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THE PARALLEL SQUAT EXERCISE: A BIOMECHANICAL COMPARISON BETWEEN NORMAL AND ANTERIOR CRUCIATE LIGAMENT DEFICIENT KNEES

presented by

Terry Alan Bemis

has been accepted towards fulfillment
of the requirements for

Master ~~of Science~~ degree in Biomechanics

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THE PARALLEL SQUAT EXERCISE: A BIOMECHANICAL COMPARISION BETWEEN NORMAL AND ANTERIOR CRUCIATE LIGAMENT DEFICIENT KNEES

By

Terry Alan Bemis

A THESIS

Submitted to
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in partial fulfillment of the requirements
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ABSTRACT

THE PARALLEL SQUAT EXERCISE: A BIOMECHANICAL COMPARISON BETWEEN NORMAL AND ANTEROR CRUCIATE LIGAMENT DEFICIENT KNEES

By

Terry Alan Bemis

The purpose of this study was to compare biomechanical parameters between normal and anterior cruciate ligament (ACL) deficient knees during a parallel squat. Besides the standard kinetic and kinematic parameters, an attempt was made to document tibial translations using motion analysis. To accomplish this, thermoplastic thigh target attachment molds were utilized to decrease skin motion, and a mathematical model was developed to describe the translations. Six subjects, with unilateral ACL deficiencies, participated. Ground reaction forces and torques, motion, and electromyographic data were collected simultaneously on each knee during seperate squats. The data were analyzed descriptively and statistically. No statistically significant differences ($p<.05$) were found for any of the parameters between groups. Translational patterns were captured and graphical differences in anterioposterior translations were present between groups. These findings suggest that squats should be used with caution following a recent ACL surgery, and that dynamic laxity can be documented using motion analysis.

DEDICATION

To my wife Emily, daughter Carrie, and sons Caleb and Joshua: Thank-you for your love, understanding, and support. Please forgive me for the missed times together.

In loving memory of my daughter Christian Joy. I miss you.

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To the United States Army and the Army Medical Specialists Corps for funding my education.

TABLE OF CONTENTS

LIST OF TABLES.....vii

LIST OF FIGURES.....viii

CHAPTERS

I. INTRODUCTION.....1

II. LITERATURE REVIEW.....4

Squatting.....4
Kinematics of the Knee.....12
Tibial Translation Studies.....22
Hamstring and Quadricep Muscle Coactivation.....26
Joint Laxity and Compressive Loading.....32
Marker Attachment Methods.....34
Literature Review Summary.....35

III. EXPERIMENTAL METHODS.....36

Equipment.....36
Pilot Study.....39
Subjects.....41
Experimental Preparation.....42
Experimental Protocol.....43

IV. ANALYTICAL METHODS.....48

V. RESULTS.....58

Kinematics.....58
Ground Reaction Forces and Moments.....69
Electromyography.....73

VI. DISCUSSION.....75

VII. CLINICAL IMPLICATIONS AND CONCLUSIONS.....87

VIII. REFERENCES.....92

LIST OF TABLES

Tables

1. Subject Characteristics.....41

2. Knee Diagnoses and Surgeries.....42

3. Mean Rotational Motions (Degrees).....60

4. Mean Translations of Tibia Relative to the Femur (cm).....69

5. Mean Flexion/Extension Moments (%BW x HGT).....71

LIST OF FIGURES

Figures

1. Femoral Based Reference Coordinate System.....19
2. Tibial Based Reference Coordinate System.....20
3. Joint Coordinate System.....21
4. Calibration Space/Lab and Force Plate Coordinate Systems.....37
5. Angles Formed by Target Triads.....40
6. Target Placement with Thigh Mold.....44
7. Ground Reaction Forces (Fz) from Three Successive Trials.....46
8. Position Vectors in Laboratory Coordinates.....49
9. Description of Translational Method.....53
10. Typical Motion Coupling Pattern over Time.....59
11. Typical Crossplot of Flexion vs Rotation.....61
12. Typical Crossplot of Flexion vs Ab/Adduction.....61
13. AP Translations over Time during Squat Repetitions.....62
14. Crossplots of Flexion vs AP Translation (Normals).....63
15. Crossplots of Flexion vs AP Translation (ACL Deficient).....64
16. Crossplots of Flexion vs ML Translation (Normals).....65
17. Crossplots of Flexion vs ML Translation (ACL Deficient).....66
18. Crossplots of Flexion vs SI Translation (Normals).....67
19. Crossplots of Flexion vs SI Translations (ACL Deficient).....68
20. Typical Ground Reaction Force Pattern.....70
21. Typical Knee Moment Patterns.....72

22. Typical EMG Patterns.....74

23. Vertical GRF Produces Posterior Tibial Shear as Tibial Plateau Approaches Vertical Position.....80

I. INTRODUCTION

Kinematics is a geometric description of motion while kinetics describe the forces which are responsible for the motion. To describe motion, systems of links are often utilized. A link is a straight line of fixed length running from one axis of motion to another. According to Gowitzke,[36] Reuleaux in 1875 introduced the term "kinematic chain" to refer to a mechanical system of links. An engineering definition of a kinematic chain, is a closed system of links joined in such a manner that if any one link is moved, all the other links will move in a predictable pattern.[36]

Steindler[96] applied the principles of kinematic chains to the body, using the term "kinetic chain" rather than "kinematic chain". He defined two seperate kinetic chain systems: open and closed. Open kinetic chains occur when the terminal link is freely movable i.e. when waving a hand or performing a knee extension exercise. Closed kinetic chains occur when the terminal link is fixed by an external resistance, and if unmovable, the proximal links move against the fixed distal link. The chin-up and squat exercises are examples of closed kinetic chain exericises. Most authors do not differentiate between the descriptions of motions or forces and continue to use the term "kinetic chain." [37,58,80,90] Gray[37] has even coined the term "chain reaction" to describe the kinematic responses of the proximal

link segments and joints in response to the foot being on the ground. Walking is an example of an activity which includes both open and closed kinetic chains. The system of links is closed during the stance phase of gait and open during the swing phase of gait.

Traditionally, most orthopedic rehabilitation exercises have been of the open chain nature. These range from the common straight leg raising and knee extension/flexion progressive resistive exercises to the variety of expensive, yet popular, isokinetic machines. Interestingly, many of the traditional rehabilitation techniques for neurological patients have incorporated functional closed chain exercises, as the sit-to-stand exercise. Recently more interest has been shown in incorporating more closed chain exercises and early on in orthopedic rehabilitation protocols.

For years, controversy has existed regarding the proper rehabilitation of the anterior cruciate ligament (ACL) deficient or post-operative ACL repair/reconstruction knee. Palmatier,[80] has described the theoretical basis for utilizing closed chain exercises over open chain exercises for ACL rehabilitative protocols. The squat exercise has frequently been used as an example of a safe closed chain exercise for ACL related knee rehabilitation because of the combined vertical loading of the knee joint and the hamstring muscle activity which theoretically reduce the strain on the ACL. Because of the scepticism that exists in the rehabilitation community, more research is needed to confirm presented theories of closed kinetic chain exercises. The purpose of this thesis is twofold. First, to explore differences in the kinetics, kinematics, and electromyographic (EMG) activity in normal and ACL deficient

knees while performing a parallel squat exercise, and second, to describe a methodology of recording translational motions at the knee using motion analysis and surface mounted targets. For the purpose of this thesis, the term "knee" will refer only to the tibiofemoral joint.

II. LITERATURE REVIEW

The survey of literature will be divided into seven sections: Squatting, kinematics of the knee, tibial translation studies, quadricep and hamstring muscle coactivation, joint laxity and compressive loading, marker attachment methods, and literature review summary.

Squatting

The act of squatting plays a role in almost all lives, and is performed several times a day as an activity of daily living. Some oriental cultures are known to spend long periods at a time in a deep squat position. In fact, attrition lesions of the articular cartilage have been attributed to the habitual squatting posture in the Japanese.[69] The history of squatting as an exercise is not well documented. However, from paintings, reliefs, mosaics and writings, we know that many ancient civilizations placed a high importance on physical activity, which included acts of strength. Many Greek stories and legends have been passed on to modern times. Among them is the story of one of the most famous Greek athletes named Milo, whose most renowned feet was lifting a four-year-old heifer onto his shoulders and carrying it the full length of the

stadium at Olympia.[62] A squat would have been necessary at the beginning and end of this exercise.

One of the first recorded feats of strength in England was that of Thomas Topham (1741) who lifted three hogshead of water weighing approximately 1800 pounds by means of a strap over his shoulders.[62] During the next century, an american physician named George Barber Winship, popularized a similar form of exercise.[62] His exercise, known as the "health lift," involved the lifter standing astride a weight with knees bent and a harness attached to the weight and placed over the shoulders. The weight was lifted by extending the knees. According to Winship and his followers, the lift had considerable therapeutic and fitness value.

The full squat exercise with a barbell was popularized in America by a young German immigrant by the name of Heinrich "Henry" Steinborn, also known by his later stage name of Milo.[98] He had learned to weightlift while in an Australian prison camp during World War I. Articles which described his squatting feats, comparing his legs to the "Pillars of Hercules" created a wave of interest in the squat exercise. As more people tried the exercise, many startling and phenomenal claims appeared in the media. By the late 1950s, the squat had become the key to any serious program of weight training.

Although, according to Todd,[98] the full squat exercise was criticized as early as 1946 by Ravell, the exercise remained popular until the early 1960's when Klein published several papers on studies indicating that the deep squat was responsible for the production of ligament instability at the knee. Klein[52,53] and

others[9,27,87] recommended that no more than the half squat (thighs approximately parallel to the floor) be performed during weight training and that deep knee bending exercises such as the duck walk be avoided. As a result, many, including the military branches, banned the deep knee bending exercises. Reports of patellar tendon ruptures during deep squats with heavy loads added to the fears.[72,105]

More recently, studies have questioned the effect of full squats on knee stability. Todd[98] criticized Klein's studies as poorly controlled and probably biased. Meyers[67] compared full squats at various speeds and with various loads against similar half squats. He found no statistical differences between groups or with pre- and post-measurements following an 8 week exercise program.

Steindler[96] studied the immediate effects of exercise on anteroposterior laxity at the knee in squat power lifters, basketball players, and distance runners with sedentary controls and subjects under general anesthesia. He noted a significant increase in laxity in the basketball players and distance runners, but no change in the other groups. However, upon closely examining the data, it was evident that the squat power lifters already had greater anterior laxity than the others indicating a chronic stretching of the ACL and posterior capsule. However, the athletes did not report instability. Steiner also admitted to a source of error: The weight lifters were not tested by the same person who performed the passive laxity tests on all the other groups.

More recently, Chandler[17] performed knee laxity tests using a KT-1000 knee arthrometer on subjects pre- and post 8 weeks of

training with the half squat versus the full squat versus controls. He also performed the same laxity tests on male power and weight lifters. He found that knee stability did not decrease by 8 weeks of training using either the half or full squat, and that long term performance of squat exercises had no effect on knee stability. He concluded that the full squat was a safe exercise in terms of not causing permanent stretching of the knee ligaments.

Kinetic, and EMG studies have also been performed on the squat exercise. Ariel[7] investigated the forces and moments of force acting about the knee in 12 experienced weight lifters during deep knee bends with heavy loads. He found that the mean maximum flexing moment occurred at the deepest position of the squat and that the shearing force was of greatest magnitude at the beginning of the descent, but commented that the knee would be more vulnerable when flexed greater than 90 degrees. He also found that forward movement of the knees was associated with the greatest shearing forces. The direction of these shearing forces was not stated.

Sagawa[89] studied squats at two rates: 60 times/min and 30 times/min. He found the maximum flexion moment (indicating extensor muscle activity) and vertical ground reaction force occurred at the deepest position of the squat. EMG activity was reported in the vastus lateralis, biceps femoris, gastrocnemius, and tibialis anterior muscles before and after reaching the deep squat position. He was perplexed by the activity of the hamstring and tibialis anterior muscles during an extensor dominant exercise, but suggested that their function was to control body velocity. Sagawa

also reported that the faster rate squats significantly increased the moments produced.

McLaughlin[63,64] studied 12 world class lifters during competition, thus no markers were applied to the subjects. He found that the estimated moments about the knee reasonably compared with those reported by Ariel.[7] He also reported that the more the subject leaned forward, the less extensor dominent the knee moment, and that the higher skilled lifters produced higher extensor moments at the knee. Thus, he recommended the trunk remain erect if the exercise is primarily for quadricep development.

Carlsoo and Molbech[16] reported increased activity in both the quadricep and hamstring muscle groups during a squat repitition. Dahlkvist[22] reported similar muscle activity during squatting, but in contrast to Ariel,[7] found the maximum shear forces in the deepest squat position. He also implicated the ACL as being in danger in this flexed position. Nisell[72] also found the maximum shear forces at the deepest part of the squat, but he was the only author to report these forces as being posteriorly directed on the tibia. He further reported the maximum knee flexing moments at the beginning of the ascent from the deepest squat postion.

Andrews and Hay[2] reported a resultant vertical knee force that was essentially constant at a slow rate (2 second descent and 2 second ascent) but widely varying during at a fast rate (1 second descent and 1 second ascent). They also reported high shear forces which maximized at the deepest squat position.

In another study, Andrews and Hay[1] compared the resultant knee shear forces produced by use of a barbell versus a weightmachine at

three different loads and rates. For all exercise conditions, maximum values of the vertical force and shear forces occurred when the greatest load was lifted at the fastest rate. The typical squat machine shear force levels were 30-40% higher than a comparable barbell squat.

Henning[41] arthroscopically strain gauge instrumented the ACL in two subjects with grade II ACL strains and measured the strain across the ACL while the subjects performed a variety of activities. One of the subjects performed a "one-leg half squat." He reported that the recorded strain was less that that recorded during an 80 pound Lachman test. Without presenting the details of his findings, he recommended that squats be perfomed within 20-90 degrees flexion.

Costigan and Reid[20] examined the effects of foot position on the radial torque produced during deep squats. In general, they found that as the knee flexed, increased lateral torques (moments) were produced. But in the extreme external rotation foot position a medial torque was produced. Interestingly, they found that the natural foot position produced torques lower than the predetermined foot positions of 20 degrees internal rotation, straight ahead, and 50 degrees external rotation.

In a 3-dimensional analysis of a squat exercise, Hattin[40] studied 10 normal males while performing 50 half squat repititions at 1 and 2 second ascent/descent interval rates. He found that the maximum anterioposterior shear and compressive forces occurred at the deepest squat position, medial/lateral shear was small, the faster cadence increased anterioposterior shear (50%) and compressive

forces (28%), and that anterioposterior shear was most affected by fatique. In contrast to Andrews and Hay,[2] Hattin reported that both the slow and fast speed cadences produced similar patterns over the squat cycle.

In a review of squat related literature, Garhammer[30] concluded that no objective scientific evidence exists which would implicate knee damage caused from the squat exercise to the "parallel" position. He did recommend that squats be performed with a slow rate of descent, avoiding any bouncing in the deep squat position.

Today, the use of the squat exercise in training programs is again gaining in popularity, but uncertainties abound. In 1989, 160 survey responses from high school football coaches were obtained regarding the use of squats in their training programs.[18] The survey found that 78% of the coaches used some form of the squat exercise, ranging from a quarter squat to a full squat. Parallel squats were used most often (56%) and the general attitude was negative for squats below parallel.

In a published roundtable discussion of well known strength coaches,[26] the attitude was overwhelmingly positive toward the squat exercise. One of the participants,Tom McLaughlin, Ph.D., stated that he considered the squat to be the best resistance exercise an athlete could do. Minor differences in techniques were also discussed. These included foot position, use of belts, and the depth of the squat. Klein advocated that the depth of the squat should be limited to 1/4 or no lower than thighs parallel to the floor for protection of the knee ligaments.

In the past, the use of squats in ACL rehabilitation protocols were never mentioned.[14,23,35,45,81] More recently forms of the squat exercise have been included. Antich[6] recommended the use of partial squats (less than 90 degrees) for chronic ACL deficient knees. To avoid anterior tibial translation, he recommended the use of volitional quadricep and hamstring muscle concontractions during the squat. He did not mention the use of squats for acute ACL deficiencies.

Shelbourne[90] compared the patients' functional outcomes of ACL surgeries to compliance with the rehabilitation protocol. He found that patients who were non-compliant (they progressed as they desired and at a faster rate) returned to "normal" function sooner (and without instability) than the compliant patients. As a result, he revised his exercise protocol and included many closed chain exercises early on. His original protocol included squats beginning at 6 months, while the revised protocol included quarter squats at 2-3 weeks, with progression as tolerated.

Others have also recommended using squats early, but the type of squat varies. Lutz[58] begins quarter squats within the range of the rehabilitation brace once the patient is able to bear full weight. He adds power squats toward the end of the strengthing stage. Silfverskiold[92] stated that quadricep and hamstring muscle rehabilitation exercises should be weightbearing, but within a safe range of 90-30 degrees of flexion. Thus, he recommends a 1/3 squat between 90-30 degrees as an ideal way to begin rehabilitation of the quadriceps and hamstring muscles. At 8-12 weeks, he adds single 1/3 squats. Montgomery[68] uses 3 sets of up to 100

repititions of 1/4-1/3 single leg squats for endurance 8 weeks post injury in the ACL deficient knee. In operated legs, he uses 1/4-1/2 squats at approximately 3 months post surgery. And, Podesta[84] incorporates squatting exercises in a range of 45-90 degrees flexion at approximately 5-6 weeks post surgery. The arc of motion is then increased from 0-90 degrees during the 7-12 week period.

Using force diagrams, Palmatier[80] has theorized that ACL strain is reduced during the squat exercise secondary to the axial orientation of the applied load and muscular cocontraction. Because the squat exercise induces hamstring contraction to stabilize the hip, its contraction is claimed to help neutralize the tendency of the quadriceps to cause anterior tibial translation. As a result, he stated that the squat exercise would be a safe exercise to use early in the ACL rehabilitation process.

From the literature, it is apparent that differences still exist in the use of squats during rehabilitation of the ACL problem knee. Other than Henning, no studies have been found which examine the effects of squatting on the ACL deficient knee.

Kinematics of the Knee

In the saggital plane, the knee's primary motion is flexion and extension, but it is not considered a true hinge joint. Various terms have been used, inconsistently, to describe the arthrokinematics or joint surface motion at the knee. The terms and definitions include:

1) Rolling[49,51,71,74,99] - the contact points on each of the joint surfaces change by an equal amount. This is analogous to a wheel rolling along with perfect traction.

2) Gliding[74] and Sliding[51] - the contact point of the moving segment remains the same while that of the stationary segment changes. This is analogous to a wheel of an automobile with locked brakes skidding on ice.

3) Spinning,[74] Rotation,[74] and Sliding[49,99] - the contact point on the stationary segment remains constant while the contact point of the moving segment changes. This is analogous to a stuck automobile wheel spinning in the snow.

For the purpose of this thesis, the terms rolling, sliding, and spinning will be utilized in the order as defined.

This mechanism of combined joint surface motions allows for economy of articular cartilage with respect to the size of the joint surface necessary for accomplishment of the movement and prevents degeneration of the joint surfaces.[51] The literature agrees that, in the sagittal plane, rolling occurs in combination with one of the other joint surface motions at the knee, but disagrees as to whether it is sliding or spinning. Nordin and Frankel,[74] and Kessler[51] report that the tibia rolls and slides on the femur as it extends and flexes. In a computerized, sagittal plane, four-bar linkage model of the knee, O'Connor[77] defined the joint center as the intersection of the cruciate ligaments. He showed that this flexion axis moved posteriorly as the knee flexed and anteriorly as it extended secondary to the changing slopes of the cruciate ligaments. By taking the articular contact point as being perpendicular to the flat

tibial plateau and passing through the joint center, he demonstrated that the distance between successive contact points on the femur were 3-4 times greater than the corresponding distances on the tibia as the femur was flexed 0-140 degrees on the tibia. He concluded that the femur must slide forward on the tibia while rolling backward during flexion and must slide backward while rolling forward during extension. Kapandji[49] described knee motion as a combination of rolling and spinning. He cited an experiment performed by the German Weber brothers in 1836 which involved marking contact points between the femur and tibia during various amounts of flexion. They noted that the point of contact on the tibia moved posteriorly during flexion and that the distance between the points of contact marked on the femoral condyles was twice as long as that between the corresponding points on the tibial condyle. Citing the same study by Weber, O'Connor[77] interpreted the combined motions as rolling and sliding.

The question is whether the differences in the distance between the contact points were due to the femur spinning or sliding. A simple analysis of sliding and spinning based upon the previously presented definitions may help clarify the issue. If the knee is extending, by the tibia moving on a fixed femur, then essentially a single contact point of the tibia is contacting several contact points on the femur. Thus, by definition, the tibia is sliding anteriorly on the femur. Because of the flat or concave shape (primarily due to the menisci) of the tibial plateau, it would seem impossible for it to spin on the convex femoral condyle. On the other hand, if the knee is extending by the femur moving on a fixed tibia, one could apply

either definition in combination with rolling and obtain the same extended position. That is, the femur could either slide posteriorly as it rolls anteriorly, or it could spin anteriorly faster than it is rolling anteriorly. Perhaps the knee functions differently in an open kinematic chain than in a closed kinematic chain situation.

A related phenomenon is the Concave-Convex Rule.[51] It states that if a concave surface (tibial plateaus) moves on a convex surface (femoral condyles), roll and slide must occur in the same direction: if a convex surface (femoral condyles) moves on a concave surface (tibial plateaus), roll and slide occur in opposite directions. Roll always occurs in the same direction of the swing of the bone. For the purpose of this thesis, it is assumed that flexion and extension of the knee is accomplished by the combination of rolling and sliding as described by the Concave-Convex Rule.

The exact amounts of rolling and sliding that occur during the full range of flexion and extension have not been determined. Kapandji[49] cited a study by Strasser (German) who in 1917 reported the ratio of rolling to sliding varied during flexion and extension. He found that starting from full extension the femoral condyle begins to roll without sliding and then the sliding motion becomes more dominant so that at the end of flexion, the condyle slides without rolling. Kapandji reported that the medial femoral condyle incorporates pure rolling during the first 10-15 degrees of flexion, while the lateral femoral condyle performs pure rolling until 20 degrees of flexion. These differences relate to the geometric size differences of the condyles.

Nordin and Frankel[74] have described a 2D method of locating the instant center pathway and joint surface motion using successive roentgenograms. They state that pure rolling could occur only when the instant center is located on the joint surface which does not happen in the normal knee. Their study indicated that a component of sliding is present at each interval of motion from full extension to flexion. Mueller[71] also seems to agree. He stated that the ratio of rolling to sliding is approximately 1:2 in early flexion and 1:4 by the end of flexion.

Whatever the ratio of rolling to sliding may be, the cruciate ligaments are believed to play a dominate role in providing the proper combination. By their alignments, the cruciates limit the amount of rolling the femoral condyles can accomplish and force the sliding component. Mueller[71] stated that the cruciate ligaments perform the function of a true gear mechanism and form the nucleus of the knee joint kinematics. The importance of the ACL in controlling motion is evident by portions of the ligament being taut during all ranges of flexion/extension.[28,65,102]

In the presence of a torn ACL, a deterioration of the rolling and sliding components is believed to occur.[71] Using an instant-centre of movement technique, Gerber[33] studied normals and subjects with different knee pathologies, including acute ACL disruptions. Six to eight radiographs were taken at 10-15 degree intervals from full extension to flexion while the subjects layed on their sides. He found that the centroids of the deficient knees shift suddenly forward and downward between 20-40 degrees flexion and move

back into a normal or slightly posterior position as 90 degrees was approached.

In the study of kinematics between two bodies, either of the bodies can be chosen as a reference or fixed body about which the other moves. At the knee, motion of the tibia may be described relative to the femur, or motion of the femur may be described relative to the tibia. Thus, external rotation of the tibia on the femur is equivalent, but opposite in direction to internal rotation of the femur on the tibia. Movements between two bodies is a displacement which involves both a rotation and a translation.[99] The knee has six degrees of freedom of motion:[13,38,73,82] three rotational and three translational. In general terms, the rotational motions are defined as flexion/extension about a mediolateral axis; abduction/adduction about an anterioposterior axis; and external/internal rotation about a superioinferior axis. The translational motions are defined as medial/lateral along the mediolateral axis; anterior/posterior along the anterioposterior axis; and superior/inferior along the superioinferior axis.

Although the dominant rotational motion of the knee is flexion/extension in the sagittal plane, it is known that motion occurs simultaneously in all planes. These combined planar motions are described as coupled motions.[13,91] The screw home mechanism is a commonly used example of the coupling motion at the knee where extension is coupled with external tibial rotation and anterior tibial translation.[39,61,76] Coupled motion at the knee is primarily dependent upon the geometry of the articular surfaces and the restraints of the ligaments.[13,49,77,99]

While the rotational motions are the most obvious and most easily documented i.e. by a simple goniometer, the translational motions are small and more difficult to document. Of the six translations described at the knee, anterior translation has received the most attention, because the primary function of the ACL is to prevent this motion.[39,65,71,91,102] Most clinicians define anterior tibial translation simply as displacement of the tibia anteriorly relative to the femur. From a biomechanists point of view, this definition is vague and confusing. Consider a femoral based reference coordinate system (Figure 1) where X defines the anterioposterior, Y the mediolateral, and Z the superioinferior axes. When the knee is fully extended, anterior tibial translation relative to the femur would occur along the X axis in a positive (anterior) direction. Most clinicians would agree that this is anterior tibial translation. However, when the knee is flexed, the X axis continues to define the anterior direction. It is obvious from the diagram that this definition is not what a clinician would describe as anterior tibial translation. It appears that most clinical literature actually describe anterior tibial translation relative to the tibia and not to the femur (Figure 2). With a tibial based coordinate system, one might be confused in describing motion of the femur. Care must be taken in defining the locations and orientations of the axes of motion as different kinematic[73,82] and kinetic[2] results can be produced.

Grood and Suntay[38] presented a joint coordinate system technique for the knee. They described three non-orthogonal unit axes, *e*1, *e*2, and *e*3 (Figure 3). In the femur, *e*2 is aligned with its Y axis, and

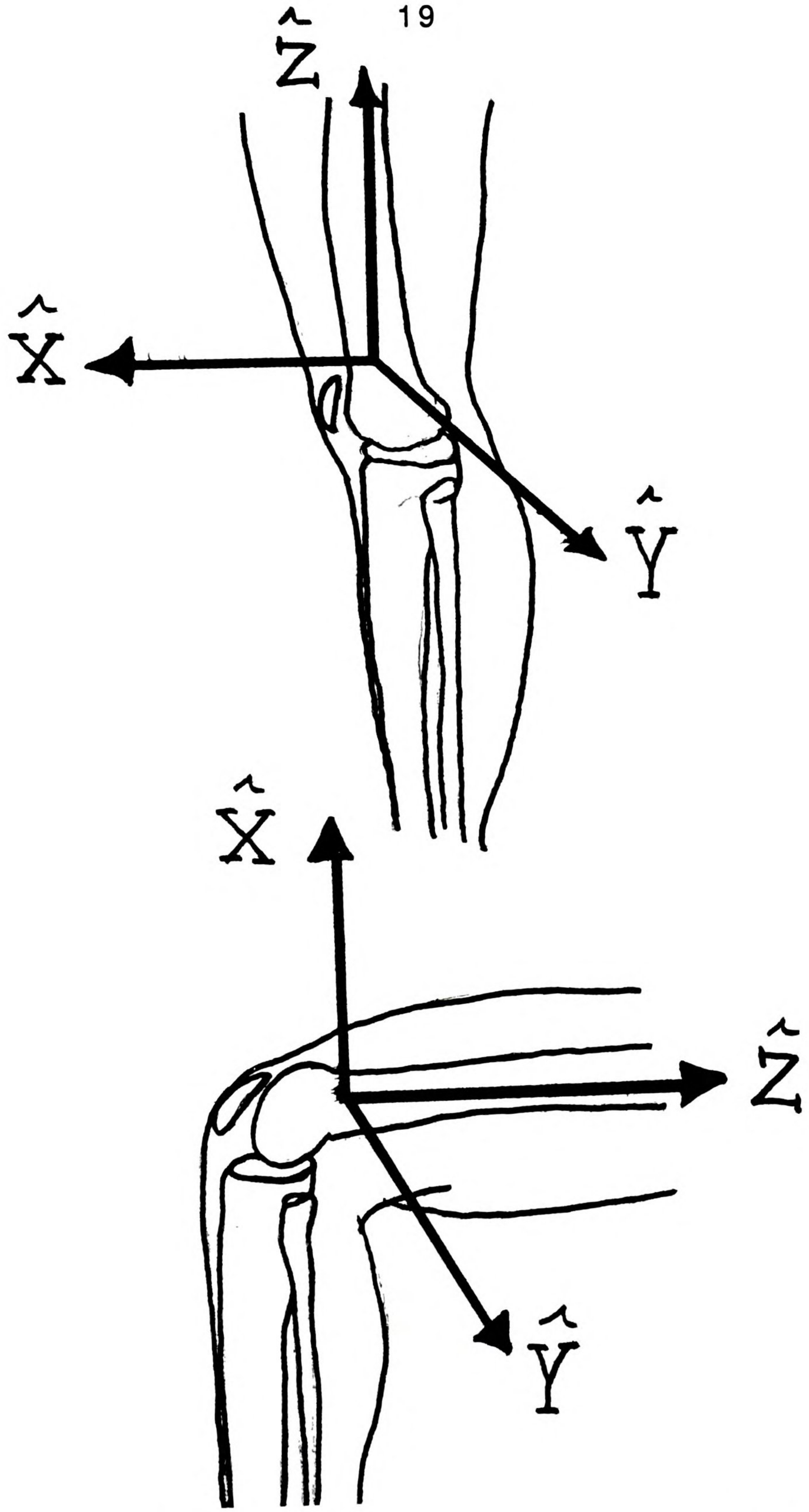


Figure 1. Femoral Based Reference Coordinate System

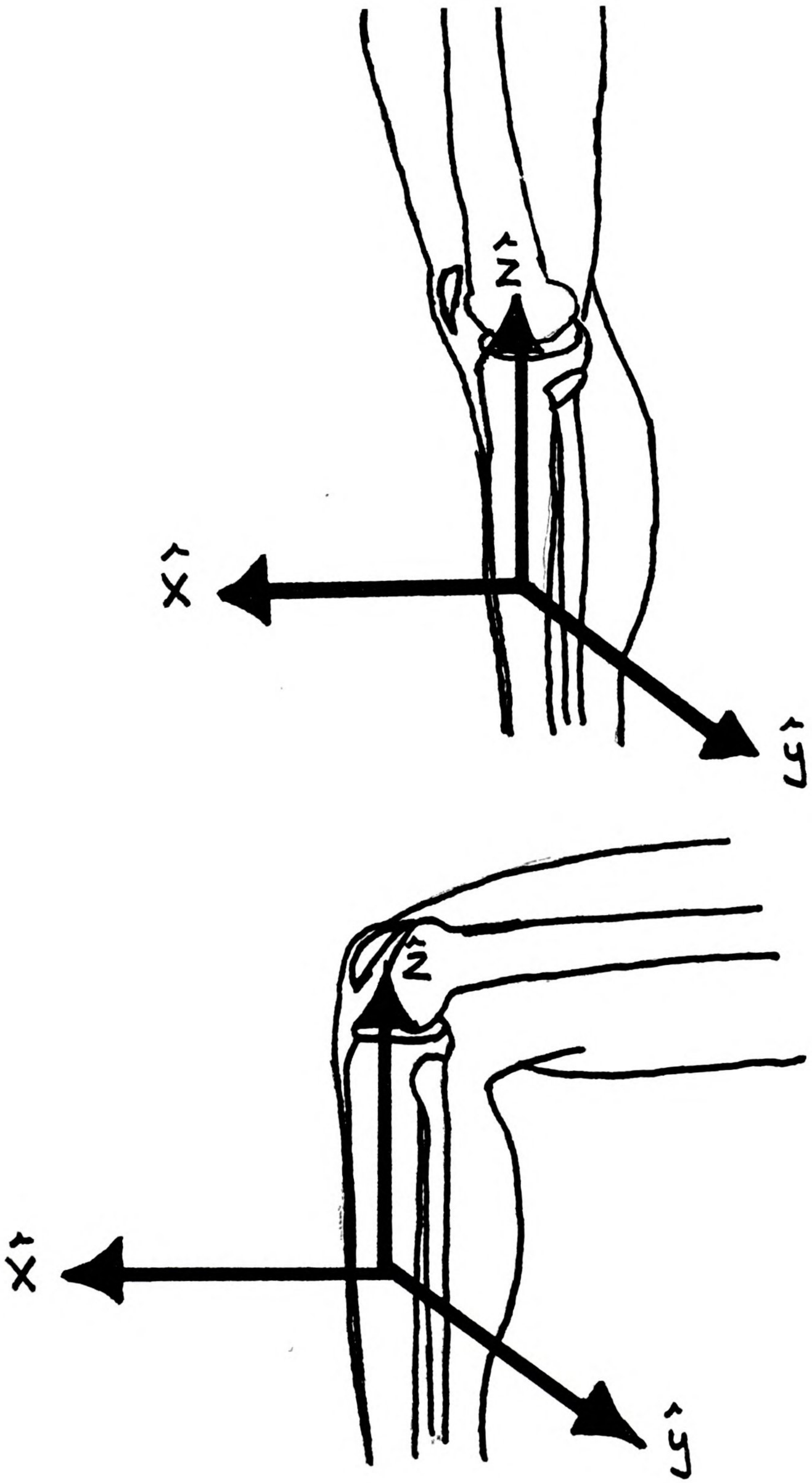


Figure 2. Tibial Based Reference Coordinate System

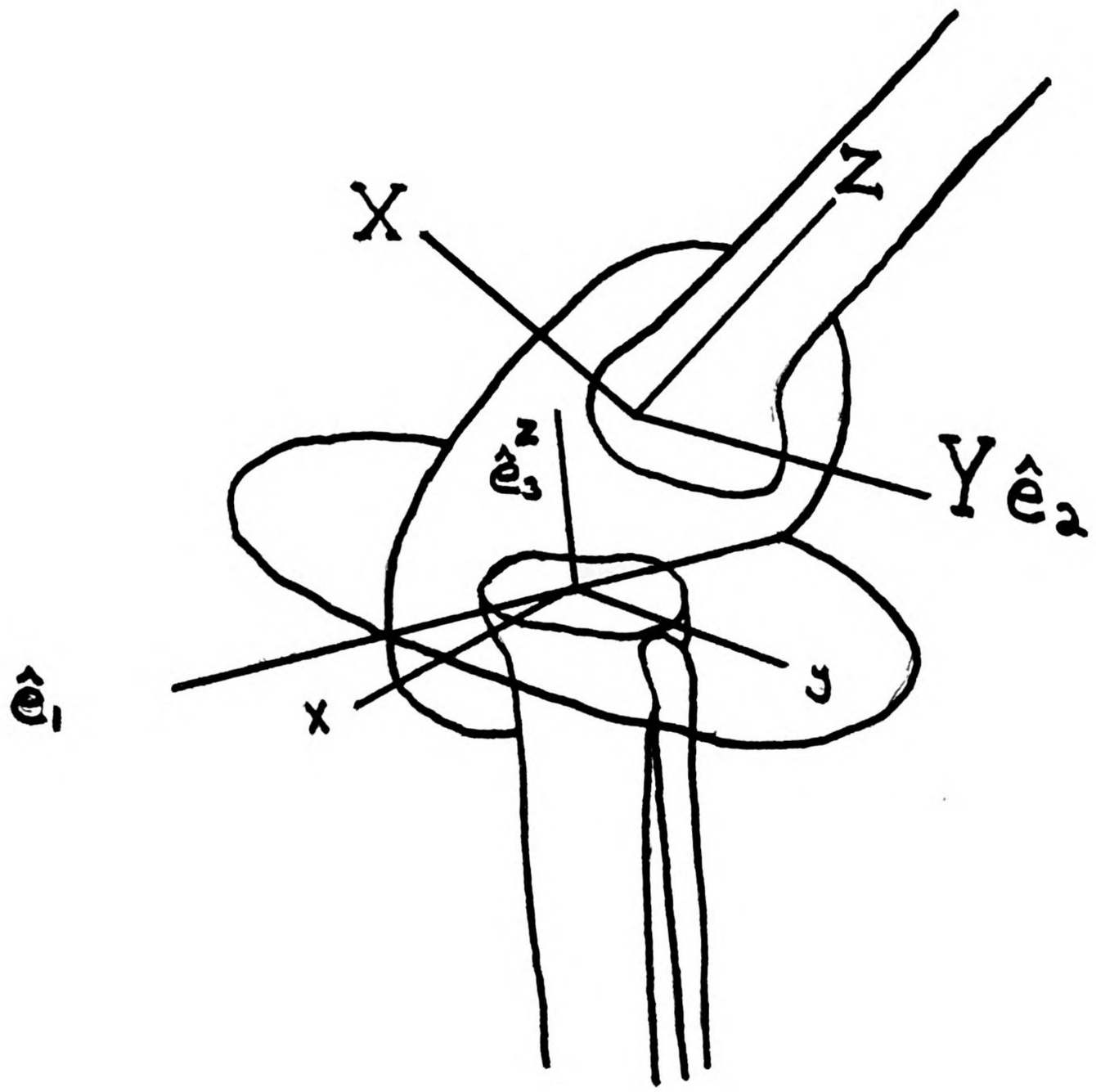


Figure 3. Joint Coordinate System

was selected as the flexion/extension axis. The axis of external/internal rotation was defined as *e*3 which is aligned with the tibial z axis. Both *e*2 and *e*3 are called "body fixed axes" since they are embedded in the two body segments. The final unit axis, *e*1, is mutually perpendicular to both the body fixed axes and is called the "floating axis" because it is not fixed to either body segment. The floating axis, *e*1, was selected as the axis of abduction/adduction, and is also the axis of anterior/posterior translation.

Tibial Translation Studies

As mentioned, the ACL is the primary passive restraint to anterior tibial translation, so a major concern of those involved with the rehabilitaion of the knee is to prevent damage to an acutely repaired or reconstructed ligament or to the secondary restraining tissues. While not describing where the axis of translation lies, investigators have studied "anterior tibial translation" using a variety of methods.

In 1984, Grood[39] documented increased anterior tibial displacement in ACL deficient cadaveric knees while performing simulated knee extensions from 30-0 degrees flexion. This anterior tibial translation was further increased with the addition of a 31 N (7 pound) weight attached to the foot. He concluded that terminal extension exercises at less than 30 degrees should be avoided after ACL repair or reconstruction to avoid stressing and damaging the ligament.

In a similar study, but with the addition of a strain transducer implanted into the ACL of fresh cadaveric knees, Arms[8] reported that simulated quadricep muscle contractions in the range of 45-0 degrees flexion increased ACL strain. He also found that simulated isometric or isotonic quadricep muscle activity did not strain the ACL when the knee was flexed to 60 degrees or more.

In agreement with Arms, Renstrom[88] also reported that simulated quadricep muscle activity at 45-0 degrees flexion significantly increased strain in the ACL. In addition, he found that simulated isolated hamstring muscle activity at all angles tested actually decreased the ACL strain.

Because of the possible errors in applying results from experiments on cadavers to live humans, some authors have sought to measure anterior tibial translation using live subjects. Using lateral radiographs of sidelying subjects during measured isometric quadricep contractions at various angles of flexion, Yasuda[104] reported the mean knee flexion angle at which the anterior tibial translation force became zero was at 45.3 $\pm$ 12.5 degrees. To be on the safe side, he recommended that isolated quadricep muscle extension exercises be performed at knee flexion angles greater than 70 degrees.

In a study of 13 subjects with unilateral ACL deficiency, Jonsson[47] implanted tantalum balls percutaneously into the distal femur and proximal tibia bilaterally. Subjects later underwent stereophotogrammetric examinations while performing a knee extension exercise from approximately 28-0 degrees flexion with a 30 N weight applied to the ankle. In this 3D analysis, he found no

differences in medial/lateral translations, but reported more pronounced distal and anterior tibial displacements in the ACL deficient knees. However, the anterior tibial displacements were only statistically significant at 10-15 degrees. He also reported no differences in the three rotational motions between the ACL intact and ACL deficient knees.

While not addressing quadricep muscle activated anterior tibial translation, Staeubli[94] measured the passive anterior translation obtained by applying 89 N of anterior force via a KT-1000 arthrometer. He also simultaneously recorded the displacement radiographically and compared both results. The measured translations, using the arthrometer, were minimally higher in both the ACL deficient and control knees over the radiographic technique, but both techniques showed significantly higher measurement of anterior displacement in the ACL deficient knees.

Using a KT-1000 arthrometer in combination with a Cybex II isokinetic dynamometer, Howell[43] compared the anterior tibial translations or quadriceps active drawer (QAD) during maximum isometric extensions at multiple knee flexion angles. He found that the anterior tibial translation was only 1-2 mm greater in the ACL deficient knees at 15 and 30 degrees of knee flexion. As in other studies, he also found a posterior tibial translation at 75 degrees flexion. He also reported that a maximal manual anterior tibial force produced an anterior translation which was nearly double that produced by the QAD. He concluded that extension exercises within 15-75 degrees of knee flexion would produce less tension in the ACL than instrumented laxity testing and questioned whether a maximal

quadricep contraction is detrimental to a recently reconstructed ACL.

In apparently the first in-vivo ACL strain gauge study, Henning[41] reported on the results of two subjects with grade II ACL sprains who performed several activities following arthroscopic instrumentation of the ACL. As a reference, an 80 lb Lachman test was considered to equal 100 units of ACL elongation. Thus activities were compared to the Lachman test result. For example, an apparent eccentric quadricep muscle exercise from 0-22 degrees flexion produced 87% and 121% of the Lachman induced ACL elongation in subjects one and two respectively. In agreement with Renstrom[88], he also found that an isometric hamstring muscle contraction produced no ACL strain. However, many activities were performed by only one subject. Based upon the range of differences reported in the quadricep muscle activity, care should be taken in making conclusions from findings on a single subject.

More recently, Howe[42] arthroscopically strain gauge instrumented the anteromedial band of normal ACLs in five subjects and calculated the strain during quadricep contractions at 30 and 90 degrees of flexion. As in previous studies, he found a significant increase in ACL strain during quadricep muscle contractions at 30 degrees and no significant change in strain measured at 90 degrees. He concluded that isometric quadricep muscle activity at 90 degrees of knee flexion could be prescribed without risk of increased strain of the ACL's anteromedial band.

In search for an accurate and non-invasive means of measuring the rotational as well as the translational motions at the knee,

Marans[60] studied 20 ACL deficient and 30 control subjects using an electrogoniometer capable of measuring all 6 degrees of freedom of knee motion. Studying only walking, he reported a significant anterior/posterior translation in the ACL deficient knee during the swing phase of gait. No statistical differences were found in mediolateral or superioinferior translations.

In probably the most accurate description of motions at the knee during walking, Lafortune[54] studied 5 normal males who underwent in-vivo insertion of Steinmann traction pins into their right femurs and tibias. Target clusters were attached to the pins, thus errors from skin motion were eliminated. The kinematics were described according to the joint coordinate system of Grood and Suntay[38] which defines the anterioposterior axis as the floating axis. During gait, they found that when the knee flexed, the tibia underwent a posterior translation and when the knee extended, an anterior translation occurred. The posterior translation was reported as 3.6 ± 1.8 mm during stance and 14.3 ± 3.3 mm during swing. Anterior translation during stance was reported as approximately 1.3 mm from a defined zero point.

Hamstring and Quadricep Muscle Coactivation

One of the main concerns with the rehabilitation of an ACL problem is how to control the anterior tibial translation caused by a quadricep contraction. Whether or not the amount is significant, it is apparent that increased anterior tibial translation occurs during nonweightbearing open chain knee extension at least from 45-0

degrees of flexion in an ACL deficient knee and that some anterior translation is present during gait in a normal knee. It is also apparent that the hamstring muscles produce a posterior tibial translation, thus decreasing the stress on the ACL. Several studies have sought the effect of concurrent hamstring and quadricep muscle contractions on the knee.

Using cadavers, Renstrom[88] simulated quadricep and hamstring muscle cocontractions at various degrees of knee flexion. He concluded that the hamstring muscles are not capable of protecting a recently repaired or reconstructed ACL at angles less than 30 degrees. Draganich[25] performed a similar experiment and reported no statistically significant strain on the ACL from 10-90 degrees. Yasuda[103] reported that cocontractions prevented anterior tibial shear forces at mean knee angles greater than 17.2 degrees in live subjects.

While these studies represent volitional isometric muscle cocontractions, others have studied antagonistic hamstring muscle activity during quadricep muscle activity. Using EMG, Carlsoo and Molbech[16] reported increased rectus femoris activity during isolated hip flexion and knee extension, and increased hamstring activity during isolated hip extension and knee flexion. Also during rapid knee flexion, increased gastrocnemius muscle activity was noted. During a squat, they reported marked activity in both the rectus femoris and biceps femoris and mild activity in the medial hamstrings. A hamstring and quadricep muscle coactivation pattern was also reported during stationary cycling. They proposed that the rectus femoris and biceps femoris muscles can act synergistically

as knee extensors, or what they described as the "paradoxical effect" of these two joint muscles.

Draginich[24] reported hamstring and quadricep muscle coactivation in a study of 6 healthy male subjects during quasistatic (20 degrees/sec) knee extension while in the seated and prone positions. The coactivation reached a maximum at terminal extension (less than 10 degrees flexion). He concluded that the hamstring muscles act in concert with the ACL to prevent anterior tibial translation.

Solomonow[93] compared 12 controls with 12 ACL deficient subjects during a maximum voluntary effort from 90-0 and 0-90 degrees flexion using a Cybex II isokinetic dynamometer. He found that the ACL deficient subjects demonstrated an anterior tibial "subluxation" at approximately 45 degrees while simultaneously producing a marked reduction of EMG activity of the quadricep muscles and increased hamstring muscle activity. Based upon a study with anesthetized cats, he suggested that the mechanoreceptors within the ACL provide reflex sensory information to the thigh muscles to assist in joint stability during deformation of the ligament. (In another study, Pope[85] was unable to reproduce the same results in anesthtized cats.) In the case of the ACL deficient subjects, Solomonow proposed that a secondary reflex arc is present through mechanoreceptors in the muscles, tendons and joint capsule. Although the hamstring muscles were not capable of preventing the anterior tibial translation, these findings suggested a reflexive attempt of the hamstring muscles to correct the instability. Solomonow also reported that subjects with "good" muscle tone were able to prevent subluxation during maximal slow

loading conditions (15 degrees/sec). Pope[86] reported that both the ligament musculoprotection reflex and the tendon-stretch reflex are too slow to protect the medial ligament complex at the knee against many sports injuries.

In a similar study Baratta and Solomonow[11] compared nonathlete controls versus high performance athletes who did not practice hamstring muscle exercises versus high performance athletes who included specific hamstring muscle exercises in their regime. During a sidelying isokinetic knee flexion/extension test at 15 degrees/sec, they found high level agonist and low level antanonist muscle activity. The subjects with muscle imbalance (hypertrophied quadricep muscles compared to hamstring muscles) demonstrated a substantial reduction in this coactivation pattern of the hamstring muscles when acting as antagonists. The same subjects placed on a 2-3 week hamstring muscle exercise program demonstrated increased coactivation activity during later testing. The authors concluded that reduced antagonist activity diminishes the stabilizing force available at the knee, which increases the risk of ligament injury.

Using a variety of evaluative methods, Walla[100] concluded that the presence of "reflex level" hamstring control was most closely associated with a high functional rating score in ACL deficient subjects. This finding was a much more important predictor of a patient's success than the findings of passive laxity testing or Cybex strength and power estimates.

Other researchers have studied muscle activity in normal subjects during such functional activities as walking,[70]

stairclimbing,[4,70] and walking up a ramp.[70] Functional activity studies with ACL deficient subjects have also been published.

Limbird[56] reported decreased quadricep and gastrocnemius muscle activity and increased hamstring muscle activity in ACL deficient subjects during the early stance phase of free walking. He postulated that the stated pattern was an attempt to generate a greater posterior force on the tibia during weight acceptance. During fast walking, however, the EMG activity was more similar between groups.

Kaalund[48] compared 9 ACL deficient subjects with 9 normals during treadmill walking. He found no differences between groups during level walking, but during uphill walking, the hamstring muscles in the ACL deficient group were activated significantly earlier in late swing than that of the controls. Yet in a similar study, Lass and Kaalund[55] reported earlier onsets and longer durations of EMG activity in the lateral hamstring and medial gastrocnemius muscles of the ACL deficient subjects during level treadmill walking. They concluded that both the hamstrings and gastrocnemius muscles contribute to the functional stability of the ACL deficient knee.

Tibone[97] compared kinetic, kinematic, and EMG data between the ACL intact and ACL deficient knees of the same subjects during a variety of functional activities (free walking, fast walking, running, straight cutting, cross cutting, ascending stairs, and descending stairs). No major differences were found except for a decreased ground reaction force in the ACL deficient subjects during running and cutting, and an earlier cessation of quadricep activity in the ACL

deficient group during stair climbing. In contrast, Branch found increased hamstring activity and decreased quadricep activity during the stance phase of cutting.

McNair[66] reported in an interlimb comparison of subjects with unilateral ACL deficiency while jogging on a treadmill. He found increased quadricep muscle activity at footstrike and earlier activity of the hamstring muscles shortly after footstrike in the ACL deficient knee. He concluded that this pattern was to decrease the anterior tibial translation.

Gauffin[32] studied one legged jumping in normal vs ACL deficient knees of the same subject. He reported that the ACL deficient extremity showed higher angles of hip and knee flexion at touchdown suggestive of the use of hamstrings as stabilizers. EMG recordings showed decreased quadricep muscle activity in the deficient knees, but no differences in the hamstring muscle activity.

Berchuck[12] and Andriacchi[5,12] described a study of 16 ACL deficient subjects with 10 controls which focused on the moment tending to flex or extend the knee joint during walking, jogging, and ascending and descending stairs. Without using EMG, they described the use of moments to reflect muscle activity. An applied "external" moment would have to be balanced by an equal, but opposite, "internal" (muscle) moment to maintain equilibrium. They found that the ACL deficient subjects tended to adopt a moment pattern that avoided peak external flexion moments, which would require knee extensor muscle activity, when the knee was near terminal extension. They defined this pattern as the "quadricep avoidance pattern" because any external flexion moment would have to be

balanced by a net quadricep muscle generated internal moment. Thus in level walking where the knee approaches full extension they found the largest percentage change in the external flexion moment at the knee in the ACL deficient knees. During jogging the external flexion moment was also lower in the ACL deficient knees. But in stair climbing, the ACL deficient knees showed no tendency to avoid or reduce the high knee flexion moments. Andriacchi explained these adaptations as a function of the knee flexion angle, since ACL strain decreases as the knee flexes. In the absence of the ACL, a patient may subconsciously avoid contracting the quadricep muscles to avoid displacing the tibia anteriorly.

Joint Laxity and Compressive Loading

The standard clinical means of determining ACL laxity is through such passive manual tests as the anterior drawer, Lachman's 30 degree anterior drawer, and pivot shift tests or via an instrumentation test with a KT-1000 knee arthrometer.[32,84] A respected orthopaedic surgeon, Dr. Hughston, was quoted as saying: "A loose knee's okay if its owner can play well on it. An athlete's knee may be stable when the foot hits the floor and that's what counts. Who cares what the knee is like when it's up in the air during a physical exam?"[46] When the foot is on the ground, as in a closed kinetic chain condition, the knee (and all the joints of the lower extremity) is in a loaded state. The loading is obviously different during open kinetic chain exercises. Studies suggest that axial loading will increase knee joint stability.

Using cadaveric knees, Markolf[61] examined the effects of compressive loads on joint stability at 0 and 20 degrees of knee flexion. He found that compressive loads increase knee stability by limiting rotations and translations. He also reported that removal of the menisci did not affect the stability of the loaded knee.

Hsieh[44] completed a similar study, but tested the knee specimens at 0 and 30 degrees of flexion. His results were the same as Markolf. He proposed that the increased knee stability in load-bearing conditions were secondary to the friction between the joint surfaces, deformation of the cartilage, and the geometrical conformity of the femoral condyles to the tibial plateaus.

Wang[101] studied 27 cadaveric knees for rotary stability during progressive axial loading from 0-100 kg. He reported a progressive decrease in rotatory laxity with progressive increases in axial loading, and that the collateral ligaments were twice as effective as the cruciate ligaments in controlling this laxity.

In a recent study,[57] the effects of joint compression on the anterior drawer stability of the knee in live subjects were explored. Subjects were tested under compressive loads of 0%, 25%, and 50% body weight at both 30 and 90 degrees of knee flexion. They concluded that increased stability resulted from progressively increasing compressive loads, but that no differences were present in either knee angle tested.

While not specifically addressing the effect of loading on knee stability, the data indicate more stability is present during the stance phase than during the swing phase of gait.[54,60] The increased

stability was indicated by the decreased translational motions present.

Marker Attachment Methods

Because of the obvious problems associated with invasive in-vivo fixation of targets, most kinematic data have been obtained by methods which include errors secondary to skin and soft tissue motion. Using skin markers, Macleod[59] measured the distance between the markers (representing a "rigid link") over time during activities. He found that movement between the skin markers was the greatest source of noise in the measurement and would certainly affect the resolution of any measurement system. Because translational motions are small, it has been assumed that any motion analysis system using externally attached targets would be incapable of evaluating translational motions.[18,32]

Because experimentors are primarily restricted to placing markers on the skin, Karlsson[50] compared the motion of markers mounted directly to the bone via skeletal pins, to the motions of markers mounted on the skin by two seperate methods: 1) markers mounted on rigid, acrylic frames strapped to the subjects limbs, and 2) markers mounted on molded, plastic forms held on the subject with a vascular stocking. He discovered that the markers mounted on molded, plastic forms produced similar results to the bone-pin data. Data from the acrylic frames were consistently noisier than data from either of the other methods. Few in-vivo studies have been completed using rigid bone fixations of targets. The study by

Lafortune[54] on 5 normal subjects during walking provides one source for comparison of results. Unfortunately, no other activities were reported.

Literature Review Summary

From the literature, it is apparent that most squat exercise studies have been 2D and completed on weightlifters. Other than the study by Henning,[41] no studies have reported on the direct effect of the squat exercise on the ACL. Because quadricep and hamstring cocontractions occur during the squat, as does vertical loading of the knee joint, it is believed to be an exercise which may prevent or decrease the harmful effects of anterior tibial translation on the ACL. Thus, the squat has been reported as a safe and effective exercise for ACL rehabilitation.[80] Its use in ACL rehabilitation has differed from one protocol to another. This study seeks to compare common biomechanical parameters between normal and ACL deficient knees of the same subject while performing a parallel squat. One of these parameters includes the documentation of translational motions at the knee. Because these motions are small, rigid bone fixation of targets would normally be required to prevent skin motion. However, it has been shown that mounting targets on a plastic mold, and firmly securing this mold to the skin can produce similar results as bone mounted targets.[50]

III. EXPERIMENTAL METHODS

A general description of the experimental methods and techniques used to collect and analyze the data are described in this chapter. All data were collected at the Biomechanics Evaluation Laboratory, Saint Lawrence Hospital Health Science Pavillion, East Lansing, Michigan. Three types of information were experimentally recorded: Kinematic activity of the knee, ground reaction forces, and surface electromyographic signals from four muscles about the knee.

Equipment

Sixteen control points were placed on the boundaries of a calibration space that measured 100.00 x 100.00 x 80.00 cm (Figure 4) and centered over the force plate. Each target in the calibration space was covered with retro-reflective tape (3M scotchlite Corporation) which has a reflectivity 1600 times the reflectivity of a white surface.

Kinematic data were collected using four solid state, shuttered 60 Hz video cameras (NEC) positioned around the calibrated space in such a manner to maximize the number of pixels in the calibration space and to obtain the best view of the later targeted extremity. To accomplish this, cameras were positioned approximately 8 feet from the center of the calibration space, one in each quadrant.

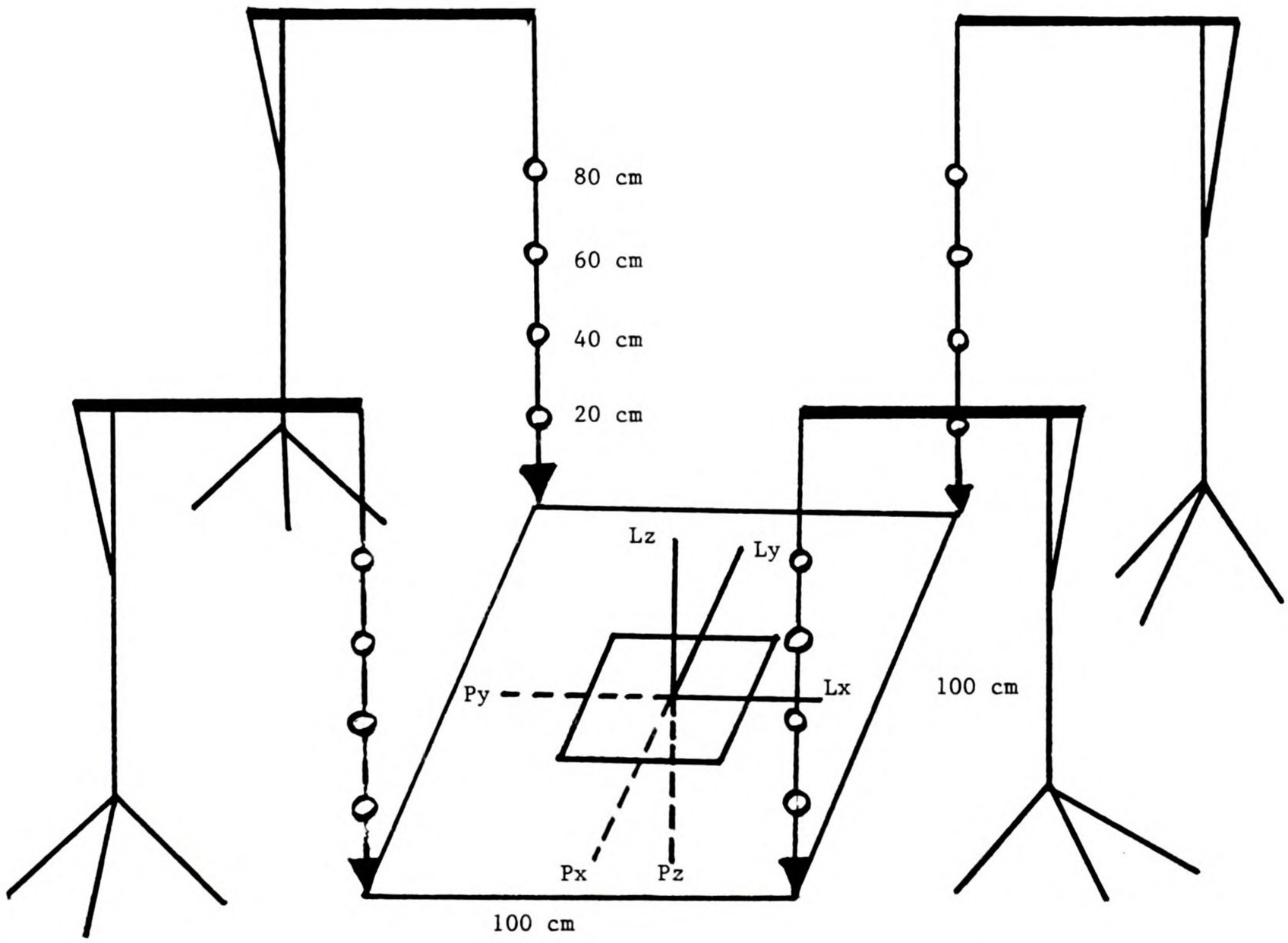


Figure 4. Calibration Space/Lab and Force Plate Coordinate Systems

Illumination of the targets was provided by flood lights attached approximately 2 inches from the center of each camera lens. The proximity of the flood light to the lens enabled the target to reflect light at maximum intensity, as the retro-reflective tape is extremely sensitive to the "observation angle." The observation angle is the angle between the incidence light ray, the reflective target, and the reflective ray returning to the camera lens.

All four cameras were synchronized by a VP-320 model dynamic processor (Motion Analysis Corporation). Each target location was determined in pixel space on the VP-320. Using the Expertvision three dimensional (EV3D) digitizing program, the centroid location of each target was located. The accuracy of the calibrated space was reported as a "norm of residuals" for each camera. The closer the residuals are for each camera, the more accurate the calibration. The residual value differences were within .15 for all testing sessions. Also, all residual values were less than .39, which fell well below the system requirements for residual values of less than 2.0, indicating an accurate calibration. The calibrated space provided a known laboratory coordinate system with the center of the force plate surface as its origin (Figure 4). Using a method of direct linear transformation, the transformation matrices were determined and stored in the environmental operator section of EV3D.

Ground reaction forces (Fx, Fy, Fz) and the moments about the instrument center were measured using an AMTI Biomechanic Force Platform Model OR6-6. The force platform, mounted in the floor, incorporated strain gauges which measure the applied forces,

amplifies the signal, and sends it to the analog to digital converter. The signal was sampled at a rate of 1000 Hz and stored on the Sun 4/260C work station. The orientation of the force plate coordinate system is also shown in Figure 4.

The electromyographic signals were collected via surface silver/silver chloride electrodes (Protrace) and telemetered to a Transkinetics receiver. The raw signals for each muscle were stored on the Sun work station.

Pilot Study

In an effort to decrease noise from skin motion, two methods of target attachment were studied: A triad of targets mounted directly on the skin and a triad mounted on thermo-plastic material molded over the limb segment and strapped to that segment. Movement was assessed at both the thigh and leg segments by comparing the changes in angles (Figure 5) formed by the triads over time while moving the knee through a full range of motion during a squat exercise. On the thigh, skin mounted targets yielded angle differences ranging from 12.06-18.17 degrees while the molded plastic reduced these differences to 3.07-6.10 degrees. At the leg, differences were less. The skin mounted targets produced angle differences of 3.16-7.66 degrees, while the molded plastic differences were 1.38-5.34 degrees. A decision was made to conduct the study with the thermoplastic mold on the thigh only, since this was the segment which produced the greatest decrease in motion.

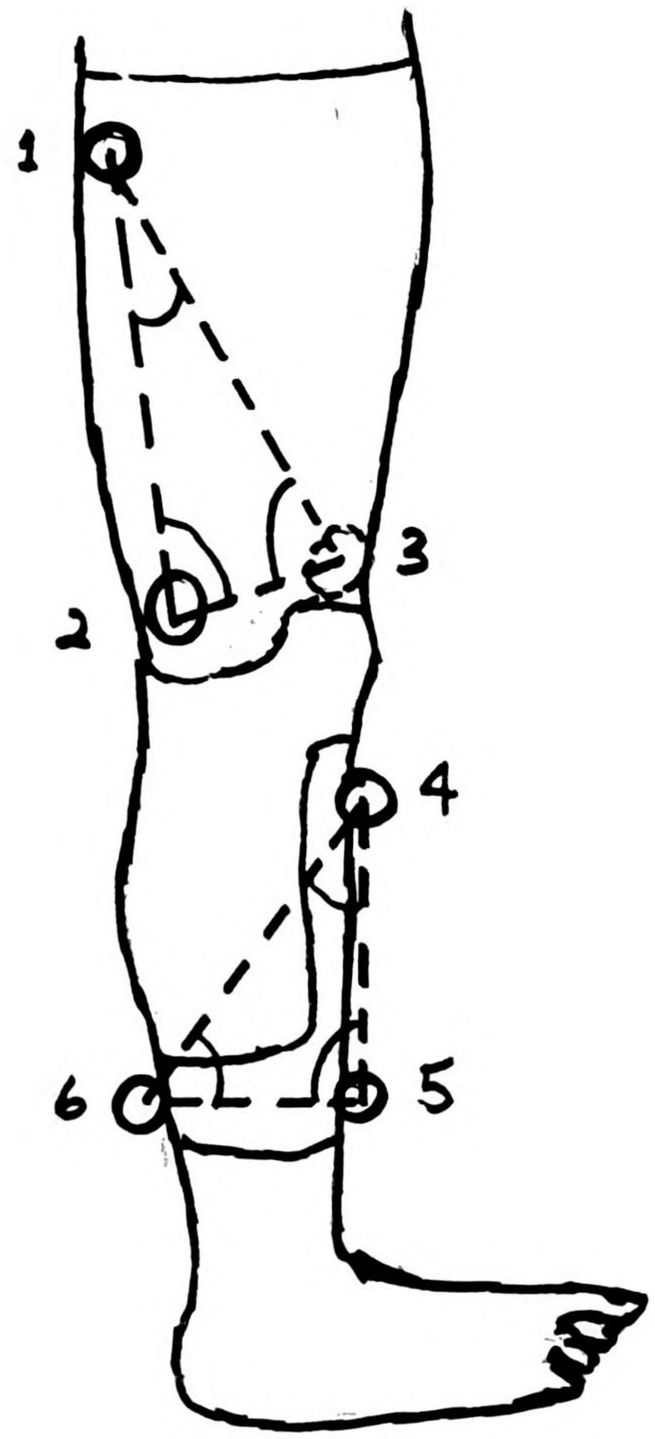


Figure 5. Angles Formed By Target Triads

Subjects

The subjects for this study consisted of 3 males and 3 females. The subjects descriptive data are included in Table 1 and a summary of surgeries and knee injuries, as reported by the subjects, is listed in Table 2. Each subject had a self reported chronic unilateral ACL deficiency. Prior to participation in the study, each subject was evaluated for the presence of passive ACL deficiency in both knees. Passive clinical laxity tests were utilized and included a 30 degree Lachman's, a 90 degree Anterior Drawer, and a pivot shift. Each subject included in this study demonstrated positive tests in the ACL deficient knee and negative results in the control knee.

Table 1. Subject Characteristics

Subject	Sex	Age	Weight(kg)	Height (cm)
TR	F	23	57.73	168.91
EM	M	34	93.64	182.25
MP	M	22	80.91	175.26
MC	F	20	63.18	165.10
RL	M	19	118.64	181.61
LH	F	22	59.77	160.02

Table 2. Knee Diagnoses and Surgeries

Subject	Diagnosis	Surgeries
TR	Complete ACL Tear (L) Grade II Lateral Collateral	1986 Arthroscopy
EM	Complete ACL Tear (L)	1989 Arthrotomy and partial meniscectomy
MP	Partial ACL Tear (R)	1985 Arthroscopy 1989 Part. meniscectomy
MC	Complete ACL Tear (L) Reinjured 1989	1988 Acute Repair
RL	Complete ACL Tear (L)	1989 Acute Repair
LH	Complete ACL Tear (R)	1987 Arthroscopy

Experimental Preparation

Prior to testing, each subject signed an informed consent and was briefed on the testing procedure. One of the lower extremities was selected at random and thermoplastic material was cut, heated, and molded over the distal one-half of the thigh. When cooled, the mold was removed. The muscle bellies of four muscles (rectus femoris, biceps femoris, medial hamstrings, and gastrocnemius) were palpated and identified. Electrode placement locations were determined by palpation and the areas were prepared by shaving and firm buffing with a dry cloth. The electrodes were placed parallel to the muscle fibers and approximately 25 mm apart. The electrodes were attached to the transmitters which were then attached to the skin in such a manner to avoid bending the electrode cables or

contact with the mold. The thermoplastic mold was again positioned over the distal thigh. To firmly attach the mold, electrodes, and transmitters to the body segments, the entire thigh and leg were tightly wrapped with an elastic wrap (Coban, 3M). The wrap alone without a mold was felt to be adequate to restrict soft tissue motion in the leg. The subject was then taken to the calibrated space in the laboratory and all transmitters were tuned to eliminate the majority of environmental noise.

Six spherical retro-reflective targets were placed on the subject (Figure 6) in the following locations (all thigh targets were placed on the mold):

1) Lateral Thigh (on a line between the greater trochanter and lateral femoral epicondyle)
2) Lateral Femoral Epicondyle
3) Medial Femoral Epicondyle
4) Proximal Anterior Tibia (beneath the tibial tubercle)
5) Distal Anterior Tibia
6) Distal Posterior Tibia

Using EV3D, links were established between targets 1-2, 2-3, 4-5, and 5-6. These target links were the basis for forming the body coordinate system according to the joint coordinate system.[38]

Experimental Protocol

The starting position involved placing the foot of the tested extremity in the center of the force plate and oriented in the anterior axis direction. From this initial foot placement, the subject stood erect with the feet comfortably spaced approximately

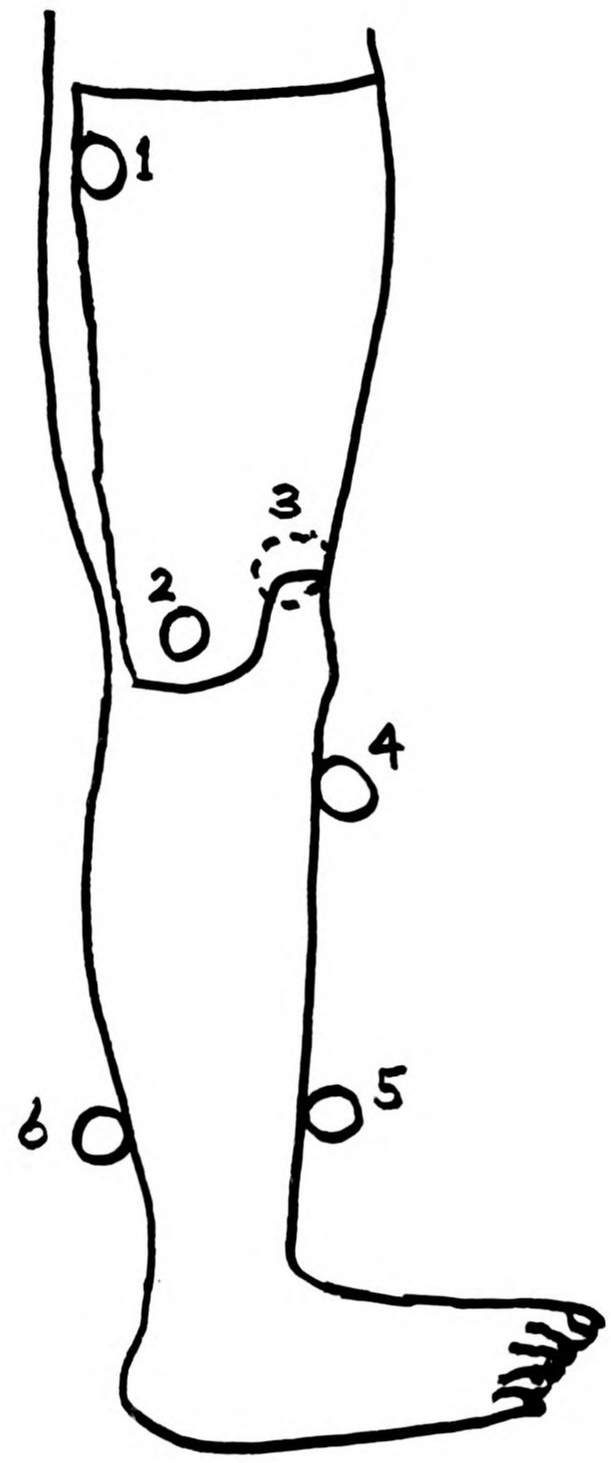


Figure 6. Target Placement with Thigh Mold

hip width apart and with a "normal/comfortable" amount of hip external rotation. The subject was instructed to perform three to four consecutive squat repititions which involved descending until the thigh was approximately parallel to the floor and ascending to the fully erect position. A limitation of the system required that for 1000 Hz sampling of force data with EMG, the maximum force plate collection time was 2 seconds. Therefore, each subject was instructed to perform each squat repitition at a constant rate of descent and ascent, but at a completion rate of 2 seconds or slightly less. Video data collection was preset for 6 seconds, and EMG data were automatically collected from 5 seconds prior to triggering the forceplate, until 5 seconds after triggering. After a period of practice, data collection began.

As the subject stood erect in the starting position, video collection was manually triggered. Thus, the initial frames of each file recorded the zero position for each trial and allowed for targeting variations between extremities and subjects. Immediately following triggering the video collection, the subject was told to "START." As close as possible to the completion of the first squat repitition, the force plate was manually triggered to begin its 2 second collection cycle. When the force plate was triggered, an event marker was automatically placed on each raw video and EMG file to allow syncronization of all components. Immediately after the trial, the ground reaction forces and EMG results could be viewed. Ground reaction forces from three successive trials could be overlayed and viewed. Figure 7 demonstrates the reproducibility of the force data during three successive trials in subject EM.

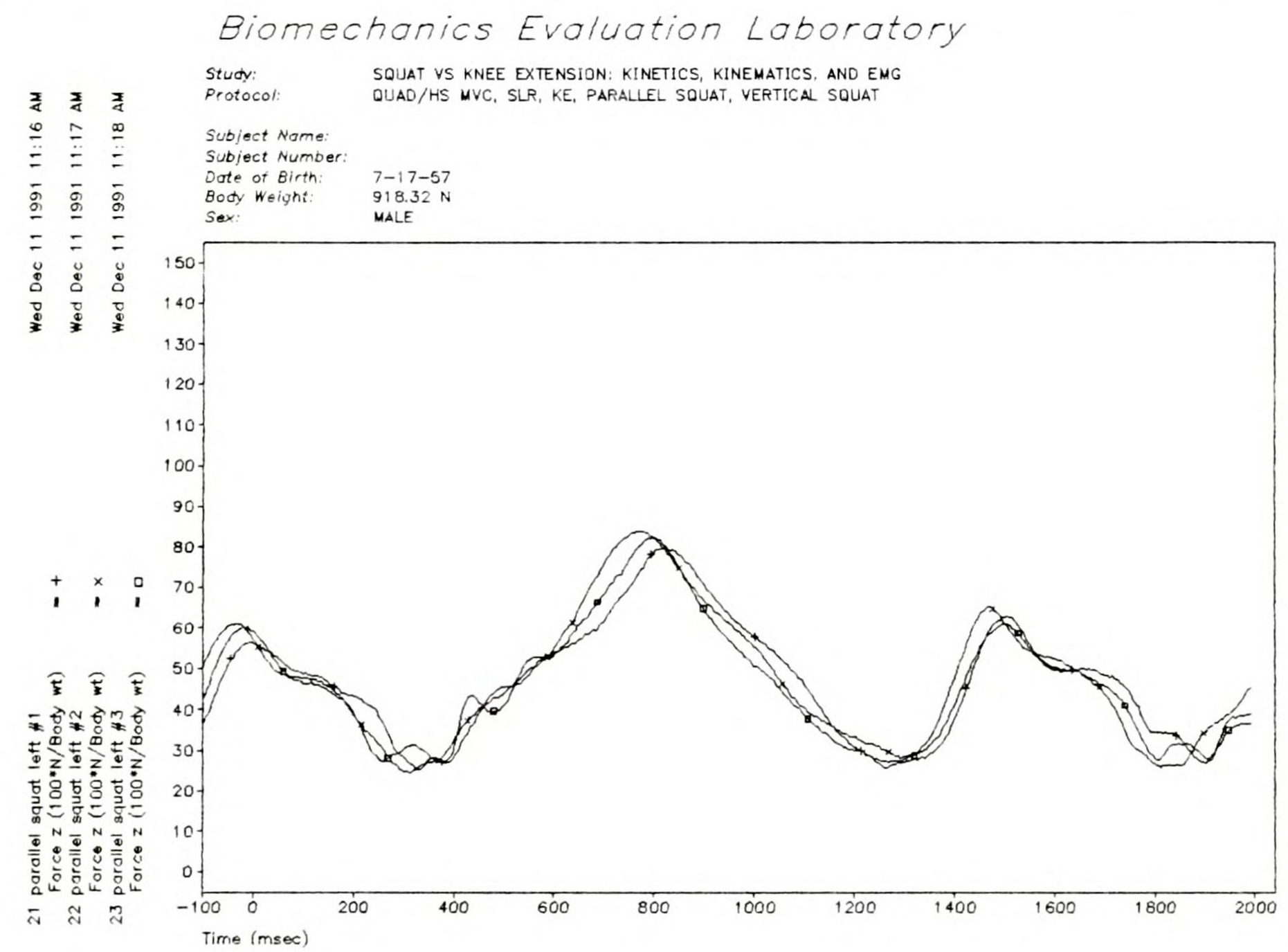


Figure 7. Ground Reaction Forces (Fz) from Three Successive Trials

A successful trial consisted of completion of the squat repitition as previously described. Following completion of all three trials, the preparation and testing processes were repeated on the contralateral lower extremity.

IV ANALYTICAL METHODS

After completion of the data collection, the video files required additional processing. As previously described, targets were placed at three points on the thigh and leg. These points established segment coordinate base vectors. Each video file was tracked using the track operator of the EV3D software. The stick figure option was utilized which allowed the targets to be viewed as rigid links. The output of the tracked motion files were the coordinates with time of each target or trajectory data, where each target had a three dimensional path of motion. Since these coordinates were measured relative to the laboratory coordinate system, they were considered as components of position vectors to each target. For example, let **P**4 be the position vector to the 4th target (Figure 8). The target trajectories were smoothed using the EV3D track editor operator. The track edited files were then processed using the MAP1 software program which collated the data so that all target positions were grouped positionally according to each increment of time.

The mapped positional data were then converted into local or segment coordinates. The thigh segment base vectors were defined as follows:[38]

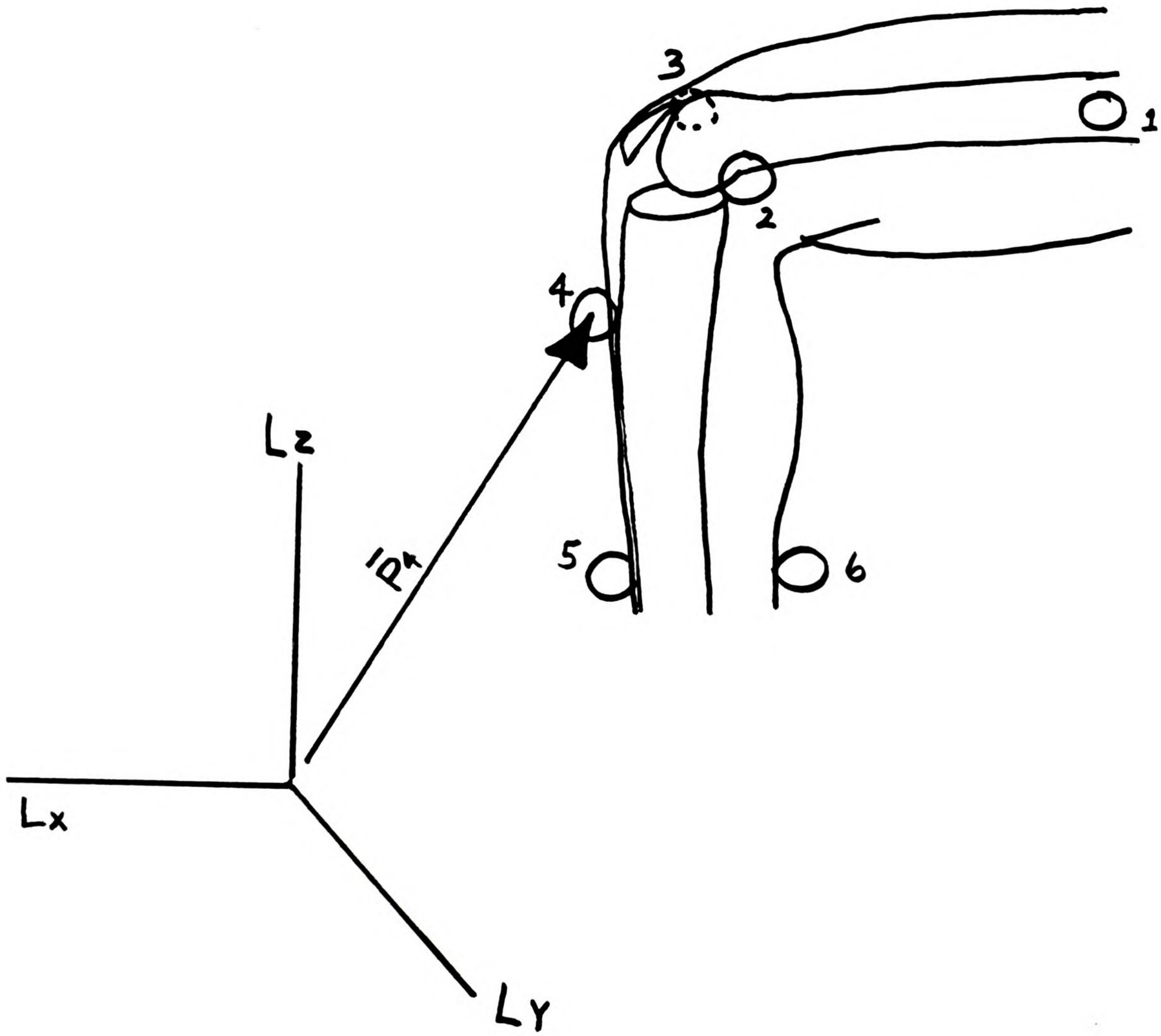


Figure 8. Position Vectors in Laboratory Coordinates

$$I_Z = (P1 - P2) / \ |(P1 - P2)|$$

$$I_X = I_Z \times (P3 - P2) \ / \|I_Z \times (P3 - P2)| \text{ (for the right side)}$$

$$= I_Z \times (P2 - P3) \ / \|I_Z \times (P2 - P3)| \text{ (for the left side)}$$

$$I_y = I_Z \times I_X$$

The leg segment base vectors were defined as follows:

$$i_Z = (P4 - P5) / |P4 - P5|$$

$$i_y = i_Z \times (P5 - P6| / |i_Z \times (P5 - P6)|$$

$$i_X = i_y \times i_Z$$

The knee angles were defined by Euler angles using the Grood and Suntay[38] "Joint Coordinate Axes." The joint coordinate axes were chosen as (fig. 3):

$$e3 = i_Z$$

$$e2 = I_y$$

$$e1 = (e2 \times e3) / |e2 \times e3|$$

Flexion and extension, as a function of time, were defined as (with flexion positive):

$$fe(t) = -\sin\ (e1 \bullet \mathbf{I}_Z)$$

Abduction and adduction, as a function of time, were defined as (with abduction positive):

$$ab(t) = \sin\ (e2 \bullet e3) \text{ for the right leg}$$
$$= -\sin\ (e2 \bullet e3) \text{ for the left leg}$$

External and internal rotation of the tibia with respect to the femur, as a function of time, were defined as (with external rotation as positive):

$$er(t) = -\sin\ (e1 \bullet \mathbf{I}_Y) \text{ for the right knee}$$
$$= \sin\ (e1 \bullet \mathbf{I}_Y) \text{ for the left knee}$$

The outputs of this program were then graphed to determine the three rotational displacements (flexion/extension, abduction/adduction, and internal/external rotation) of the knee over time. Crossplots of flexion vs abduction/adduction, and flexion vs rotation were also produced for pattern comparisons.

At this point a decision was made to describe the translational motions in tibial coordinates instead of joint coordinates. The reason for this choice was that most clinicians think of anterior/posterior translations as being approximately parallel to

the tibial plateau. A tibial based coordinate system would describe translations in this manner. To accomplish this description, the following method was utilized (Figure 9):

1) The midpoint between the lateral and medial femoral epicondyle targets was defined by the position vector **D**, where

$$\mathbf{D} = (\mathbf{P2} + \mathbf{P3}) / 2$$

2) The first frames of data in the motion file were for standing with the knees extended. These position data were used as a neutral reference of translation of the femur relative to the tibia.

3) A relative position vector **R** was created for this neutral position from **P4** to a point on the "tibia extended" that coincided with the femoral point **D** in the neutral position.

$$\mathbf{R} = \mathbf{D} - \mathbf{P4}$$

4) Since there is no single point of rotation for the rolling and sliding of the knee, the reference point for relative translation was selected as **D** on the femur and an aligned point (**P4** + **R**) on the tibia extended.

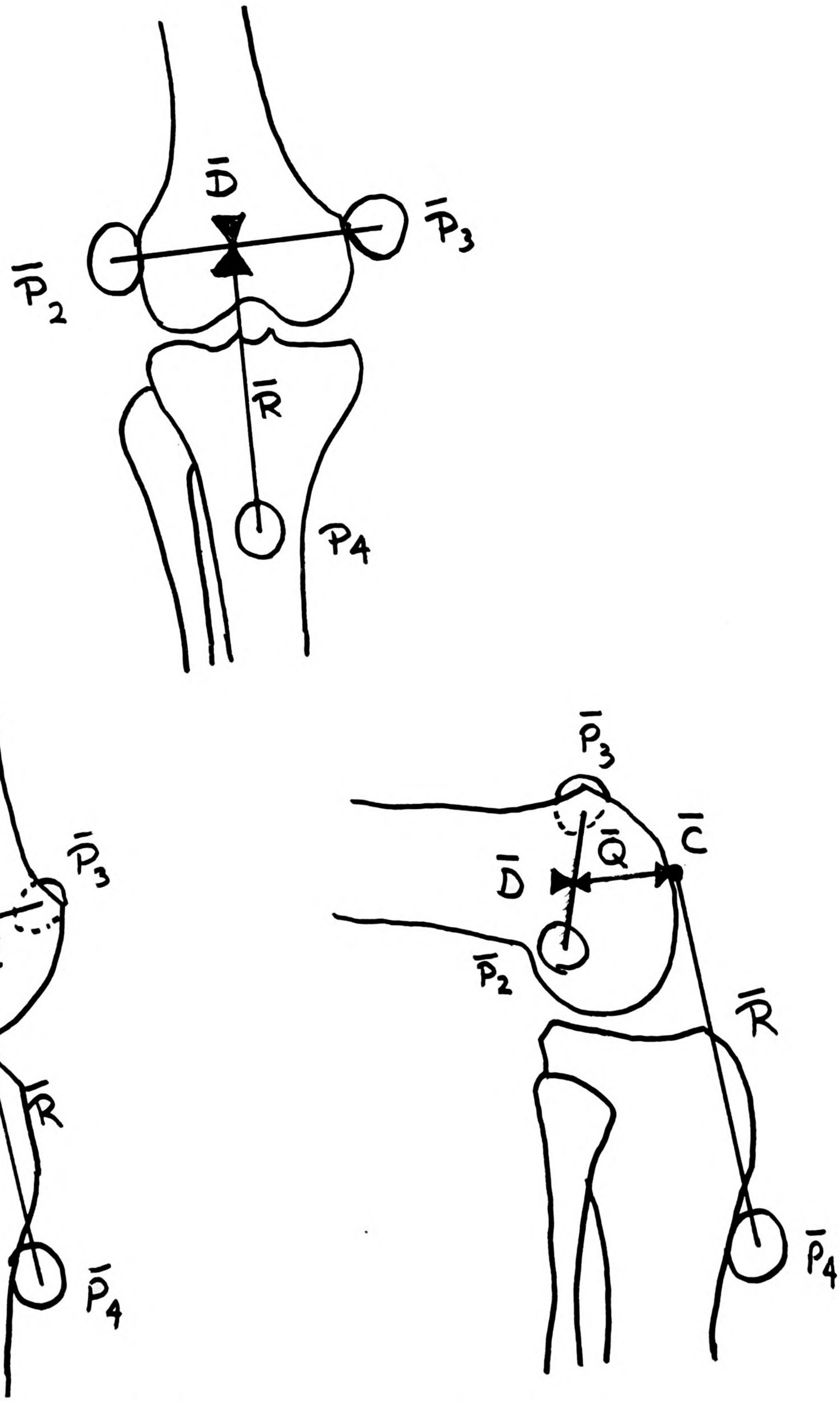


Figure 9. Description of Translational Method

5) The transformation matrix from laboratory coordinates to tibial coordinates for any time was written as:

$$M(t) = \begin{bmatrix} l_{xx} & l_{xy} & l_{xz} \\ l_{yx} & l_{yy} & l_{yz} \\ l_{zx} & l_{zy} & l_{zz} \end{bmatrix}$$

where l_{xy}, for example, is the y component of $\mathbf{i}_x$ in laboratory coordinates.

6) The position vector **R** (from the neutral position) may be written in tibial coordinates as:

$$\mathbf{R}' = [M]_n \, \mathbf{R}$$

where M_n is the transformation matrix for neutral position.

7) The corresponding position of a point on the tibia extended, **C**, for any time was defined as:

$$\mathbf{C}(t) = \mathbf{P}4(t) + [M]^T(t) \, \mathbf{R}'$$

8) The femoral reference point, **D**(t), was defined as before:

$$\mathbf{D}(t) = [\mathbf{P}2(t) + \mathbf{P}3(t)] \; / \; 2$$

9) Translation of D(t) relative to C(t) at the knee may be written as:

$$\mathbf{Q}(t) = \mathbf{D}(t) - \mathbf{C}(t)$$

10) This may be expressed in tibial coordinates as:

$$\mathbf{Q}'(t) = [M](t)\ \mathbf{Q}(t)$$

This mathematical process for describing the motion of the femur relative to the tibia is equivalent to permanently fixing a rigid rod to the proximal tibia at the site of target number 4 which is oriented exactly to the midpoint of the medial and lateral femoral epicondyle targets during the first frame of the video file. The "tibia extended" then refers to the end of this rod and the relative motion between the femoral reference point (**D**) and the tibia extended (**C**) can be determined over time.

The outputs of this program were then plotted as translations over time and as crossplots of degrees flexion versus translations. Comparisons were then made between the amount of translations and patterns produced by the normal and ACL deficient knees.

Three dimensional ground reaction forces (GRF) were taken directly from the force plate as percent bodyweight, printed in hard copy, and analyzed for the mean, maximum and minimum values and for time of occurrence. Comparisons were made between the normal and ACL deficient knees.

The moment was considered to act about the center of the joint which again was selected as the midpoint between the lateral and medial femoral epicondyles. This moment about the knee was defined as:

$$\mathbf{M}_k = \mathbf{J} \times \mathbf{F} + \mathbf{T}$$

where **J** = a vector from the knee joint center to the center of pressure (COP).

F = ground reaction force (GRF)

T = torque on the force plate surface

For the purpose of this analysis, the inertial effects were ignored and each event (instance in time) was treated as quasi static. The moments about the knee were determined by inputing the previously mentioned mapped files along with the force files. Since the video data were collected at 60 Hz and the force data at 1000 Hz, the moment program initially synchronized the two seperate data sources to 100 Hz. To make comparisons between subjects, the magnitude of the moments were normalized by:

$$|\mathbf{M}_k| \; / \; \%(BW)(HGT)$$

where $|\mathbf{M}_k|$ = the magnitude of the knee moment

BW = bodyweight

HGT = height

This normalization reduced the variability of the moment that could be associated with subjects of different weights (force) and heights (length). Moments reported are external or applied moments, so the internal or muscle moments are considered as equal and opposite. Thus, if a flexion moment was applied, it was assumed than an equal, but opposite extension moment from the quadriceps was necessary to maintain equilibrium.

Raw EMG data were analyzed for trends only. At the time of this study, further processing was not possible to make quantitative statements.

Statistical analysis was performed in two steps. First, Fischer's f-test[34] was used to check for the presence of homogeneous variance between group data. Homogeneous variance was assumned present if the null hypothesis was accepted ($p>.20$). Secondly, if homogeneous variance was present, a two-sample t-test[34] was used to test for differences between means. If heterogeneous variance was present, an alternate form[34] of the two-sample t-test was utilized. Mean differences were considered statistically significant at $p<.05$.

V. RESULTS

For this study, each subject performed 3 sets of squat repetitions. While video data were captured for all repetitions, the force plate could only capture one repetition. The corresponding video repetition was used for kinematic comparisons.

Kinematics

Because each knee was tested seperately, a concern was that the mean durations for the squat repetitions might differ between groups. The total squat durations on the normal knees ranged from 1.3 to 2.0 seconds ($x = 1.66 \pm .22$ SD), and from 1.5 to 2.0 seconds ($x = 1.67 \pm .17$ SD) on the ACL deficient knees. Using the Fischer f-test, the assumption of homogeneous variance proved true ($p>.20$), so the two-sample t-test was utilized.[34] No statistically significant differences were found between the squat durations of either group.

For both groups, the typical motion coupling pattern consisted of tibial external rotation and aduction with flexion, and tibial internal rotation and abduction with extension (Figure 10). In addition, the majority of subjects demonstrated a small amount of external rotation in terminal extension. The mean rotational motions for each group are presented in Table 3.

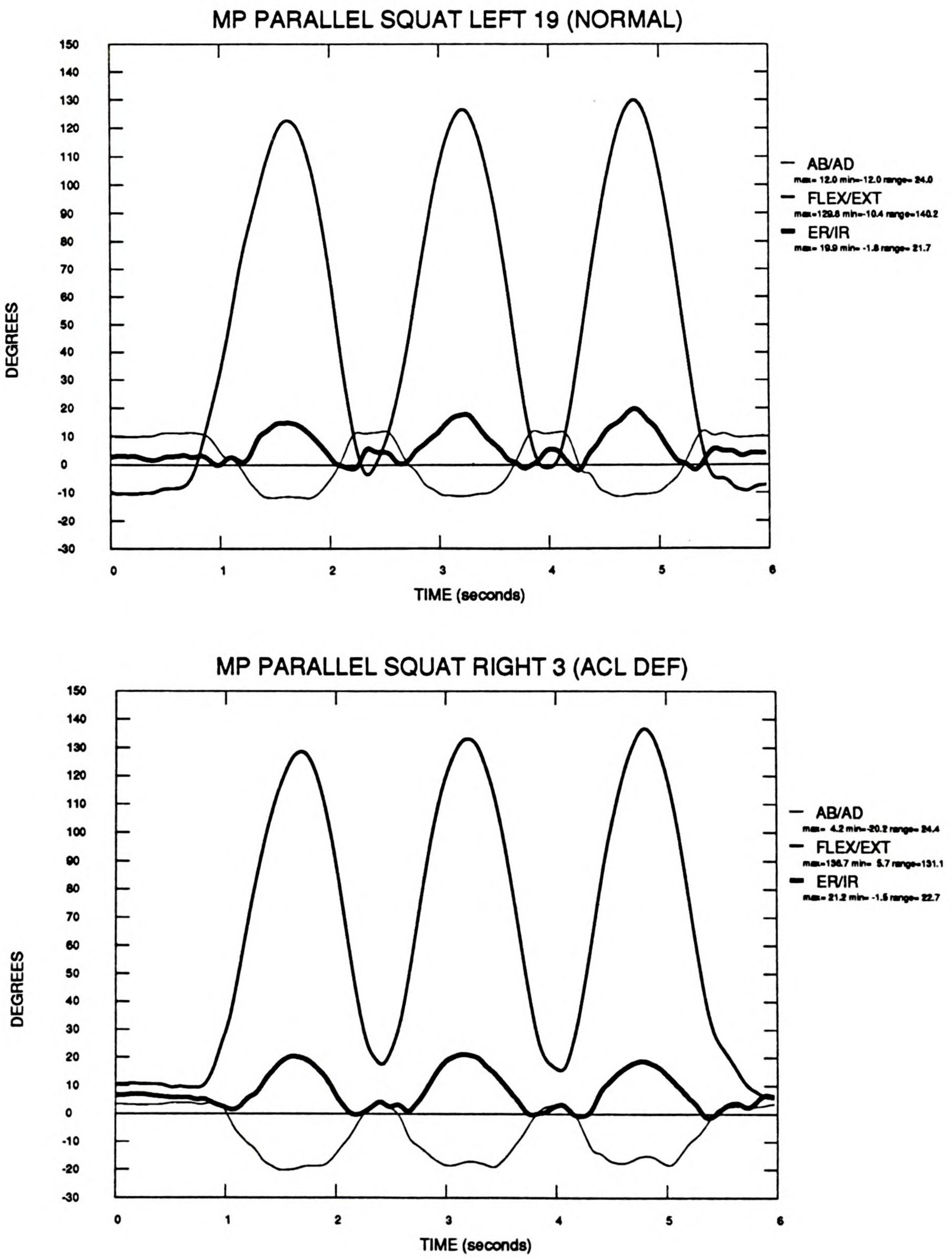


Figure 10. Typical Motion Coupling Pattern Over Time

Table 3. Mean Rotational Motions (Degrees)

	Flex/Ext	Ext/Int Rotation	Ab/Ad
Normal	121.67 ± 11.06 SD	14.67 ± 4.39 SD	19.06 ± 2.24 SD
ACL Def.	122.33 ± 9.40 SD	17.88 ± 4.63 SD	20.06 ± 4.49 SD

Typical crossplots of flexion versus rotation and flexion versus abduction/adduction are seen in Figures 11 and 12. Preliminary f-tests again indicated that homogeneous variance was present between groups ($p>.50$), so mean differences were tested using the two-sample t-test. No statistical differences were found for any motion between groups ($p>.05$); although, it appears that the ACL deficient group demonstrated more external/internal rotation, while performing a squat repitition, than the normals (Table 3).

Translations of the femoral reference point, relative to the tibia "extended", along the three axes of the tibial coordinate system were calculated as described in the methods section. However, graphs of these motions refer to the tibia translating relative to the femur. Graphs of anterior-posterior translations over time are presented in figure 13. Crossplots of degrees flexion versus anterioposterior (AP) tibial translations are seen in Figures 14 and 15. Crossplots of degrees flexion versus mediolateral (ML) (Figures16 and 17) and superioinferior (SI) translations (Figures. 18 and 19) are also presented. The mean translations (cm) are presented in Table 4.

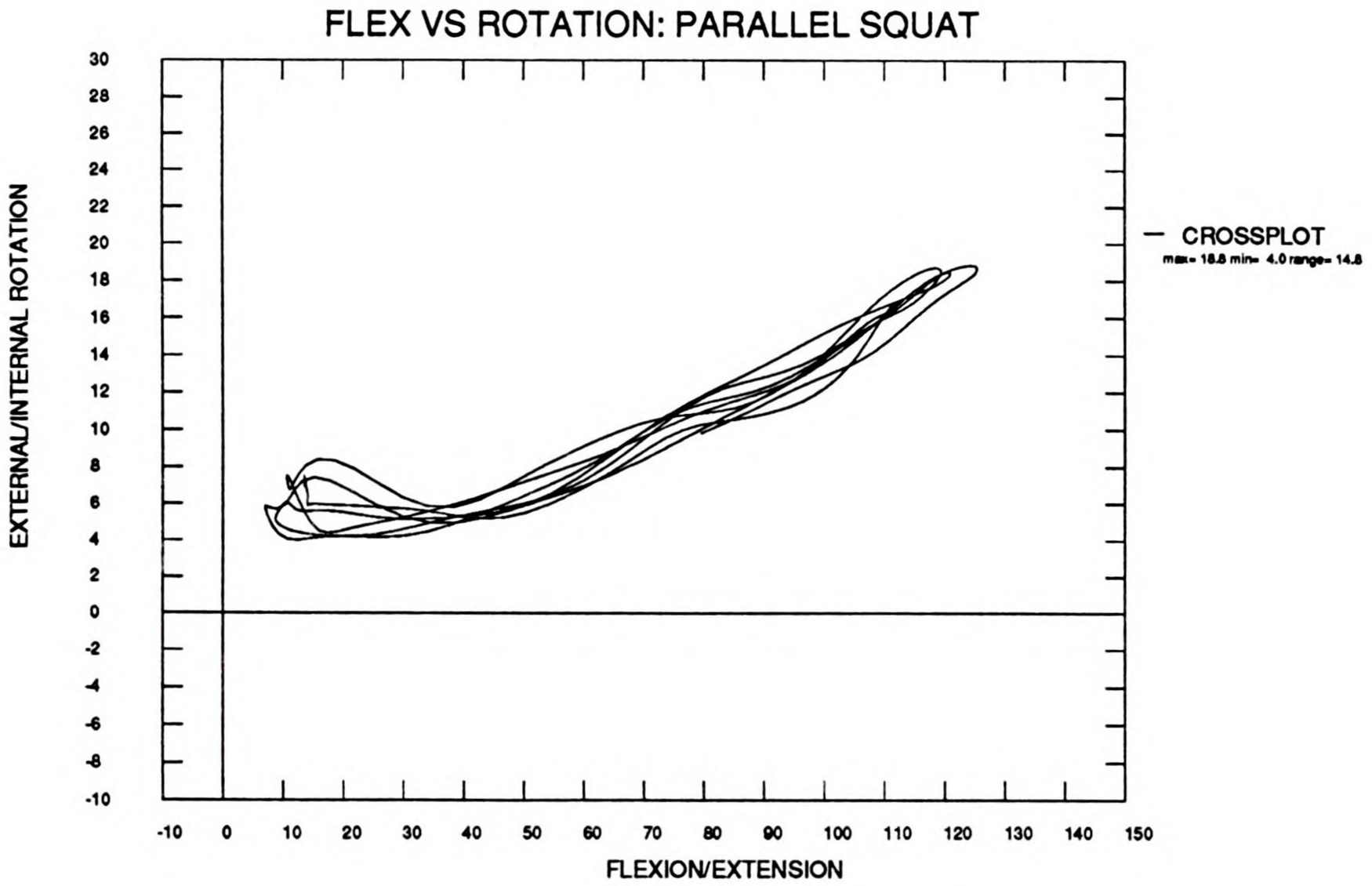


Figure 11. Typical Crossplot of Flexion vs Rotation

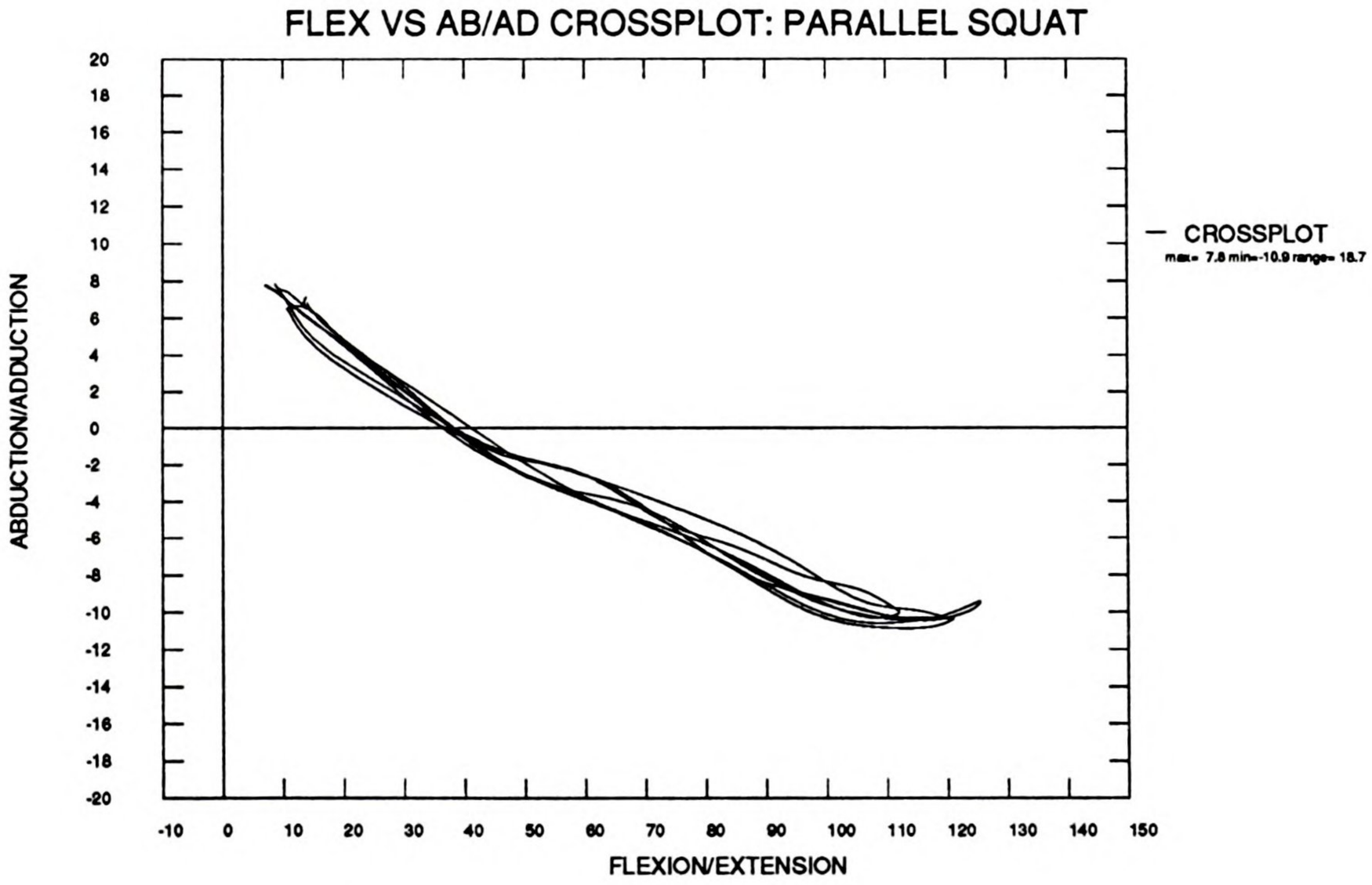


Figure 12. Typical Crossplot of Flexion vs Ab/Adduction

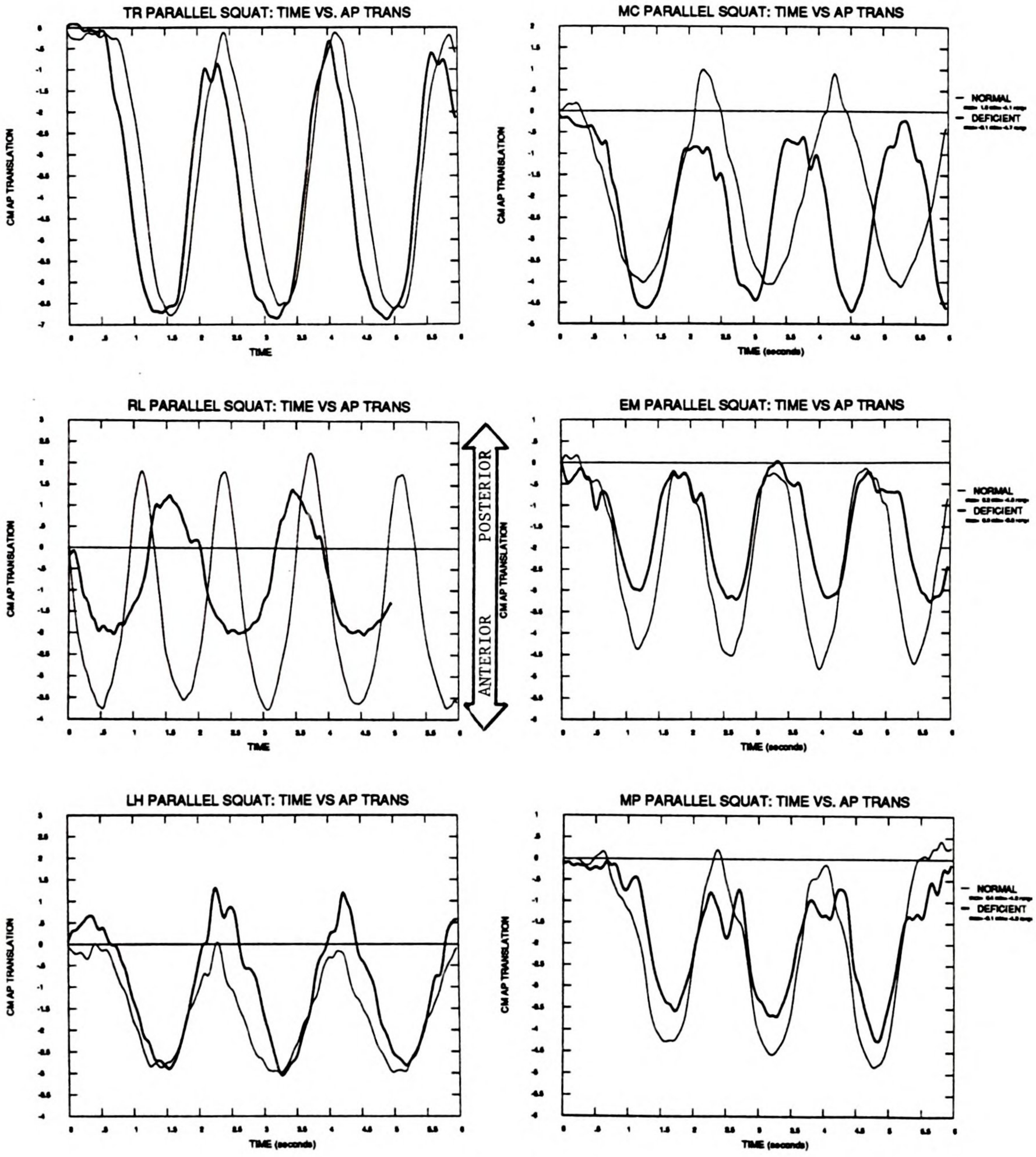


Figure 13. AP Translations over Time during Squat Repetitions

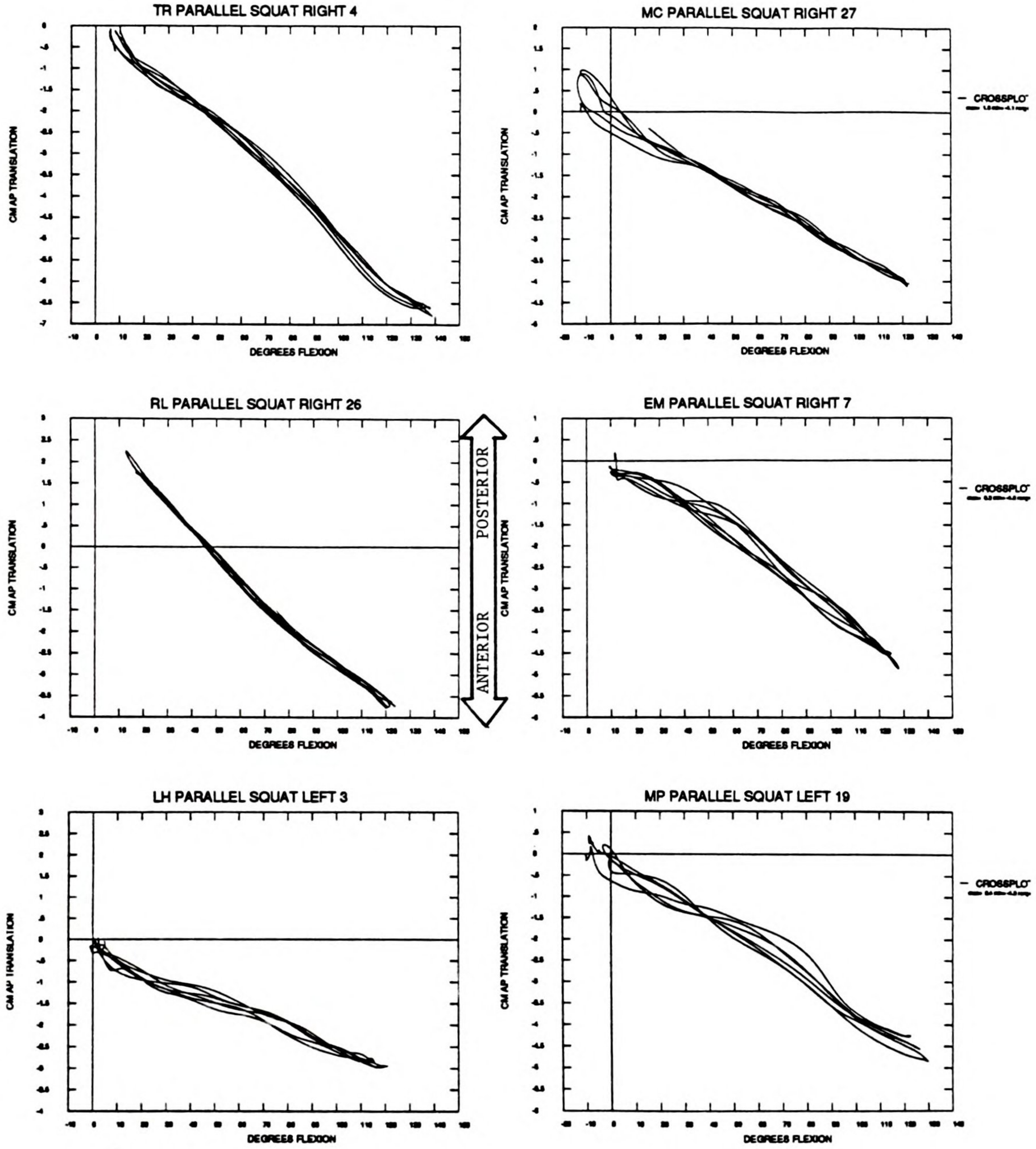


Figure 14. Crossplots of Flexion vs AP Translation (Normals)

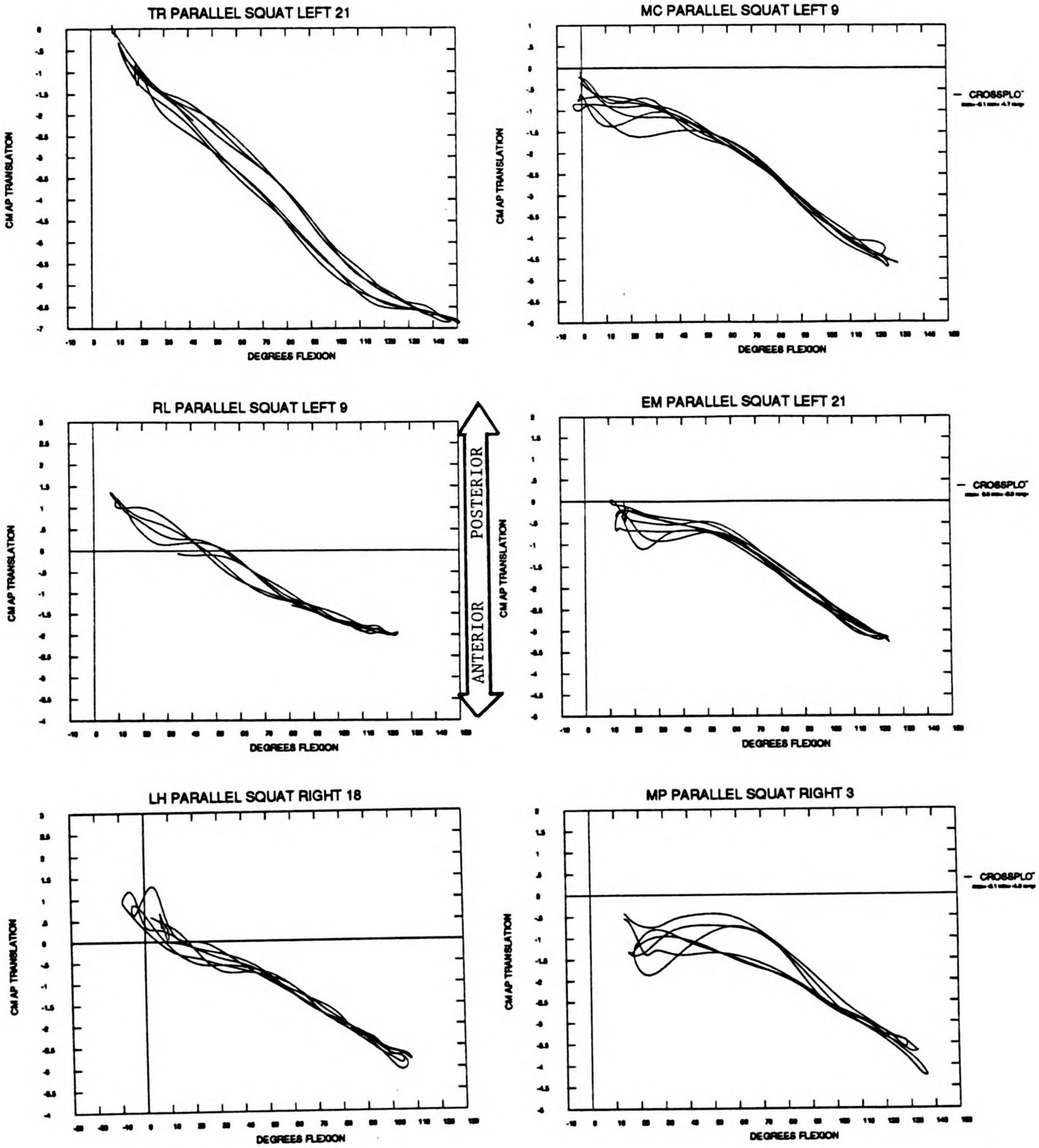


Figure 15. Crossplots of Flexion vs AP Translation (ACL Deficient)

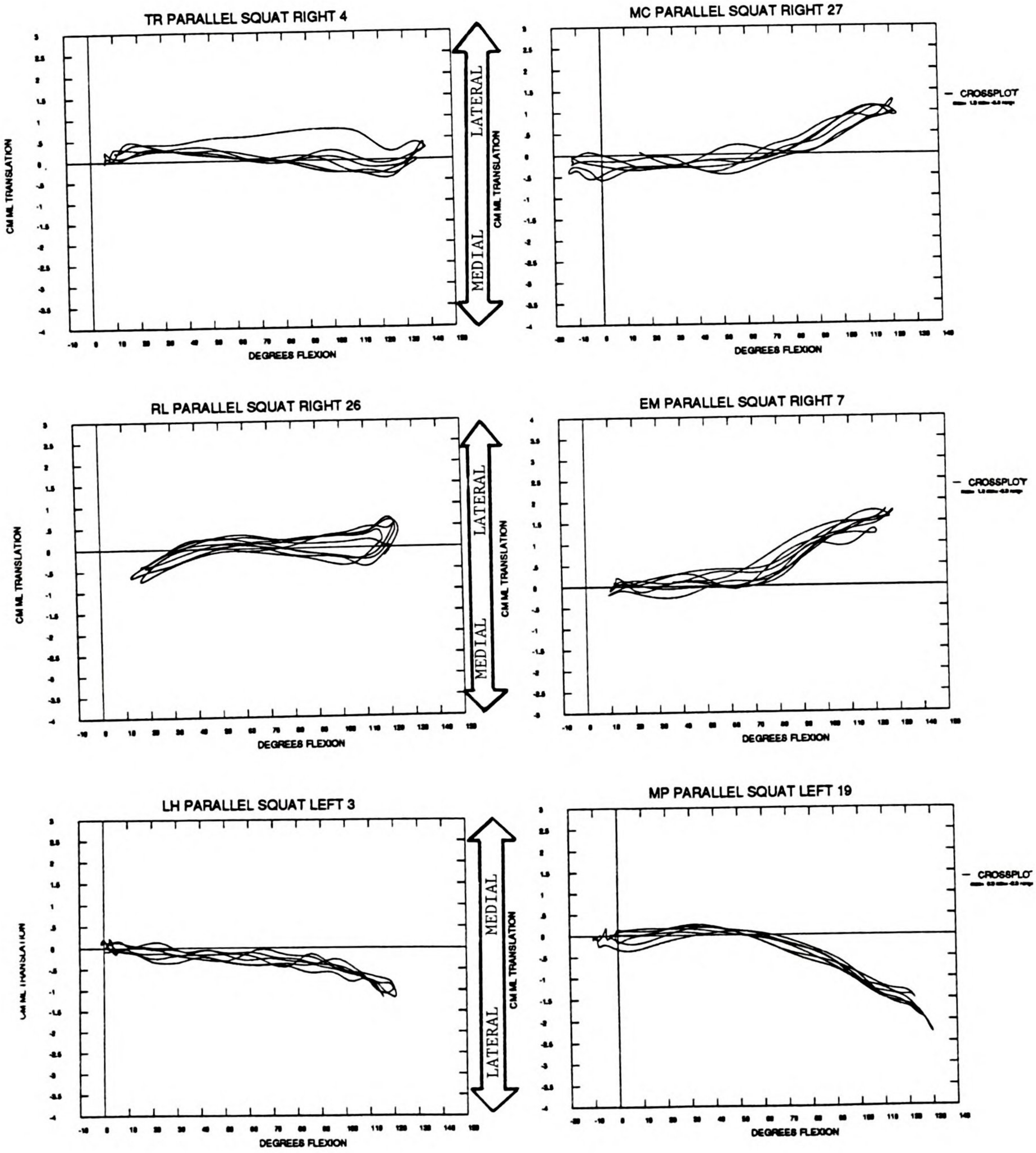


Figure 16. Crossplots of Flexion vs ML Translation (Normals)

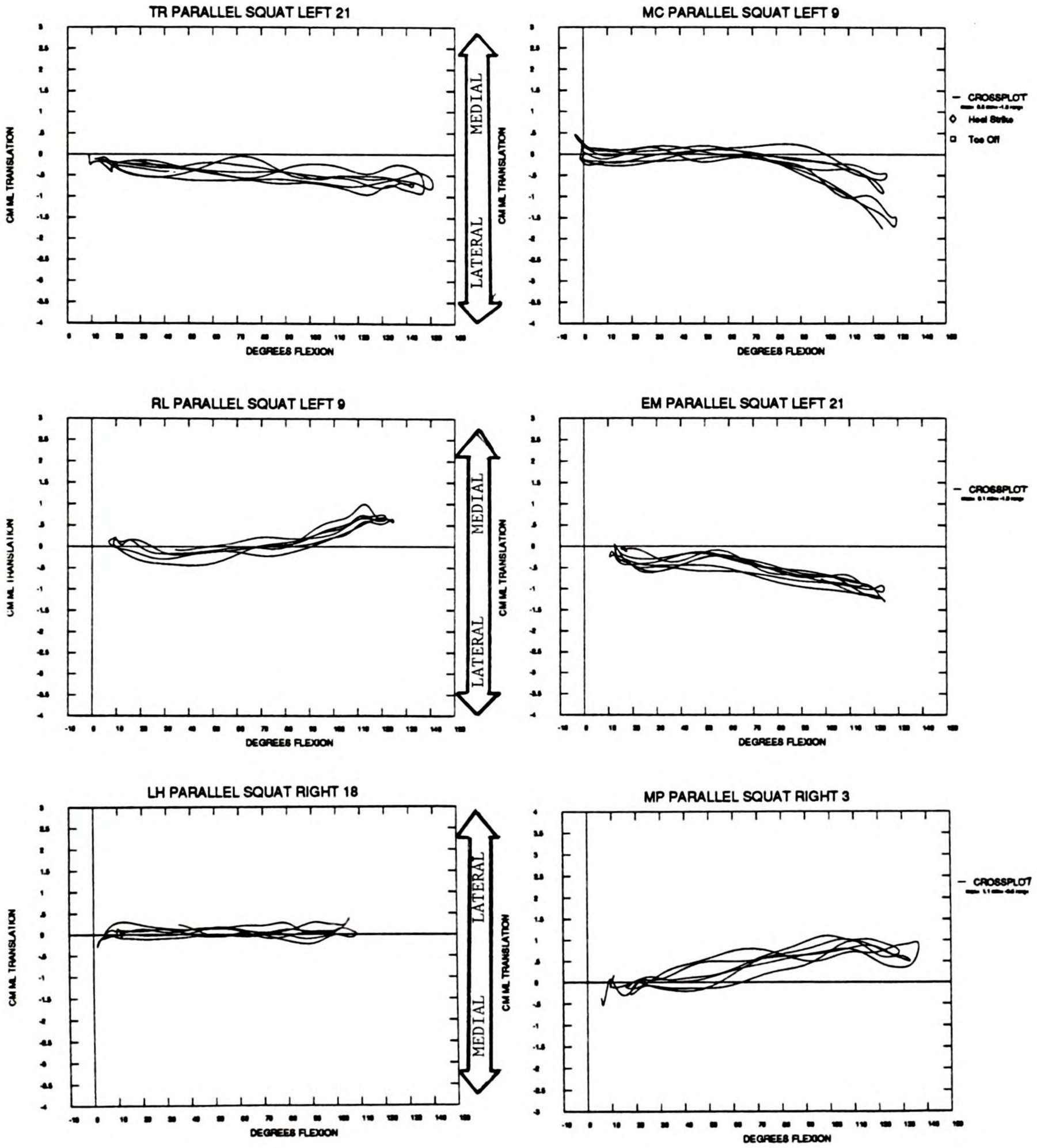


Figure 17. Crossplots of Flexion vs ML Translation (ACL Deficient)

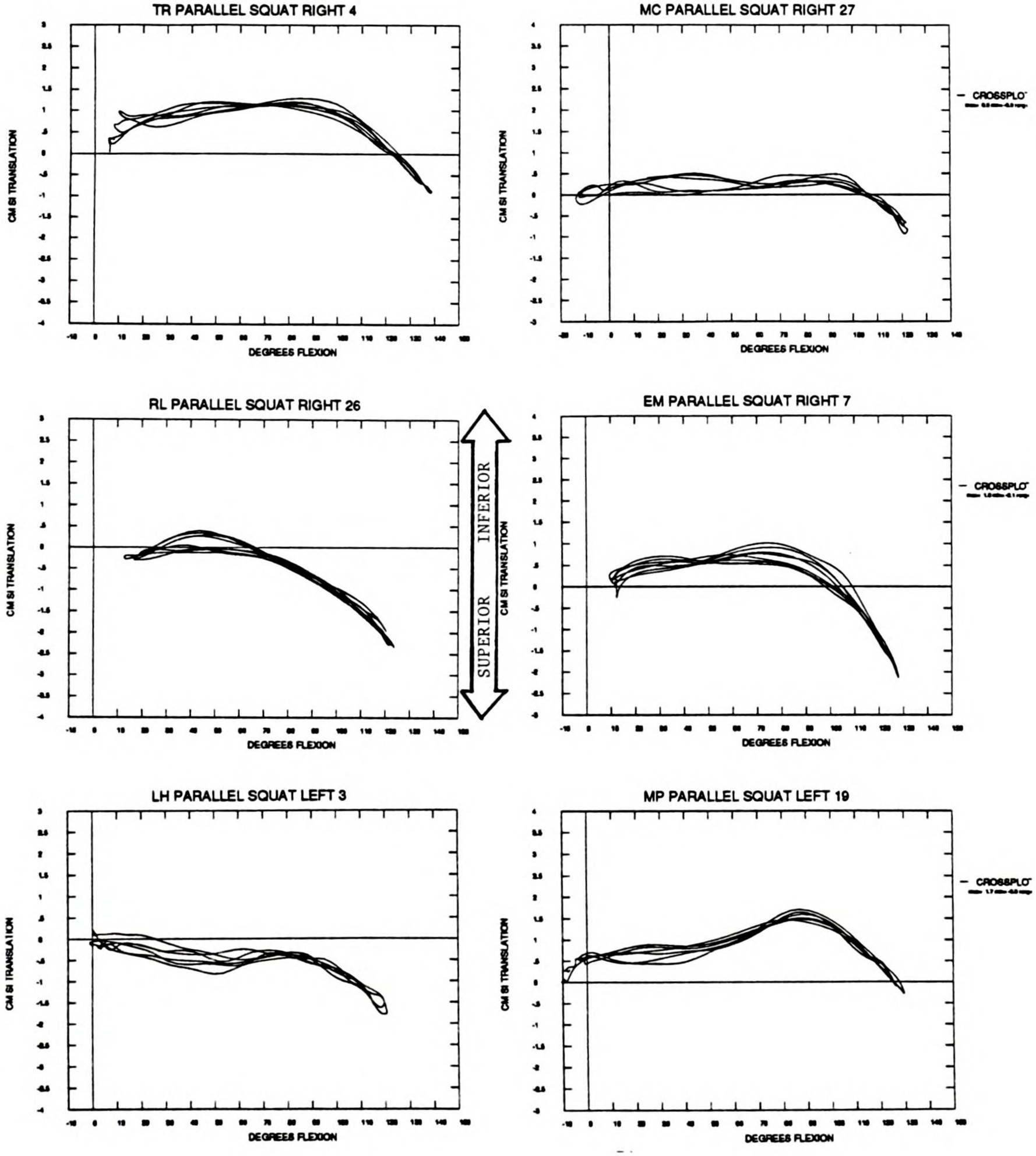


Figure 18. Crossplots of Flexion vs SI Translation (Normals)

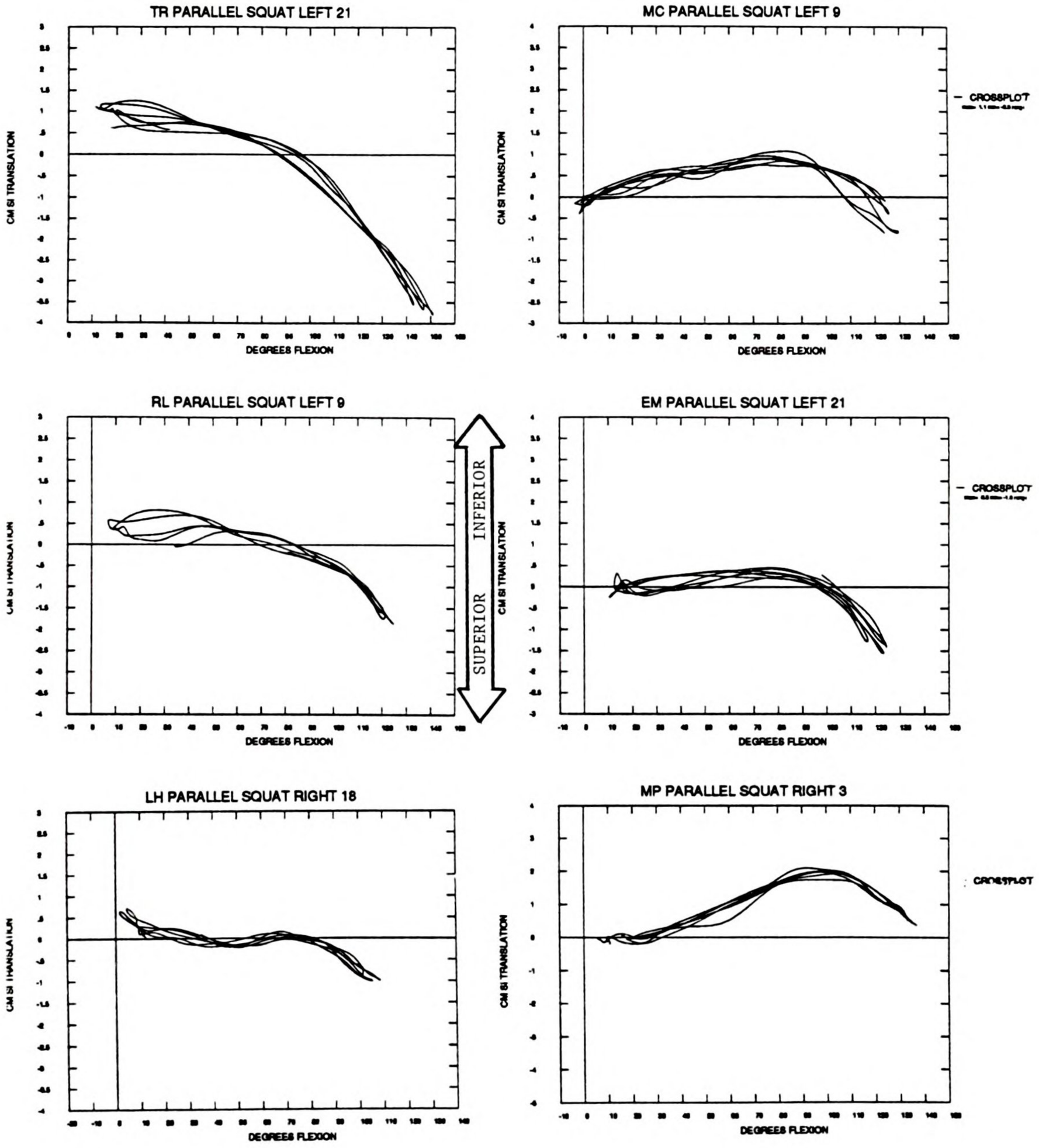


Figure 19. Crossplots of Flexion vs SI Translation (ACL Deficient)

Table 4. Mean Translations of Tibia Relative to the Femur (cm)

	AP	ML	SI
NORMAL	4.8 ± 1.25	1.5 ± .37	2.3 ± .50
ACL DEF	4.0 ± 1.22	1.3 ± .11	2.2 ± .35

Ground Reaction Forces and Moments

The mean peak vertical ground reaction force (GRF) was 81.8% bodyweight (BW) ± 6.6 SD in the normal knees and 79.4% BW ± 3.4 SD in the ACL deficient knees. The GRF patterns were the same for both groups (Figure 20). In both groups the peak vertical GRF consistently occurred at the deepest squat position (at the point of maximum flexion). The mean minimal vertical GRF was 27.2% BW ± 6.5 SD in the normal knees and 24.9% BW ± 6.2 SD in the ACL deficient knee, and consistently occurred between 20-40 degrees flexion. Homogeneous variance was present and two-sample t-tests revealed no significant differences between groups for either the mimimum or maximum GRFs. No pattern of weight bearing avoidance was evident on the ACL deficient knee compared to the normal knee. As expected the %BW was highly dependent upon the speed of the repitition. The slower speeds showed %BW as low as 72%, and the fastest repitition reached 102% BW. The mediolateral and anterioposterior GRFs were considerably smaller. During the squat,

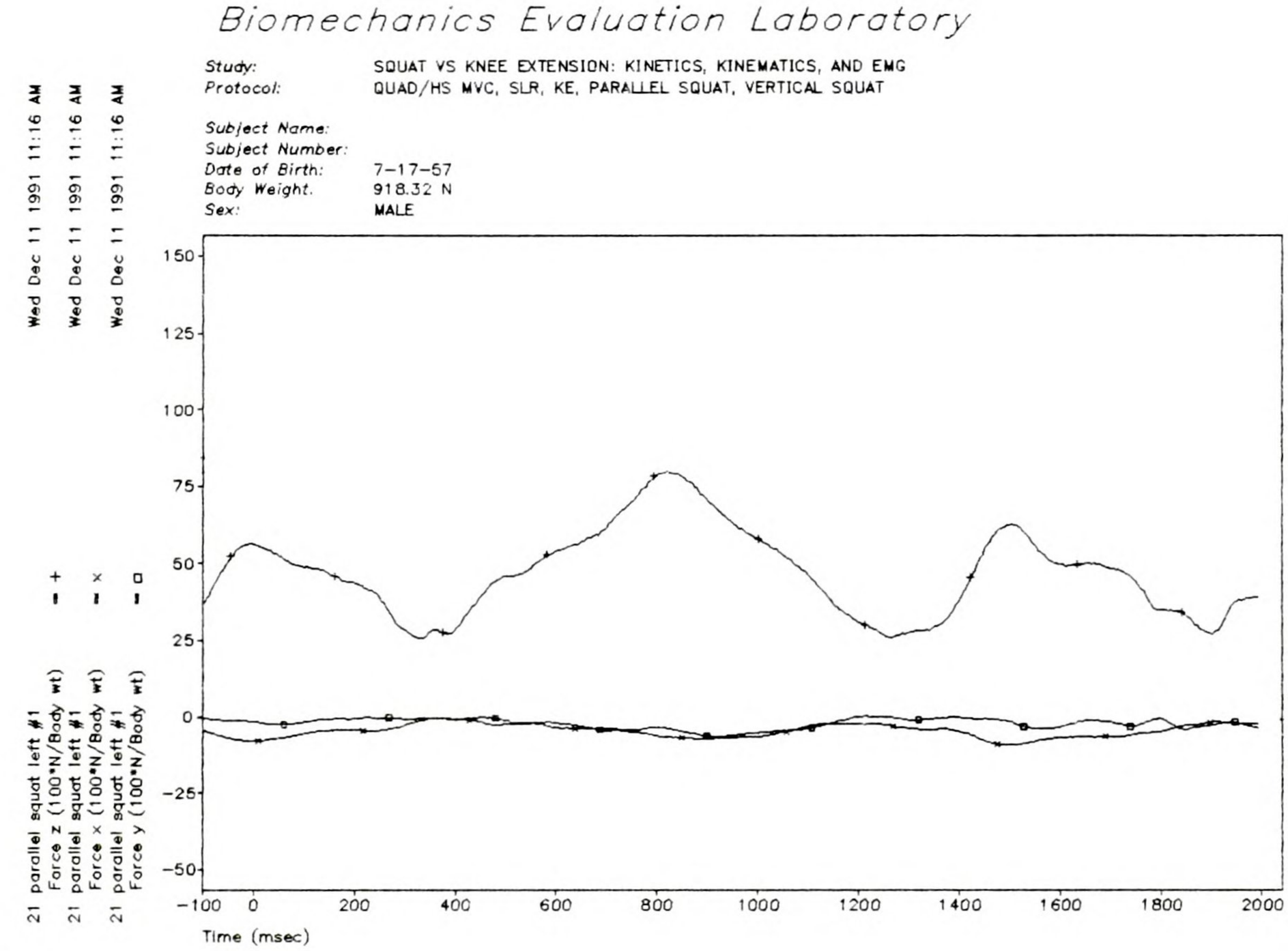


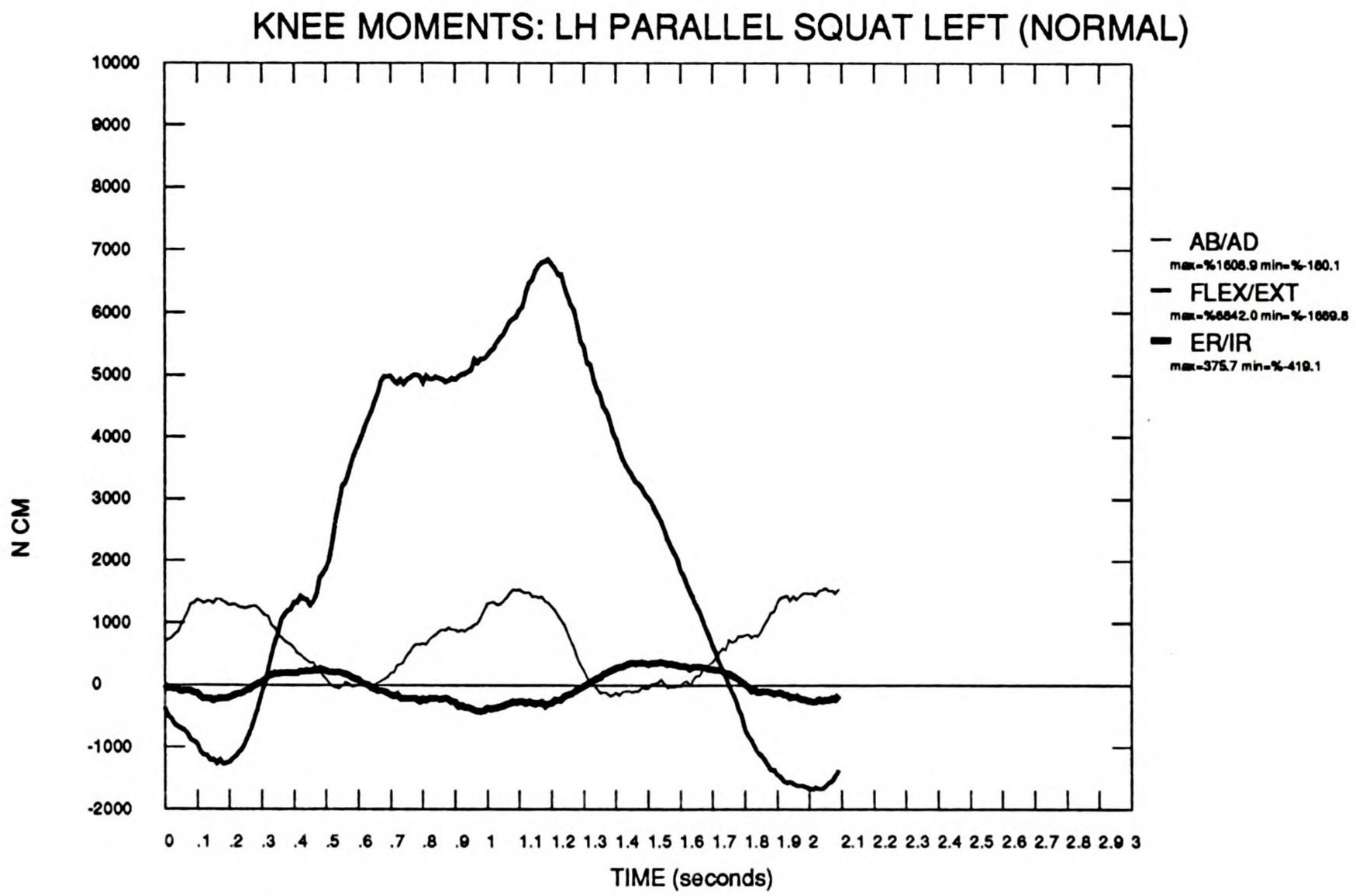
Figure 20. Typical Ground Reaction Force Pattern

all GRFs along the mediolateral axis were medially directed (indicating the subjects applied laterally directed force) and wavered between 0-14% BW. The AP GRFs were predominately posteriorly directed (anteriorly applied), but brief episodes of anteriorly directed GRFs were also observed. AP GRFs ranged from 8% BW posteriorly to 5% BW anteriorly.

As expected, all squats demonstrated dominant applied flexion moments throughout the range of movement, and the majority of subjects demonstrated small extension moments at the extended position (Figure 21). No pattern differences were observed between groups. In both groups, the peak flexion moments occurred just as the subject began the ascent from the maximally flexed position. Thus, the peak moment immediately followed the peak GRF. The mean normalized peak flexion and extension moments are presented in Table 5. Again, no statistically significant differences were found between the groups ($p>.30$). The abduction/adduction and external/internal rotation moments were comparably smaller than the flexion moments, and as such were considered insignificant.

Table 5. Mean Flexion/Extension Moments (%BW x HGT)

	Flexion	Extension
Normal	8.78 + .90 SD	.69 + .62 SD
ACL Def	8.14 + 1.37 SD	.79 + .56 SD



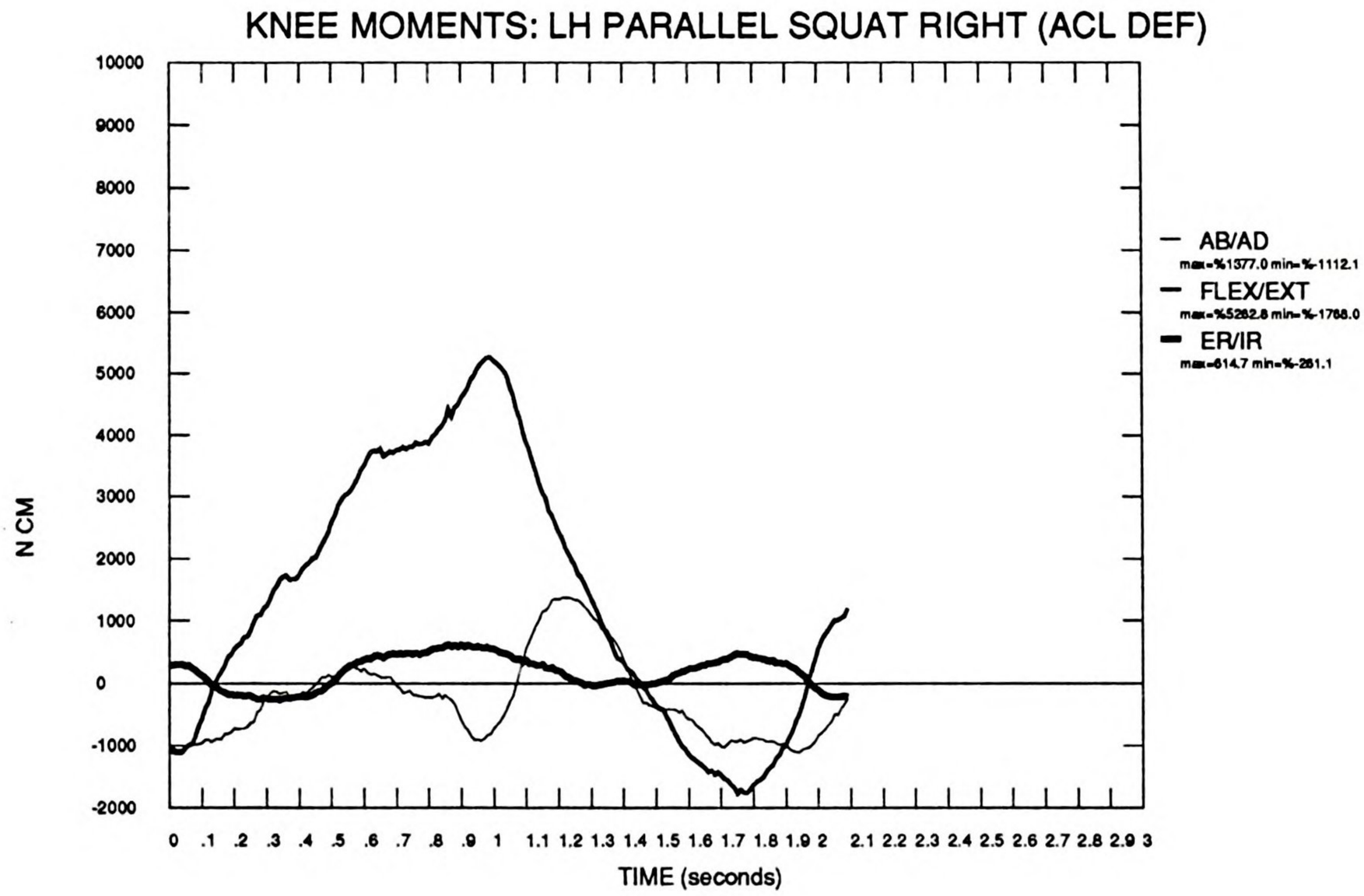


Figure 21. Typical Knee Moment Patterns

EMG

Muscle activity patterns were compared over the periods of squat cycles. Figure 22 displays a typical EMG graph of a normal knee. For all subjects, in both groups, the rectus femoris (quadriceps) demonstrated progressively increasing signal amplitude, reaching its highest amplitude in the maximally flexed position of the squat. The signal then tapered off during extension. The quadriceps appeared to be relatively baseline quiet in extension. Similarly, the trend was for both the medial and lateral hamstring muscle groups to display cocontraction patterns with the quadriceps. The gastrocnemius muscle patterns were often at baseline level or blocked by noise. In the trials where the signal was discernable, the trend was for increased amplitude in the flexed position.

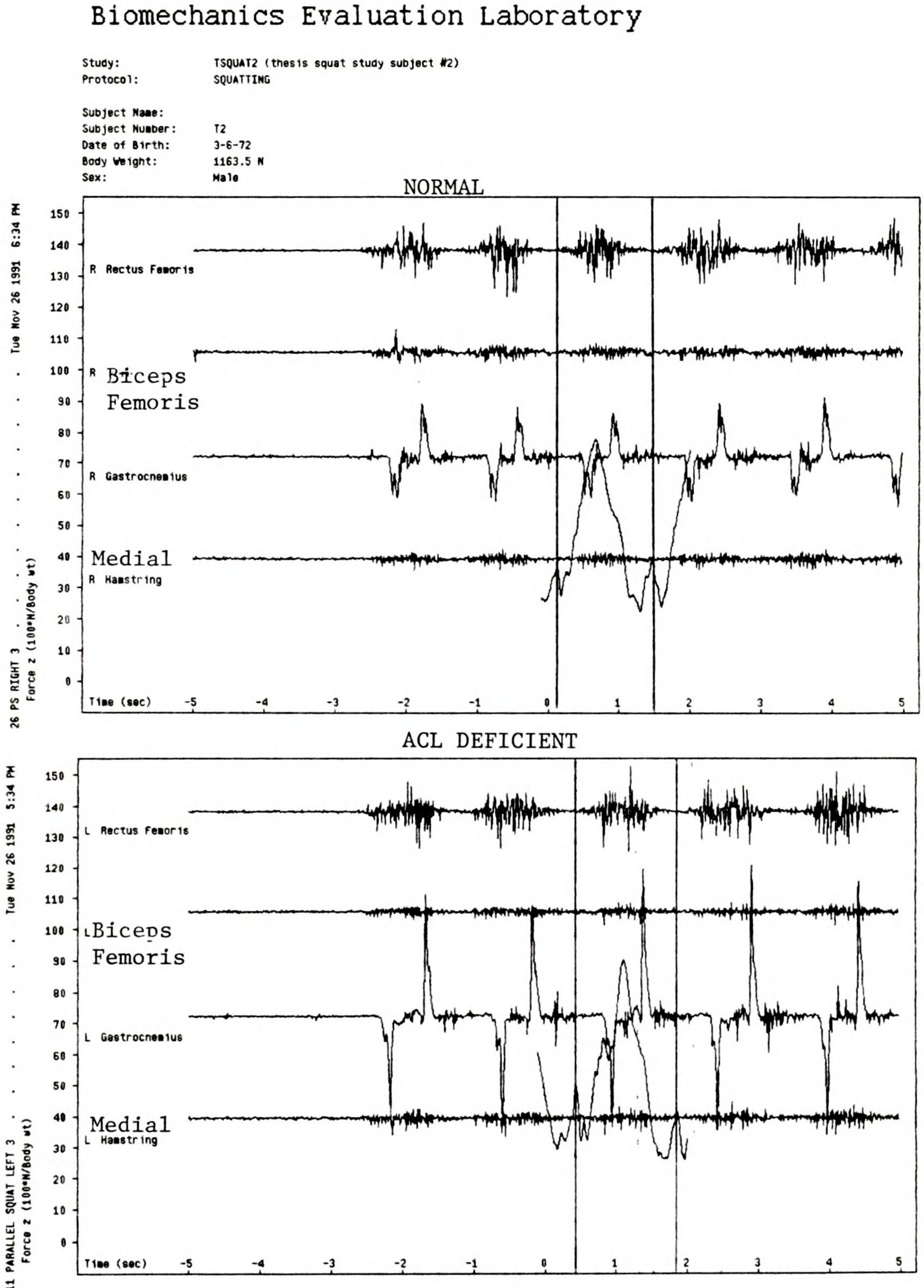


Figure 22. Typical EMG Patterns

VI. DISCUSSION

Recently, the squat exercise has been tauted as an effective and safe exercise following ACL repair or reconstruction.[80] Effective, because the quadricep muscles can be exercised and the deliterious effects of atrophy prevented. And safe because, theoretically, the ACL would be protected by the combined effects of hamstring muscle cocontractions and the increased stability gained from the vertical loading of the knee joint.

EMG recordings during the current study specifically demonstrate quadricep muscle activity during the squat (Figure 22). Since muscle activity is present during both flexion and extension of the knee, the squat provides opportunities for both eccentric and concentric contractions of the quadriceps. These forms of contractions would be desirable for most rehabilitation programs. Without additional processing of the EMG signals, the temporal quadricep firing patterns for both the normal and ACL deficient knees appear the same. An early decrease of quadricep activity has been reported in ACL deficient knees during isokinetic knee extension,[93] walking,[56] and cutting.[15] In this study, both groups appear to be baseline quiet as extension is reached.

Andriacchi[5] has proposed the use of moments as a sensitive indicator of muscle activity. In agreement with other studies,[8,64,89] the dominant moment observed was a flexion moment. In comparison

to the flexion moments during the squat, the abduction/adduction and rotational moments were small and no differences between groups were noted. The dominant flexion moments suggest the predominance of knee extensor muscle contractions required to balance the kinematic link system and prevent a loss of balance and collapse. Just as the peak quadricep muscle firing amplitude occurred around the point of maximum flexion, the peak flexion moment occurred at the earliest point of rising from this flexed position suggesting that a greater quadricep contraction is required at this point. Nissel[72] also reported the peak flexion moment at this point. The fact that the peak flexion moment immediately followed the peak ground reaction force may be attributed to the trunk shifting the center of mass posteriorly and increasing the flexion moment arm at the knee as the subject begins the ascent phase. Most subjects also demonstrated an extension moment when the knee was near extension. The presence of this moment is further supportive of the baseline quadricep muscle activity during this phase of the squat. An extension moment would be produced anytime the center of pressure (COP) lies anterior to the flexion/extension axis of rotation. Thus, if the subject shifted his weight anteriorly onto the forefoot when the knee was near extension, the COP would lie anterior to this axis. Plagenhoff[83] has described the change in moments produced secondary to trunk position which would essentially change the COP. In contrast to Andriacchi's[5,12] findings of adaptations made by ACL deficient knees during walking, no such patterns were seen during the squat exercise. These differences may be secondary to the nature of the activities evaluated, as knee

extension during gait is different than knee extension during a squat. Andriacchi also used limb segment masses and angular acceleration to determine his moments, however, angular acceleration would be higher during gait than during a squat. McLaughlin[64] also reported less than 10% error in using a static model over a dynamic model for determining moments.

In agreement with Carlsoo[16] and Draganich[25], this study also demonstrated the presence of hamstring muscle activity during the squat exercise and in cocontraction with the quadriceps. The key to this cocontraction is where it occurs. Referring to the raw EMG data (Figures 19 and 20), one can observe repeatable medial and lateral hamstring muscle activity concurrently with the quadricep muscle activity in maximal flexion. One difference between the muscle activities is the hamstring muscles appear to be active longer during the ascent stage of the squat. Since the hamstring muscles are biarticular, they will effect both the hip and knee joints. The muscle activity may reflect their role in either hip or knee extension or both. Although the hamstrings are knee flexors, they have been shown to act as knee extendors.[16] Since the hamstrings impart a posterior translatory force on the tibia when the knee is flexed[88] and the EMG data suggest hamstring muscle activity during flexion, it can be assumed that they would play a role in preventing anterior tibial translation when the knee is flexed. When the knee approaches extension, the line of action of the hamstring muscles become more parallel to the tibia, decreasing their mechanical ability to impart a posterior force on the tibia. In the presence of

knee hyperextension, the hamstrings may even produce an anterior translatory force on the tibia.[76]

Studies of open chain knee extension exercises on cadavers have reported that hamstring cocontractions can not prevent anterior tibial translation at less than 30 degrees flexion[88] or 10-30 degrees flexion.[24] A similar in vivo study[104] revealed the hamstrings could prevent anterior tibial shear forces at flexion angles greater than 17.2 degrees. These studies represented volitional isometric hamstring muscle contractions and can not be applied to dynamic knee exercises. During a Cybex isokinetic study where the knee actively flexed and extended from 0-90 degrees, Solomonow[93] reported the presence of tibial subluxation at approximately 45 degrees. Similarly, Yasuda[103] reported positive tibial anterior translatory forces at flexion angles less than a mean of 45.3 degrees. Andriacchi5 also reported that adaptations in the ACL deficient knees deminished as the knee was flexed in the range of 40 degrees. These studies suggest that the hamstrings are not capable of preventing anterior tibial translation during a dynamic open chain (and possibly closed chain) exercise at less than 45 degrees flexion.

Can the hamstrings in combination with axial loading, as in a closed chain squat exercise, prevent anterior tibial translation? Studies have demonstrated that knee stability increases with progressive increases in axial loading.[44,57,101] Although precise measurement could not be made, the current study suggests that there was not significant hamstring muscle activity as the knee approached extension in the ACL deficient knees. Even if it is accepted that the hamstring muscles in the ACL deficient knees are

contracting to prevent anterior tibial translation in extension, the question remains as to how effective they could be since the line of action on the tibia is the same during a knee extension exericise as it is during a squat. The primary difference lies in the vertical loading.

In agreement with other squat studies,[40,89] the vertical GRF was the dominant force and its peak occurred at the point of maximum flexion. Although the knee joint shear forces were not calculated, the AP shear forces would also be maximal at the point of maximum flexion because it is here that the tibial plateaus approach parallel alignment with the vertical GRF (Figure 23). It is also apparent that the shear force would be directed anteriorly on the femur and posteriorly on the tibia. Nissel[72] reported a considerable tibio-femoral shear force, with the femur tending to slide anteriorly on the tibia, in the flexed squat position. He also found that the shear force changed from negative (posterior drawer) to positive (anterior drawer) between .5 to .75 seconds into a 1.25 second ascent. This would probably place the knee in the range of40-45 degrees flexion.

During a squat repitition, knee flexion is coupled with tibial external rotation and adduction, and knee extension is coupled with tibial internal rotation and abduction. During approximately the final10-20 degrees of extension, the tibia again externally rotates indicating the presence of the screw home mechanism. The motion characteristics of this closed chain exercise coincide with that reported for an open chain knee extension exercise. Jonsson[47] studied knee extension from 30-0 degrees flexion. He reported

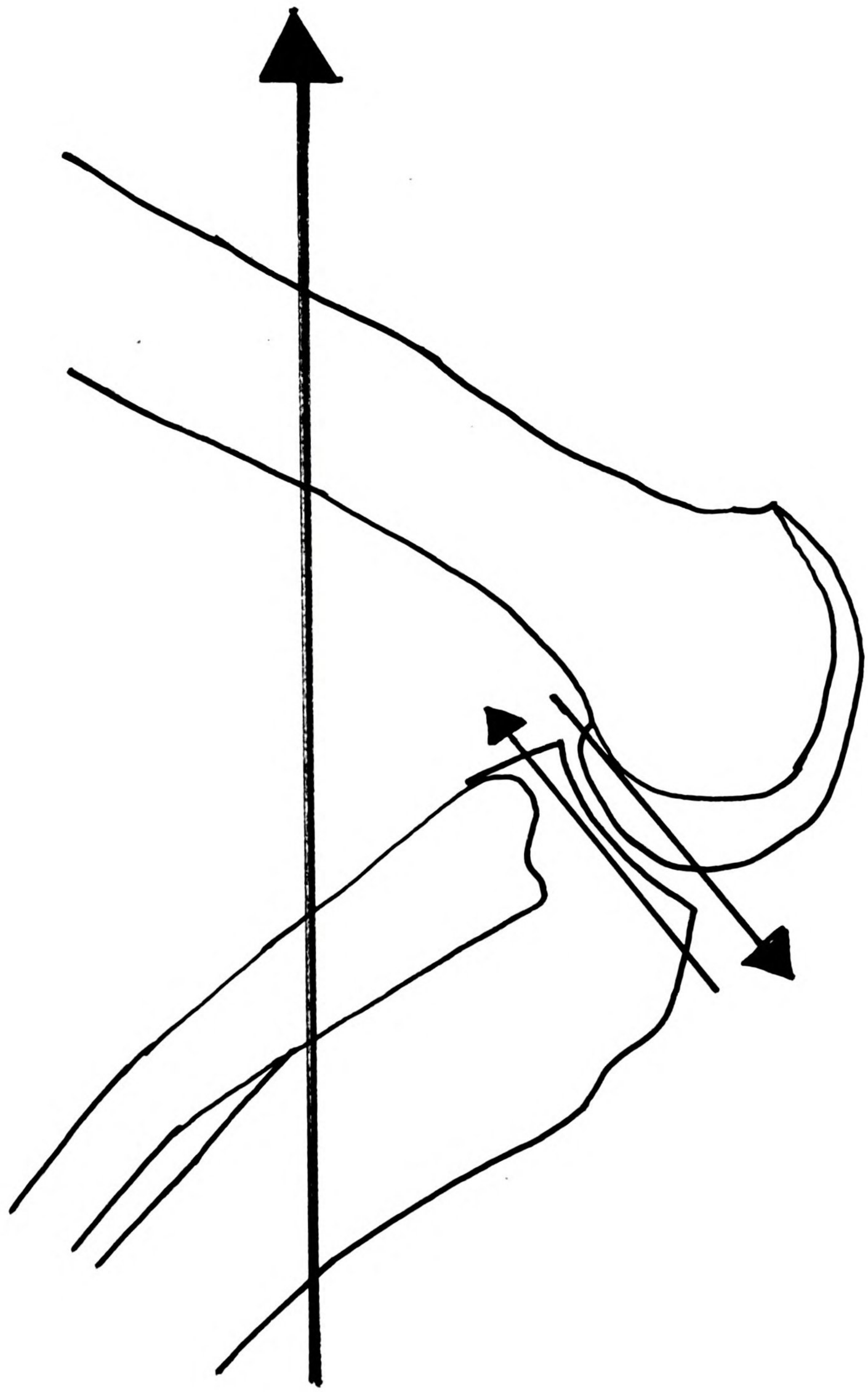


Figure 23. Vertical GRF Produces Posterior Tibial Shear as Tibial Plateau Approaches Vertical Position

tibial internal rotation which maximized at 15-20 degrees and external tibial rotation from 10-0 degrees flexion. He also reported adduction coupled with flexion and abduction with extension. While the ranges of flexion and rotation fall within the limits of motion reported at the knee,[13] abduction/adduction is well above. Since no other squat studies have described the rotational motions, comparisons are impossible. Three-dimensional studies of gait have reported varying amounts of abduction/adduction. Lafortune[54] reported a mean abduction of 6.4 degrees, while Morans[60] reported 3.9 degrees. The current study required a much larger range of flexion motion than required for normal gait. It might be argued that the shape of the femoral epicondyles could influence the varus/valgus angulation as flexion progresses in a loaded position, but targeting errors must be considered. Even if errors are present, relavent motions between groups can still be compared. In agreement with other studies,[21,47,60] this study found no statistically significant differences between the rotational motions of the normal and the ACL deficient knees. A larger sample may have been able to show a significant difference in external/internal rotation.

Regarding translational motions, the trend in the current study was for the tibia to translate posteriorly, medially, and inferiorly during extension and anteriorly, laterally, and superiorly during flexion. The closest activity found for comparison to these findings were those reported by Lafortune.[54] During the stance phase of gait, he reported that the tibia shifted anteriorly, medially, and into compression with extension and posteriorly, laterally, and into

distraction with flexion. When the knee extended during the swing phase of gait, he found that the tibia translated anteriorly, laterally, and into compression, and when it flexed, the tibia tranlated posteriorly, medially, and into distraction. Jonsson[47] studied open chain knee extension from 30-0 degrees flexion. He reported the same findings as Lafortune, except that the tibia translated inferiorly during extension. These differences suggest some confusion and the possibility of differences between open and closed chain exercises. An obvious difference between Lafortune's study and the current study is in the anterior and posterior translations. These differences, as discussed later, relate to the method used to describe translational motions. Although no statistically significant differences were found between the translational motions of the two groups, the mean translational motions in the ACL deficient knees were less than the normal knees. While no explanations are available for the mediolateral or superioinferior motions, the differences in the anterioposterior appear to contradict the findings in all the studies mentioned to this point. Again, these findings relate to the method utilized.

It has already been established that flexion and extension of the knee are accomplished by a combination of rolling and sliding. In this study, the relative AP translations between the femoral reference point and the tibia extended include the combined effects of the rolling and sliding that occurs arthrokinematically. In the case of a squat exercise, the femur primarily rolls and slides on the tibia. During flexion, the femoral condyles roll posteriorly on the tibial plateau, while they simutaneously slide anteriorly. The net

result is a posterior progression of the femur and a relative anterior displacement of the tibia in the sagittal plane (Figure 9). As the knee flexes, the ACL moves from a more vertical to a more horizontal position, following the path of the joint center. This positioning does not necessarily mean that more tension is placed upon a normal ACL in the flexed position. In fact, the shearing forces in a flexed position would tend to displace the femur anteriorly on the tibial plateaus (posterior drawer), with the majority of the strain on the PCL.[28] Also, when the knee is flexed, the hamstring muscles[76] and the iliotibial band[29,65,71] are in a position of mechanical advantage to prevent anterior tibial translations. The relatively constant slopes seen in the crossplots of flexion versus AP translation in the normal knees (Figure 14) suggest smooth progressions of the reference points anteriorly and posteriorly. In contrast, abrupt changes in the slope are seen in five of the six ACL deficient knees (Figure 15). This abrupt change in the slope of the progression of the reference points represents a deterioration of the rolling and sliding mechanism in this ACL deficient knee. The natural question at this point would be, "What caused this disruption?"

To answer this question, the examination will start at the fully flexed position and work toward extension. The crossplots (Figure 15) clearly demonstrate a constant line of anterior progression of the femoral reference point from the fully flexed position to the area of disruption. This suggests that the femoral condyles were rolling and sliding appropriately until the area of disruption was reached. When corrected for the targeting offset angle, it was

determined that progression disruption occurred at approximately 20 degrees. At this point, the femur is continuing to roll anteriorly, as indicated by the continued extension of the knee, but a simutaneous increase in the posterior sliding of the femoral condyles has also occurred, displacing the tibia anteriorly. During flexion, the mean angle where this disruption was corrected was 40 degrees. Graphs of time versus AP translation further demonstrate the abrupt disruptions in motion (Figure 13). What these graphs appear to demonstrate is the knee "giving out" or what has been described as a "pivot shift." The pivot shift is a phenomen commonly used as a clinical test to demonstrate the presence of ACL deficiency in the form of anterolateral rotary instability. Although different modifications of the test have been presented in the literature, the traditional test is performed on a supine subject by the evaluator applying internal tibial rotational and valgus stresses to the knee while simutaneously flexing the knee slowly from 0-30 or 40 degrees. In a positive test, an instantaneous "jump" of the lateral tibial plateau is witnessed as the knee passes through the 30-40 degree flexion range. The proposed mechanism for the pivot shift test is as follows:[29,65,71] As the knee is held in extension, the tibia is already anteriorly displaced or subluxed. By internally rotating and abducting the tibia, additional anterior subluxation of the lateral tibial plateau occurs. In the extended position, the iliotibial band (ITB) lies anterior to the lateral femoral epicondyle and joint center. Thus, in extension, the ITB acts a knee extensor. As the knee is flexed, tension is increased in the ITB and it eventually passes over the lateral epicondyle (30-40 degrees

flexion) and posterior to the joint center. At this point the ITB becomes a flexor and instantaneously produces a posterior force on the anteriorly displaced tibial plateau, reducing it to its normal position. The reduction is the "jump" which is witnessed. It should be emphysized that the presented mechanism of the pivot shift applies to the clinical, non-weightbearing laxity test. The pivot shift has been described during knee extension exercises.[33,93] It has also been proposed that this phenomenom occurs during foot planting and cutting activities[75] as evidenced by the subjective complaint of the knee "giving out," but to this author's knowledge, no objective record of a pivot shift during a functional activity has been reported in the literature.

One subject (TR) did not demonstrate a functional pivot shift during any of the trials; however, differences between the two knees can be identified. One might summise that the lack of "tightness" in the ACL deficient knee is still indicative of some amount of functional laxity. The fact that this subject did not demonstrate a pivot shift may suggest that this ACL deficient knee is more functionally stable than the other ACL deficient knees tested. On the other hand, the subjects in this study did not necessarily have isolated ACL pathology, and certainly all ACL deficient knees are not the same.[9] However, the pattern of changes in the AP translations seen in the presented graphs are different between groups and reflect some of the underlying patho-biomechanics of the tested knees. The role of associated pathology as a contributory factor to knee stability is yet to be determined.

Further research is needed to compare motion patterns on subjects with arthroscopically confirmed pathologies with normal knees

VII. CLINICAL IMPLICATIONS AND CONCLUSIONS

The documentation of the relative translational motions between the femoral and tibial reference points and the observed disruption of the rolling and sliding at the knee (functional pivot shift) during a squat have at least two important clinical implications. The first relates directly to the squat exercise, while the second relates to possibilities for objective clinical testing. First, the squat implications will be discussed.

Several cadaveric[8,39,88] and in-vivo[43,47,103] studies have demonstrated that anterior tibial translation occurs with open chain knee extension exercises. Other studies have reported anterior tibial translation during such functional activities as walking.[54,60] Partly because of the assumptions that vertically loading the knee would increase its stability[44,57,61,101] and that antagonistic muscle contractions occur while squatting,[16,25] some have assumed that squats would be a safe and effective means of exercising an ACL deficient or post-operative ACL repair/reconstruction knee.[80] By arthroscopically strain-gage implimenting the ACL in a single subject, Henning[41] documented the strain within the ACL while the subject performed a single leg half squat. Based upon his results he recommended that squats be performed within 20-90 degrees flexion. Some exercise protocols have implemented his recommendations.[84,92] The current study, though on a small sample

size, suggests that the hamstring muscles are not capable of preventing anterior tibial translation as the knee approaches extension or during early flexion during a squat performed at an approximate two-second rate. Caution is advised when a subject with an ACL deficiency performs a squat at less than 40 degrees flexion. The anterior tibial subluxation or pivot shift phenomenon that occurred could potentially damage a recently repaired/reconstructed ligament and possibly contribute to the frequently observed deterioration of the joint. Erosive changes have been attributed to the hyperexcursions and shearing with each episode of "giving way."[29,75]

On the other hand, other factors may have influenced the results of this study. This study demonstrated, in agreement with Ariel,[7] that faster squats produce wider ranges in the vertical forces. It also was observed that the pivot shift occurred in a range where these vertical compressive forces were relatively low. The speed of the squat may have affected the results. A slower squat would produce a relatively constant vertical ground reaction force that might have provided added stability to the lax joint. Solomonow[93] demonstrated that the subluxation could be prevented during slow isokinetic exercise on a select subject sample. Further studies should be performed to determine the effect of the rate of exercise on the presented phenomenon.

Ohkoshi[78,79] has reported that forward bending at the trunk while standing with various degrees of knee flexion increased the amplitude of EMG activity in the hamstring muscles. Although hinting that these muscle contractions would increase the posterior

shear forces at the knee, his data revealed that only at knee flexion angles greater than 30 degrees did a posterior shear force occur in all normal subjects. Since the hamstring muscles are biarticular, one would expect increased activity with trunk flexion. And, because they lie essentially parallel to the femur they would have minimal mechanical advantage in attempting to reduce an anteriorly displaced tibial plateau.[76] Although the current study demonstrated antagonistic muscle activity, it occurred primarily in the deep squat position, and certainly, in the majority of subjects did not prevent the anterior tibial displacement when approaching extension. Forward bending the trunk does not change the angle of pull on the tibia, although it does increase hamstring tension. For the same reasons, volitional cocontractions of the hamstrings while performing the squat, as suggested by Antich,[6] may also be questionable in preventing strain on the ACL. Again, studies should be performed to confirm or dispute these theories.

This study also has some important implications for clinical testing of outcomes. Gauffin[32] has described three types of laxity:

1) Static laxity has been described in this thesis as passive laxity. This is the most frequently used measure of post-operative outcomes.[84]

2) Dynamic laxity is the intrinsic translational movements that occur in the joint during functional motions. This motion is what has been documented in this thesis inspite of claims that it could not be accomplished.[19,32]

3) Functional instability is the subjective symptoms presented by the patient, such as a feeling of the knee giving way. Some have

attempted to assess functional outcomes by pacing the patient through a variety of functional activities[10,31] or by simply questioning them on their symptomotology.[14]

While each of these has merit, a method of documenting dynamic or functional stability would be the most objective and meaningful. Clearly, there are limitations to any system which requires the placement of external devices to measure internal movement of boney segments. Skin motion and joint center approximation will limit the ultimate resolution of any measurements taken. The best data would come from a motion analysis method which uses rigid bone fixation. Since this is not practical for the general patient population, we owe it to the patients, the surgeons who refer the patients for evaluation, and to the field of biomechanics to develop the best non-invasive methods for obtaining the desired information. The method described in this study for recording translations at the knee during a squat is a step in the desired direction. This study has shown that small translational motions can be recorded using an exoskeletal method, and one which is less cumbersome than an electrogoniometer. However, further studies are needed to compare findings with a variety of activities and to refine the techniques utilized. These might include comparing various marker attachment methods with a bone fixation method. On a research basis, bone fixation techniques could be justified. Once the best methods have been developed, then valuable objective information could be provided in regard to pre- and post-surgical and rehabilitation outcomes. Dynamic laxity could be documented while the patient is run through a battery of test situations including walking, running,

cutting, squatting, and a multitude of sport/activity specific maneuvers.

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