated rehabilitation. *Medicine and Science in Sports and Exercise*, **30**, 1481-1488.
- Diaz,G.Y., Averett,D.H., and Soderberg,G.L. (1997) Electromyographic analysis of selected lower extremity musculature in normal subjects during ambulation with and without a Protonics knee brace. *Journal of Orthopaedic and Sports Physical Therapy*, **26**, 292-298.
- Dienst,M., Burks,R.T., and Greis,P.E. (2002) Anatomy and biomechanics of the anterior cruciate ligament. *Orthopaedic Clinics of North America*. **33**, 605-620.
- Doorenbosch,C.A.M. and Harlaar,J. (2003) A clinically applicable EMG-force model to quantify active stabilization of the knee after a lesion of the anterior cruciate ligament. *Clinical Biomechanics* **18**, 142-149.
- Eastlack,M.E., Axe,M.J., and Synder-Mackler,L. (1999) Laxity, instability, and functional outcome after ACL injury: Copers versus noncopers. *Medicine and Science in Sports*. **31**, 210-215.
- Ekholm,J., Eklund,G., and Skoglund,S. (1960) On the reflex effects from the knee joint of the cat. *Acta Physiologica Scandinavica*. **50**, 167-174.
- Ellison,A.E. and Berg,E.E. (1985) Embryology, Anatomy, and Function of the cruciate ligament. *Orthopaedic Clinics of North America* **16**, 3-14.

- Evans, N.A. and Stannish, W.D. (2001) The natural history and tailored treatment of ACL injury. *The Physician and Sportsmedicine*. **29**, 19-24.
- Feagin, J. and Curl, W. (1976) Isolated tear of the anterior cruciate ligament: 5 year follow-up study. *The American Journal of Sports Medicine* **4**, 95-100.
- Feagin, J. and Lambert, M. (1985) Mechanisms of injury and pathology of anterior cruciate ligament injuries. *Orthopaedic Clinics of North America*. **16**, 41-45.
- Fink, C., Hoser, C., Hack, W., Navarro, R.A., and Benedetto, K.P. (2001) Long-term outcome of operative or non-operative treatment of anterior cruciate ligament rupture: Is sports activity a determining variable. *International Journal of Sports Medicine*. **22**, 304-309.
- Fitzgerald, G.K., Axe, M.J., and Snyder-Mackler, L. (2000) A decision-making scheme for returning patients to high-level activity with nonoperative treatment after anterior cruciate ligament rupture. *Knee Surg, Sports Traumatol, Arthrosoc* **8**, 82.
- Fitzgerald, G.K., Axe, M.J., and Snyder-Mackler, L. (2000a) Proposed practice guidelines for non-operative anterior cruciate ligament rehabilitation of physically active individuals. *Journal of Orthopaedic and Sports Physical Therapy*. **30**, 194-203.
- Fitzgerald, G.K., Axe, M.J., and Snyder-Mackler, L. (2000b) The efficacy of perturbation training in nonoperative anterior cruciate ligament rehabilitation programs for the physically active individuals. *Physical Therapy*. **80**, 151.
- Fleming, B.C., Renstrom, P.A., Ohlen, G., Johnson, R.J., Peura, G.D., Beynon, B.D., and Badger, G.J. (2001) The gastrocnemius muscle is an antagonist of the anterior cruciate ligament. *Journal of Orthopaedic Research*, **19**, 1178-1184.
- Flynn, J.M., Holmes, J.D., and Andrews, D.M. (2004) The effect of localized leg muscle fatigue on tibial impact acceleration. *Clinical Biomechanics* **19**, 726-732.
- France, E.P., Cawley, P.W., and Paulos, L.E. (1990) Choosing functional knee braces. *Clinical Sports Medicine* **9**, 743.
- France, E.P. and Paulos, L.E. (1994) Knee Bracing. *Journal of American Academy of Orthopaedic Surgeons*. **2**, 281-287.
- Frank, C.B. and Jackson, D.W. (1997) The science of reconstruction of the anterior cruciate ligament. *Journal of Bone and Joint Surgery (Am)* **79**, 1556-1576.
- Freeman, M.A.R. and Wyke, B. (1967) The innervation of the knee joint. An anatomical and histological study in the cat. *Journal of Anatomy* **101**, 505-532.
- Furman, W., Marshall, J.L., and Girgis, F.G. (1976) The anterior cruciate ligament. *The Journal of Bone and Joint Surgery* **58A**, 179-185.

- Fuss,F.K. (1989) Anatomy of the cruciate ligaments and their function in extension and flexion of the human knee joint. *The American Journal of Anatomy* **184**, 165-176.
- Geiringer,S.R. (1994) *Anatomic localization for needle electromyography*. Hanley & Belfaus, Philadelphia, PA.
- Giove,T.P., Miller,S.J., Kent,B.E., Sanford,T.L., and Garrick,J.G. (1983) Non-operative treatment of the torn anterior cruciate ligament. *The Journal of Bone and Joint Surgery*.
- Girgis,F.G., Marshall,J.L., and Al Monajem,A.R.S. (1975) The cruciate ligaments of the knee joint. *Clinical Orthopaedics and Related Research* **106**, 216-231.
- Goldblatt,J.P. and Richmond,J.C. (2003) Anatomy and biomechanics of the knee. *Operative Techniques in Sports Medicine* **11**, 172-186.
- Goldstein,J. and Bosco,J. (2001) The ACL-Deficient Knee: Natural history and treatment options. *Bulletin Hospital for Joint Diseases*. **60**, 173-178.
- Gollehon,D.L., Torzilli,P.A., and Warren,R.F. (1987) The role of the posterolateral and cruciate ligaments in the stability of the human knee: a biomechanical study. *Journal of Bone and Joint Surgery (Am)* **69**, 233-242.
- Grabiner,M.D., Campbell, K.R.,Hawthorne, D.L., and Hawkins, D.A. (1989) Electromyographic study of the anterior cruciate ligament-hamstrings synergy during isometric knee extension. *Journal of Orthopaedic Research*, **7**, 152-155.
- Grabiner,M.D. and Weiker,G.G. (1993) Anterior Cruciate Ligament Injury and Hamstrings Coactivation. *Clinical Biomechanics* **8**, 215-219.
- Graham,V.L., Gehlsen,G.M., and Edwards,J.A. (1993) Electromyographic evaluation of closed and open kinetic chain knee rehabilitation exercises. *Journal of Athletic Training* **28**, 23-30.
- Granata,K.P. and Orishimo,K.F. (2001) Response of trunk muscle coactivation to changes in spinal stability. *Journal of Biomechanics* **34**, 1117-1123.
- Granta,K.P. and Wilson,S.E. (2001) Trunk posture and spinal stability. *Clinical Biomechanics* **16**, 650-659.
- Gross,M.T., Tyson,A.D., and Burns,C.B. (1993) Effect of knee angle and ligament insufficiency on anterior tibial translation during quadriceps muscle contraction: A preliminary report. *Journal of Orthopaedic and Sports Physical Therapy* **17**, 133-143.
- Halata,Z. (1977) The ultrastructure of the sensory nerve endings in the articular capsule of the knee joint of the domestic cat (Ruffini corpuscles and Pacinian corpuscles). *Journal of Anatomy*, **124**, 717-729.

- Haus, J. and Halata, Z. (1990) Innervation of the anterior cruciate ligament. *International Orthopaedics*. **14**, 293-296.
- Hirokawa, S., Solomonow, M., Luo, Z., Lu, Y., and D'Ambrosia, R. (1991) Muscular co-contraction and control of knee stability. *Journal of Electromyography and Kinesiology*, **1**, 199-208.
- Hoffman, A.A., Wyatt, R.W.B., Bourne, M.H., and Daniels, A.U. (1984) Knee stability in orthotic knee braces. *The American Journal of Sports Medicine*. **12**, 371-374.
- Hong, S.H., Choi, J.-Y., Lee, G.K., Choi, J.-A., Chung, H.W., and Kang, H.S. (2003) Grading of anterior cruciate ligament injury: Diagnostic efficacy of oblique magnetic resonance imaging of the knee. *Journal of Computer Assisted Tomography* **27**, 814-819.
- Howell, S.M. and Usafr, M. (1990) Anterior tibial translation during a maximum quadriceps contraction: Is it clinically significant? *The American Journal of Sports Medicine*. **18**, 573-578.
- Iles, J.F., Stokes, M., and Young, A. (1990) Reflex Actions of Knee-Joint Afferents During Contraction of the Human Quadriceps. *Clinical Physiology* **10**, 489-500.
- Imran, A. and O'Connor, J.J. (1998) Control of knee stability after ACL injury or repair: interaction between hamstrings contraction and tibial translation. *Clinical Biomechanics* **13**, 153-162.
- Isear, J.A., Erickson, J.C. (Jr.), and Worrell, T.W. (1997) EMG analysis of lower extremity muscle recruitment patterns during an unloaded squat. *Medicine and Science in Sports and Exercise*, **29**, 532-539.
- Jackson, D.W., Arnoczky, S.P., Woo, S.L., Frank, C.B., and Simon, T.M. (1993) *The anterior cruciate ligament: Current and future concepts*. Raven Press, New York.
- Jakob, R.P. and Staubli, H.U. (1990) *The knee and the cruciate ligaments: Anatomy, biomechanics, clinical aspects, reconstruction, complications, and rehabilitation*. Springer-Verlag, Berlin.
- Johansson, H., Sjolander, P., and Sojka, P. (1991) Receptors in the knee joint ligaments and their role in the biomechanics of the joint. *Critical Reviews in Biomedical Engineering* **18**, 341-368.
- Johnson, D.S. and Smith, R.B. (2001) Outcome measurement in the ACL deficient knee: What's the score? *The Knee*. **8**, 51-57.
- Johnson, R. (1988) Prevention of cruciate ligament injuries. In *The crucial ligaments: Diagnosis and treatment of ligamentous injuries about the knee* (Edited by Feagin, J.Jr.) Pp. 349-356. Churchill Livingstone, New York.

- Kalund,S., Sinkjaer,T., Arendt-Nielsen,L., and Simonsen,O. (1990) Altered timing of hamstring muscle action in anterior cruciate ligament deficient patients. *The American Journal of Sports Medicine*. **18**, 245-248.
- Kandel E.R., Schwartz J.H., and Jessell T.M. (2000) *Principles of neuroscience*. McGraw Hill, New York.
- Kao,J.T., Giangarra,C.E., Singer,G., and Martin,S. (1995) A Comparison of Outpatient and Inpatient Anterior Cruciate Ligament Reconstruction Surgery. *Arthroscopy* **11**, 151-156.
- Kapandji,I.A. (1987) *The physiology of joints: Volume 2*. Churchill Livingstone, Edinburgh.
- Katz,J.W. and Fingerroth,R.J. (1986) The diagnostic accuracy of ruptures of the anterior cruciate ligament comparing the Lachman test, the anterior drawer sign, and the pivot shift test in acute and chronic knee injuries. *The American Journal of Sports Medicine* **14**, 88-91.
- Kean, C.O., Behm,D.G., and Young, W.B. (2006) Fixed foot balance training increases rectus femoris activation during landing and jump height in recreationally active women. *Journal of Sports Science and Medicine*, **5**, 138-148.
- Kennedy,J.C., Alexander,I.J., and Hayes,K.C. (1982) Nerve supply of the knee and its functional importance. *The American Journal of Sports Medicine* **10**, 329-335.
- Kirkendall,A.L., Lopez,J.M., and Gonzalez,R.V. (2002) A kinematic analysis on an ACL deficient knee. *Advances in Bioengineering (BED)*. **53**, 1-2.
- Kvist,J. and Gillquist,J. (2001) Sagittal plane knee translation and electromyographic activity during closed and open kinetic chain exercises in anterior cruciate ligament-deficient patients and control subjects. *The American Journal of Sports Medicine* **29**, 72-82.
- Larson,R.L. and Grana,W.A. (1993) *The knee: Form, function, pathology, and treatment*. W.B. Saunders Company, Philadelphia, PA.
- Laurence,M. and Strachan,J.C.H. (1970) The dynamic stability of the knee. *Proceedings of the Royal Society of Medicine* **63**, 34-35.
- Lehmkuhl,L.D. and Smith,L.K. (1983) *Brunnstroms clinical kinesiology*. F.A. Davis Company, Philadelphia.
- Liggins,A.B. and Bowker,P. (1991) A quantitative assessment of orthoses for stabilization of the anterior cruciate ligament deficient knee. *Proceedings of the Institution of Mechanical Engineers (Part H)*. **205**, 81-87.

- Limbird,T.J., Shiavi,R., Frazer,M., and Borra,H. (1988) EMG profiles of knee joint musculature during walking: Changes induced by anterior cruciate ligament deficiency. *Journal of Orthopaedic Research* **6**, 630-638.
- Liu,S.H., Lunsford,T., Gude,S., and Vangsness,T. (1994) Comparison of functional knee braces for control of anterior tibial displacement. *Clinical Orthopaedics and Related Research*. **303**, 203-210.
- Liu,W. and Maitland,M.E. (2000) The effect of hamstring muscle compensation for anterior laxity in the ACL-deficient knee during gait. *Journal of Biomechanics* **33**, 871-879.
- Lloyd,D.G. and Besier,T.F. (2003) An EMG-driven musculoskeletal model to estimate muscle forces and knee joint moments in vivo. *Journal of Biomechanics* **36**, 765-776.
- Lloyd,D.G. and Buchanan,T.S. (1996) A model of load sharing between muscles and soft tissues at the human knee during static tasks. *Journal of Biomechanical Engineering* **118**, 367-376.
- Loosli,A. and Oshimo,T. (1994) Conservatively treated anterior cruciate ligament injuries: activities, symptoms, and roentgenographic changes. *Clinical Journal of Sports Medicine*. **4**, 232.
- Lu,T.-W., O'Connor,J.J., Taylor,S.J.G., and Walker,P.S. (1998) Validation of a lower limb model with in vivo femoral forces telemetered from two subjects. *Journal of Biomechanics* **31**, 63-69.
- Lysholm,J. and Gillquist,J. (1982) Evaluation of knee ligament surgery results with special emphasis on use of a scoring scale. *The American Journal of Sports Medicine*. **10**, 150-154.
- Mackenzie,R. and Dixon,A. (2001) Knee anatomy and MRI. *Clinical MRI/Developments in MR* **11**, 65-70.
- MacWilliams,B.A., Wilson,D.R., DesJardins,J.D., Romero,J., and Chao,E. Y.S. (1999) Hamstrings cocontraction reduces internal rotation, anterior translation and anterior cruciate ligament load in weight-bearing flexion. *Journal of Orthopaedic Research*. **17**, 817-822.
- Marans,H.J., Jackson,R.W., Glossop,N.D., and Young,C. (1989) Anterior cruciate ligament insufficiency: A dynamic three-dimensional motion analysis. *The American Journal of Sports Medicine*. **17**, 325-332.
- Markhof,K.L., Graff-Radford,A., and Amstutz,H.C. (1978) In vivo knee stability: A quantitative assessment using an instrumented clinical testing apparatus. *The Journal of Bone and Joint Surgery* **60A**.

- Markolf, K.L., Kochan, A., and Amstutz, H.C. (1986) Measurement of knee stiffness and laxity in patients with documented absence of the anterior cruciate ligament. *Journal of Bone and Joint Surgery* **58A**, 583.
- Marras, W.S. and Sommerich, C.M. (1991) A three dimensional motion model of loads on the lumbar spine: I. Model Structure. *Human Factors* **33**, 123-137.
- Marx, R.G., Jones, E.C., Wickiewicz, T.L., and Warren, R.F. (2003) Beliefs and attitudes of members of the American Academy of Orthopaedic Surgeons regarding the treatment of anterior cruciate ligament injury. *Arthroscopy: The Journal of Arthroscopic and Related Surgery*. **19**, 762-770.
- Matsumoto, H. and Seedhom, B.B. (1993) Three-dimensional analysis of knee joint movement with biplanar photography, with special reference to the analysis dynamic knee instabilities. *Journal of Engineering in Medicine*. **207**, 163-173.
- Miyasaka, K.C., Daniel, D.M., and Stone, M.L. (1991) The incidence of knee ligament injuries in the general population. *American Journal of Knee Surgery* **4**, 3-8.
- Moore, K.L. (1992) *Clinically oriented anatomy*. Williams & Wilkins, Baltimore, MD.
- Moore, S.L. (2002) Imaging the anterior cruciate ligament. *Orthopedic Clinics of North America*. **33**, 674.
- Muller, W. (1983) *The knee: Form, function, and ligament reconstruction*. Springer-Verlag, Berlin.
- Nichols, T.R. (1994) A biomechanical perspective on spinal mechanisms of coordinated muscular action: an architecture principle. *Acta Anatomica* **151**, 1-13.
- Nemeth, G., Lamontagne, M., Tho, K.S., and Eriksson, E. (1997) Electromyographic activity in expert downhill skiers using functional knee braces after anterior cruciate ligament injuries. *The American Journal of Sports Medicine* **25**, 635-641.
- Newman, S.A., Jones, G., and Newham, D.J. (2003) Quadriceps voluntary activation at different joint angles measured by two stimulation techniques. *European Journal of Applied Physiology* **89**, 496-499.
- Noyes, F.R., Barber, S.D., and Mangine, R.E. (1991) Abnormal Lower-Limb Symmetry Determined by Function Hop Tests After Anterior Cruciate Ligament Rupture. *The American Journal of Sports Medicine* **19**, 513-518.
- Noyes, F.R., Grood, E.S., Butler, D.L., and Malek, M. (1980) Clinical laxity tests and functional stability of the knee: Biomechanical concepts. *Clinical Orthopaedics and Related Research* **146**, 84-89.



Noyes,F.R., Mooar,P.A., Matthews,D.S., and Butler,D.L. (1983a) The symptomatic anterior cruciate-deficient knee: Part I: The long-term functional disability in athletically active individuals. *The Journal of Bone and Joint Surgery* **65A**, 154-162.

Noyes,F.R., Mooar,P.K., Matthews,D.S., and Grood,E.S. (1983b) The symptomatic anterior cruciate-deficient knee. Part II: The results of rehabilitation, activity modification, and counseling on functional disability.

Noyes,F.R., Schipplein,O.D., Andriacchi,T.P., Saddemi,S.R., and Weise,M. (1992) The Anterior Cruciate Ligament-Deficient Knee with Varus Alignment - An Analysis of Gait Adaptations and Dynamic Joint Loadings. *The American Journal of Sports Medicine* **20**, 707-716.

O'Connor,J.J. (1993) Can muscle co-contraction protect knee ligaments after injury or repair? *The Journal of Bone and Joint Surgery (British)*. **75B**, 41-48.

Odensten,M. and Gillquist,J. (1985) Functional anatomy of the anterior cruciate ligament and a rationale for reconstruction. *Journal of Bone and Joint Surgery* **67A**, 257-262.

Ott,J.W. and Clancy,W.G., Jr. (1993) Functional knee braces. *Orthopedics* **16**, 171-176.

Paluska,S.A. (2000) Knee braces: Current evidence and clinical recommendations for their use. *American Family Physician* **61**, 411.

Panjabi,M.M. (1992) The stabilizing system of the spine. Part 1. Function, dysfunction, adaptation, and enhancement. *Journal of Spinal Disorders* **5**, 383-389.

Perotto,A.O., Delagi,E.F., Iazzetti,J., and Morrison,D. (1994) *Anatomical guide for the electromyographer*. Charles C Thomas Pub Ltd, Springfield, MO.

Podesta,L. and Sherman,M.F. (1988) Knee bracing. *Orthopedic Clinics of North America*. **19**, 737-745.

Pope,T.L. (1994) MRI of knee ligaments. *Seminars in Ultrasound, CT, and MRI* **15**, 366-382.

Potvin,J.R. and Brown,S.H.M. (2005) An equation to calculate individual muscle contributions to joint stability. *Journal of Biomechanics*. **38**, 973-980.

Ramsey,D.K., Wretenberg,P.F., Lamontagne,M., and Nemeth,G. (2003) Electromyographic and biomechanic analysis of anterior cruciate ligament deficiency and functional knee bracing. *Clinical Biomechanics* **18**, 28-34.

Regalbuto,M.A., Rovick,J.S., and Walker,P.S. (1989) The forces in a knee brace as a function of hinge design and placement. *The American Journal of Sports Medicine*. **17**, 535-543.

- Reider,B., Areand,M.A., Diehl,L.H., Mroczek,K., Abulencia,A., Stroud,C.C., Palm,M., Gilbertson,J., and Staszak,P. (2003) Proprioception of the knee before and after anterior cruciate ligament reconstruction. *Arthroscopy: The Journal of Arthroscopy and Related Surgery* **19**, 2-12.
- Roberts,D., Friden,T., Stomberg,A., Lindstrand,A., and Moritz,U. (2000) Bilateral proprioceptive defects in patients with a unilateral anterior ligament reconstruction: A comparison between patients and healthy individuals. *Journal of Orthopaedic Research*. **18**, 565-571.
- Roberts,D., Friden,T., Zatterstrom,R., Lindstrand,A., and Moritz,U. (1999) Proprioception in people with anterior cruciate ligament-deficient knees: Comparison of symptomatic and asymptomatic patients. *Journal of Orthopaedic and Sports Physical Therapy*. **29**, 587-594.
- Rozendaal,L.A. (1997) On the quasi-static and dynamic stability of the shoulder. Delft University of Technology Publication. *Proceedings of the First Conference of the ISG*.
- Rudolph,K.S., Axe,M.J., and Snyder-Mackler,L. (2000) Dynamic stability after ACL injury: who can hop? *Knee Surg.Sports Traumatol Arthrosc.* **8**, 262-269.
- Ryder,S.H., Johnson,R.J., Beynnon,B.D., and Ettliger,C.F. (1997) Prevention of ACL injuries. *Journal of Sports Rehabilitation* **6**, 80-96.
- Saitou,K., Masuda,T., Michikami,D., Kojima,R., and Okada,M. (2000) Innervation zones of the upper and lower limb muscles estimated by using multichannel surface EMG. *Journal of Human Ergology* **29**, 35-52.
- Schultz,R.A., Miller,D.C., Kerr,C.S., and Micheli,L. (1984) Mechanoreceptors in human cruciate ligaments. *The Journal of Bone and Joint Surgery* **66A**, 1072-1076.
- Schutte,M.J., Dabezies,E.J., Zimny,M.L., and Happel,L.T. (1987) Neural anatomy of the human anterior cruciate ligament. *The Journal of Bone and Joint Surgery* **69A**, 243-247.
- Segawa,H., Omori,G., and Koga,Y. (2001) Long-term results of non-operative treatment of anterior cruciate ligament injury. *The Knee* **8**, 5-11.
- Sinkjaer,T. and Arendt-Nielson,L. (1991) Knee stability and muscle coordination patterns in patients with anterior cruciate ligament injuries: An electromyographic approach. *Journal of Electromyography and Kinesiology* **1**, 209-217.
- Smith,B.A., Livesay,G.A., and Woo,S.L.Y. (1993) Biology and biomechanics of the anterior cruciate ligament. *Clinics in Sport Medicine* **12**, 637-670.
- Smith,J., Malanga,G.A., Yu,B., and An,K.-N. (2003) Effects of functional knee bracing on muscle-firing patterns about the chronic anterior cruciate ligament-deficient knee. *Archives of Physical Medicine and Rehabilitation* **84**, 1680-1686.

- Snyder-Mackler,L., De Luca,P.F., Williams,P.R., Eastlack,M.E., and Bartolozzi,A.R. (1994) Reflex inhibition of the quadriceps femoris muscle after injury or reconstruction of the anterior cruciate ligament. *The Journal of Bone and Joint Surgery* **76A**, 555-560.
- Snyder-Mackler,L., Fitzgerald,K., Bartolozzi,A.R., and Ciccotti,M.G. (1997) The relationship between passive joint laxity and functional outcome after anterior cruciate ligament injury. *The American Journal of Sports Medicine*. **25**, 191-195.
- Solomonow,M., Baratta,R., Zhou,H., Shoji,E.E., Bose,W., Beck,C., and D'Ambrosia,R. (1987) The synergistic action of the anterior cruciate ligament and thigh muscles in maintaining joint stability. *The American Journal of Sports Medicine*. **15**, 207-213.
- Suter,E., Herzog,W., and Bray,R. (2001) Quadriceps activation during knee extension exercises in patients with ACL pathologies. *Journal of Applied Biomechanics* **17**, 87-102.
- Swanik,C.B., Lephart,S.M., Giannantonio,F.P., and Fu,F.H. (1997) Reestablishing proprioception and neuromuscular control in the ACL-injured athlete. *Journal of Sport Rehabilitation* **6**, 182-206.
- Tashman, S. and Anderst,W. (2003) In-Vivo measurement of dynamic joint motion using high speed biplane radiography and CT: *Application to canine ACL deficiency*. *Journal of Biomechanical Engineering*, **125**, 238-245
- Thacker,S.B., Stroup,D.F., Branche,C.M., Gilchrist,J., Goodman,R.A., and Kelling,E.P. (2003) Prevention of knee injuries in sports: A systematic review of literature. *Journal of Sports Medicine and Physical Fitness*. **43**, 165-179.
- Tortora,G.J. and Grabowski,S.R. (1996) *Principles of anatomy*. Harper Collins, Menlo Park, CA.
- Tsuda,E., Okamura,Y., Otsuka,H., Komatsu,T., and Tokuya,S. (2001) Direct evidence of the anterior cruciate ligament-hamstring reflex arc in humans. *The American Journal of Sports Medicine* **29**, 83-87.
- van Eijden,T.M.G.J., Kouwenhoven,E., Verburg,J., and Weijs,W.A. (1986) A mathematical model of the patellofemoral joint. *Journal of Biomechanics* **19**, 219-229.
- Vandertuin,J.F. and Grant,J.A. (2004) The role of functional knee braces in managing ACL injuries. *Athletic Therapy Today*. **9**, 58-62.
- Wei,S.-H. (2000) Dynamic joint and muscle forces during knee isokinetic exercise. *Proceedings of the National Science Council* **24**, 161-168.
- White,A.A. and Panjabi,M.M. (1990) *Clinical biomechanics of the spine*. JB Lippincott, Philadelphia, PA.

- White,S.C., Yack,H.J., and Winter,D.A. (1989) A three dimensional musculoskeletal model for gait analysis. Anatomical variability estimates. *Journal of Biomechanics* **22**, 885-893.
- Williams,G.N., Snyder-Mackler,L., Barrance,P.J., and Buchanan,T.S. (2005) Quadriceps femoris muscle morphology and function after ACL injury: a differential response in copers versus non-copers. *Journal of Biomechanics* **38**, 685-693.
- Winter,D.A. (2005) *Biomechanics and motor control of human movement*. John Wiley & Sons, Inc., Hoboken, NJ.
- Wojtys,E.M. and Huston,L.J. (1994) Neuromuscular performance in normal and anterior cruciate ligament-deficient lower extremities. *The American Journal of Sports Medicine*. **22**, 89-104.
- Wojtys,E.M., Huston,L.J., Taylor,P.D., and Bastian,S.D. (1996a) Neuromuscular adaptations in isokinetic, isotonic, and agility training programs. *The American Journal of Sports Medicine* **24**, 187-192.
- Wojtys,E.M., Kothari,S.U., and Huston,L.J. (1996b) Anterior cruciate ligament functional brace use in sports. *The American Journal of Sports Medicine* **24**, 539-546.
- Yamaguchi,G.T. (2001) *Dynamic modeling of musculoskeletal motion: A vectorized approach for biomechanical analysis in three dimensions*. Kluwer Academic Publishers, Norwell, MA.
- Yanagawa,T., Shelburne,K., Serpas,F., and Pandy,M. (2002) Effect of hamstrings muscle action on stability of the ACL-deficient knee in isokinetic extension exercise. *Clinical Biomechanics* **17**, 705-712.
- Zajac,F.E. (1989) Muscle and tendon: Properties, models, scaling, and application to biomechanics and motor control. *Critical Reviews in Biomedical Engineering* **17**, 359-411.
- Zimny,M.L., Schutte,M., and Dabezies,E. (1986) Mechanoreceptors in the human anterior cruciate. *The Anatomical Record*. **214**, 204-209.
- Zimny,M.L. and Wink,C.S. (1991) Neuroreceptors in the tissues of the knee joint. *Journal of Electromyography and Kinesiology*. **1**, 148-157.

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