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GAIT CHARACTERISTICS EXHIBITED BY SUBJECTS
WITH UNINJURED AND INJURED KNEES
UNDER UNFATIGUED AND FATIGUED CONDITIONS

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By

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ABSTRACT

GAIT CHARACTERISTICS EXHIBITED BY SUBJECTS WITH UNINJURED AND INJURED KNEES UNDER UNFATIGUED AND FATIGUED CONDITIONS

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The purpose of the study was to perform a three dimensional kinematic analysis to examine the effects of muscle fatigue on gait characteristics of individuals with ACL reconstruction, compared to uninjured individuals. Subjects were tested under unfatigued and fatigued conditions. Fatigue was produced utilizing a submaximal bicycle ergometer test. In addition, electromyographic patterns of quadriceps and hamstrings firings were analyzed.

The ACL group exhibited greater muscular activity when unfatigued and fluctuated between lower and higher magnitudes of measured linear and angular parameters when fatigued, than did the control group. Under fatigue, changes in gait characteristics were found for all subjects in the study, with higher magnitudes in angular parameters, when compared to the unfatigued condition.

To my parents, Jorge and Mirna Angeli for believing in my dreams, and for all the sacrifices and constant support.

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CHAPTER I

INTRODUCTION

Interest in healthy lifestyles has caused an increase in the number of physically active individuals in the United States. Along with a greater level of activity, there has been an increase in the number of injuries reported. The leading cause of sports injuries is overuse (Pecina and Bojainc, 1993). Statistics show that soccer, tennis and football are the sports in which the majority of all the knee injuries occur (Nicholas and Hershman, 1995). Within the broad category of knee injuries, the anterior cruciate ligament (ACL) and the medial colateral ligamnet (MCL) are the most commonly injured tissue of the joint, responsible for over 50% of the injuries reported. Because of the anatomy of the knee, an injury to the anterior cruciate ligament is often accompanied by an injury to other soft tissues of the joint. Most injuries to the knee ligaments occur due to some type of contact or force which produces an abnormal movement at the joint. Non contact injuries to the ACL are produced by abnormal loading of the knee joint during landing in an unbalanced position (Nicholas et al., 1995).

While the cruciate ligaments guide and restrict the tibia and femur to a gliding and sliding motion, the anterior cruciate ligament is the primary stabilizer of the knee joint (Jackson and Drez, 1987). An injury to the anterior cruciate ligament negates normal function, and jeopardizes joint stability. Ligament sprains are classified into three categories: first degree, in which the ligament fibers are injured at the microscopic level, and no stability is lost in the joint; second degree, a partial tear of the ligament fibers occurs, and some joint stability is compromised; and third degree, in which the ligament fibers are completely torn, accompanied by a complete loss of stability (Macnicol, 1986). Several factors will determine the treatment following the injury: severity of the tear, age of the patient, level of activity of the patient, and desire to continue an active lifestyle. Reconstruction of the ACL is usually recommended for second and third degree tears, and for individuals who plan to continue a high level of activity. Rehabilitation without surgical intervention is often sought by individuals who do not plan to engage in high levels of activity, or have experienced a first degree sprain to the ACL, without any damage to other knee structures (Jackson and Drez, 1987).

Further consideration should be given to the issue of ACL reconstruction versus rehabilitation. The consequences of an injury to the ACL include loss of knee stability as well as modifications to the knee mechanics. Considering these factors, ACL reconstruction should focus on restoring knee stability, maintaining the normal mechanics of the knee joint, and preventing the premature degeneration of the joint. The surgical procedure for reconstruction is beyond the scope of this paper. However, it is important to understand that replacement for the anterior cruciate ligament must exhibit similar mechanical characteristics as the original ACL in order to allow for the same joint kinematics.

Chronic ACL deficiency may result from untreated injuries to the ACL. If the patient decides only to rehabilitate the knee after the injury, chronic knee laxity can become a factor. Jackson and Drez (1987), identified most chronic ACL deficient patients as having laxity greater than 7.5 millimeters in the injured knee and a difference greater than 2 millimeters between both lower limbs. The authors also reported that repairs of acute ACL injuries surgically treated soon after the incident had a 6% better outcome than repairs performed on chronically deficient knees. As a result of frequent subluxation of the joint in the ACL deficient knee, other tissues may become damaged; if surgery is desired at this point, the results may not be as successful since degeneration of the tissue has already began.

The knee kinematics are compromised as a consequence of an injury to the anterior cruciate ligament. In addition, muscle action is altered to compensate for abnormal movements (Shiavi, Zhang, Limbird, and Edmondstone, 1992). The topic of injuries to the anterior cruciate ligament has been of interest to several investigators Andriacchi (1990 and 1993); Timoney, Inman, Quesada, Sharkey, Barack, Skinner and Alexander (1993); Berchuck, Andriacchi, Bach, and Reider (1990), who all used gait analysis to examine the adaptations caused by an ACL injury. The general results of these studies showed differences in angular displacement at the knee joint. Other research has focused on change in muscle action due to the injury (Shiavi, Zhang, Limbird, and Edmondstone, 1992; Schipplein and Andriacchi, 1991; Solomonow, Baratta, Zhou, Shoji, Bose, Beck, and D'Ambrosia, 1987; Hirokawa, Solomonow, Lu, Lou, and

D'Ambrosia, 1992; Renstrom, Arms, Stanwyck, Johnson, and Pope, 1986; and O'Connor, 1993).

Even though the topic of anterior cruciate ligament injuries has been very popular in the past two decades, little, if any, importance has been given to the issue of fatigue in relation to an ACL injury. Muscle fatigue can play an important role in the reoccurrence of any type of injury. In the case of an ACL injury, muscle involvement becomes a critical factor in the early stages of the rehabilitation process. When a second degree injury occurs, the anterior cruciate ligament has an 86% reduction in its capacity to restrain joint movements (Jackson and Drez, 1987). With a reduction in knee stability produced by the injured ACL, the muscles assume a more important role in knee function (McNair and Wood, 1993).

I. NEED FOR THE STUDY

Research in the area of muscular fatigue as it relates to gait characteristics is rare. Studies in which the effects of fatigue on the function of muscles as primary stabilizers of the knee after an injury to the anterior cruciate ligament are scarce. When the anterior cruciate ligament is injured, knee stability is jeopardized. In combination with the kinematic adaptations present at the knee joint, muscles of the lower extremity become joint stabilizers, restricting excessive movement of the bones (Jackson and Drez, 1987). After a period of approximately six to nine months, the individual is usually ready to return to preinjury activity level. However, gait analysis studies have shown that adaptations in the patients' walking characteristics are still present after one year of the injury (Andriacchi, 1990; Andriacchi, 1993; Berchuck et al., 1990).

Return to physical activity prior to full restoration of the mechanical properties of the injured ligament can increase the chances of reinjury. Noyes, Torvik, Hyde, and DeLucas, (1976) reported a 39% decline in the maximum failure load of the ligament after an eight week period of immobilization. There were no additional benefits of isometric exercises during the immobilization period. After 20 weeks of rehabilitation exercises, the ligament showed almost normal stiffness properties, but strength properties were incomplete. Athletes and coaches must be sensitive to the results obtained by Noyes et al. (1976), as these results have implications for preventing ligament reinjury and further damage to the joint itself.

There has been little research performed on the effects of fatigue on reinjury of the anterior cruciate ligament. Injury to the anterior cruciate ligament is one of the most common injuries to the lower extremity in competitive sports. Muscle fatigue becomes a concern due to the role of the muscles of the lower extremity as primary stabilizers of the joint. The athletes' training styles do not typically change after returning from a rehabilitation period (S. Nogle, personal communication, March 21, 1996). Long hours of intense workout have always been a characteristic of training, and have not been thought as a situation in which the athlete is at risk of injury.

The lack of studies in which fatigue is considered as a variable to the kinematic analysis of an injury related situation, can be due to the controversy of the topic. Fatigue has been a very poorly understood topic, due to its complexity in physiological characteristics, as well as the lack of understanding of the psychological effects at its onset. Most of the research done on fatigue is to gain a better understanding from a physiological point of view. Research conducted on the effects of fatigue on the kinematic characteristics of uninjured and injured individuals can assist in the development of safer training practices for injured individuals. This study will focus on the kinematic characteristics of individuals with an anterior cruciate ligament reconstruction not older than 15 months, under fatigued and unfatigued states of the muscles of the lower extremity. The same kinematic characteristics and the adaptations due to fatigue will be analyzed in uninjured individuals.

II. PURPOSE OF THE STUDY

Kinematic adaptations during gait of individuals with ACL deficiency or recent reconstruction have been well documented (Andriacchi, 1990 and 1993; Timoney et al., 1993; Tibone, Antich, Fanton, Moynes and Perry, 1986; and Berchuck et al., 1990). Functional changes in the knee joint as a result of the injury to the ACL, are exhibited in gait analysis data as well as electromyographical (EMG) data. The purpose of this study was to perform a three dimensional kinematic analysis utilizing EMG in order to examine the effects of muscle fatigue on the gait characteristics of individuals with a reconstructed anterior cruciate ligament, and compare the changes to those experienced by uninjured individuals. The following hypotheses were formed on the basis of previous research:

Hypothesis 1: individuals with ACL reconstruction will exhibit greater muscular activity during the stance phase under normal conditions than the control subjects.

Hypothesis 2: under the fatigued state, individuals with ACL reconstruction will have lower angular and linear displacements, velocities and accelerations occurring at the knee joint, than under normal conditions.

Hypothesis 3: subjects with ACL reconstruction will experience adaptations of greater magnitudes occurring at the knee joint under the fatigued state than the control group.

III. ASSUMPTIONS AND LIMITATIONS

It will be assumed that each individual reaches muscle fatigue prior to each trial under the fatigued condition, and the fatigued state is maintained throughout the trial. Fatigue will be assumed to be present when the subject cannot maintain the predetermined mechanical output on the bicycle ergometer, the EMG pattern decreases in frequency and in amplitude, and the subject reaches 8 on the Borg's 10 point perceived exertion scale.

Due to the short duration, high intensity characteristic of the fatiguing protocol, muscle fatigue will occur due to depletion of energy sources in the muscle. Different factors may contribute to the onset of fatigue after long duration, moderate intensity exercise. For this study, it will be assumed that the type of fatigue does not affect the adaptations in gait characteristics.

IV. SIGNIFICANCE OF THE STUDY

The findings of this study will contribute to the field of injury prevention. The study will provide information for coaches, trainers, and athletes on how the knee joint specifically adapts to the factors of muscle fatigue after an injury to the ACL. Understanding the changes in kinematic characteristics during gait as a consequence of fatigue of the muscles of the lower extremity, in an individual with recent ACL reconstruction, can help in the development of safer training practices for the injured athlete. Understanding the effects of muscular fatigue in a healthy individual will also allow for the planning of safer practices for prevention of injuries.

V. DEFINITIONS

<u>Anterior cruciate ligament deficiency</u>: a condition experienced after a partial or complete tear of the anterior cruciate ligament. The knee joint demonstrates instability during clinical examinations, Lachman and anterior drawer tests. Normal knee mechanics may only be reinstated through reconstruction.

<u>Anterior cruciate reconstruction</u>: replacement of the injured anterior cruciate ligament through surgical procedure, allowing a close reproduction of normal knee mechanics.

<u>Cocontraction</u>: simultaneous contraction of two or more muscle groups.

<u>Concentric contraction</u>: shortening of muscle fibers when tension is produced.

<u>Eccentric contraction</u>: lengthening of muscle fibers when tension is produced.

<u>Electromyography</u>: (EMG) method use for measuring electrical signals during muscle activity. Surface electrodes will be used as sensing devices.

Fatigue: inability of the muscle to maintain a constant mechanical output at approximately 75% of maximum workload.

Gait cycle: cycle which refers to the period between heelstrike of one limb to heelstrike of the same limb. Each cycle is composed of the following events (see Figure 1):

Heelstrike (HS): beginning of the stance phase, when the foot first makes contact with the ground.

Midstance: point at which deceleration changes to acceleration. The position of the shank is at 90 degrees to the surface.

Toe off (TO): end of the stance phase, when the foot leaves the contact surface as it enters the swing phase.

Stance phase: period during the gait cycle in which the limb is in contact with the ground. This period starts with heelstrike of either foot and ends with toe off of the same foot. The stance phase can be divided into two events:

Single support phase: period during the gait cycle in which one limb is in contact with the ground and the opposite limb is in the swing phase.

Double support phase: period during the gait cycle in which both limbs are in contact with the ground.

Swing phase: period during the gait cycle in which the limb is advancing free of contact with the ground. This period starts at toe off and ends at heelstrike.

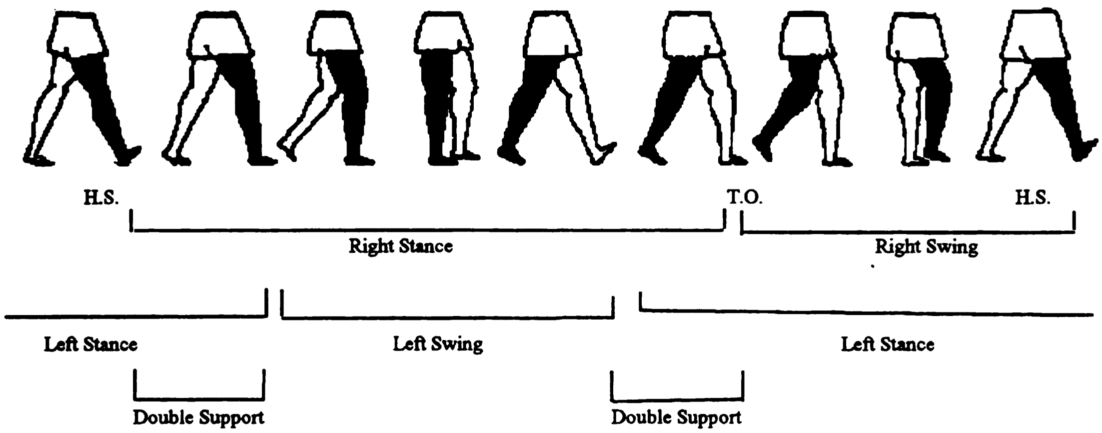


Figure 1: Gait Cycle Components

Knee stability: restriction of excessive joint motion. The values of laxity tests are within the normal range.

Ligament tear: injury to the ligamentous tissue. The following classification specifies the severity of the injury:

First degree tear: injury to the ligament fibers at the microscopic level. No joint instability is detected during clinical examinations.

Second degree tear: partial injury to the ligament fibers. Increased instability may be detected during clinical examinations.

Third degree tear: complete tear of the ligament fibers. All joint stability is lost.

Normal gait: walking pattern kinematics within normal ranges and demonstrating no characteristics and signs of injury.

Stride length: distance between heel contact of one foot to heel contact of the same foot after a complete cycle. The distance is measured in the sagittal plane.

<u>Valgus deformity</u>: medial opening at the knee joint.

<u>Varus deformity</u>: lateral opening at the knee joint.

CHAPTER II

REVIEW OF LITERATURE

Walking is one of the most common locomotive activities performed by humans. Analysis of the components of gait of the lower extremity can assist in the identification of pathological conditions. A kinematic gait analysis provides information about linear and angular displacements, velocities, and accelerations, which can be used in the detection of variations in walking. Kinetic data can complement the kinematic analysis by providing information about forces, moments, and power. Aside from individual differences in the gait data, adaptations made because of a pathological condition can be identified by the discrepancies in the kinematic values when compared to range values of data from non pathological subjects. From the digitization of markers placed over the specific joint, a biomechanical model can be developed. Angular and linear calculations can be made from the data collected, with body position being determined first, followed in order by velocity and acceleration calculations. The kinematic data can be combined with data from force plates, electromyographs, and electrogoniometers, which enrich the analysis (Harris and Wertsch, 1994).

Patterns of walking have not been found to be mechanically predetermined, suggesting that the step length and frequency during walking can be changed if desired.

As these changes are made, the walking pattern must always meet the prerequisites of equilibrium, body stability, and strength to maintain a coordinated movement. Another characteristic of normal gait is the symmetry of movement with right and left steps being approximately the same length. After the age of four, the actions involved in walking start to become automatic and the movements are controlled by the central nervous system. The central nervous system produces a basic locomotion pattern which is intended to propel the body in a specific direction, controls the balance of the body and adapts the gait pattern to different environmental demands. Changes in the environment as well as changes within the locomotive system result in adaptations to the gait pattern. Adaptations of any of the components of the gait pattern are made to maintain both body balance and gait mechanics close to the normal range for that individual (Zatsiorky, Werner, and Kaimin, 1994).

I. KNEE ANATOMY

A brief review of the structure of the knee joint will be provided to facilitate the understanding of the studies reviewed in this chapter. The femur and the tibia articulate in a hinge joint, which has as part of the supportive structure, dense connective tissue called the cruciate ligaments. The ACL is attached to the posterior aspect of the medial surface of the lateral femoral epicondyle. At the tibia, it attaches to a fossa, anterior and lateral to the anterior tibial spine. As it attaches to the tibia, the ACL passes under the transverse meniscal ligament, and a few fascicles of the ligament may blend with the anterior part of the lateral meniscus. This blending of tissues is often the reason for injury to the ACL. The posterior cruciate ligament (PCL) is attached to the posterior aspect of the lateral surface of the medial femoral epicondyle. At the tibia, the PCL attaches to the depression found posteriorly on the articular surface of the tibia. The cruciate ligaments cross each other anteriorly and posteriorly, as they extend from the femur to the tibia. This orientation in space is critical to the function of the cruciate ligaments as they serve to constrain anterior/posterior joint motion. The anterior cruciate ligament extends anteriorly, medially, and distally as it crosses from the femur to the tibia. The ACL also turns into a small outward spiral, due to the attachment sites. The posterior cruciate ligament, extends posteriorly, laterally, and distally through the joint, as shown in Figures 2a, 2b and 2c. The orientation of the femoral attachments is responsible for the ligament's tension throughout the range of motion. Both ligaments act to limit extension, and prevent hyperextension at the knee joint. If the knee reaches a hyperextended position, the ligament that is most exposed to an injury is the anterior cruciate (Feagin, 1988). The menisci rest on the head of the tibia, and articulate with the epicondyles of the femur, as shown in Figure 2c. Each cartilage is attached to the inside aspect of the capsule in the knee. Each meniscus covers the outer two-thirds of the articulating surface of the tibia. The medial meniscus is semicircular; its anterior aspect is attached to the depression on the anterior margin of the head of the tibia. Its posterior aspect is attached to the depression behind the spine of the tibia. The lateral meniscus is almost a full circle, and covers a larger surface than the medial meniscus. The anterior portion of the lateral meniscus is attached anteriorly to the spine of the tibia, and the posterior aspect of the lateral meniscus is attached behind the spine (Gray, 1974).

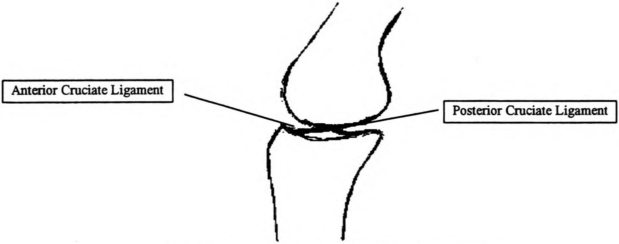


Fig. 2a: Right Knee - Medial view of Sagittal View

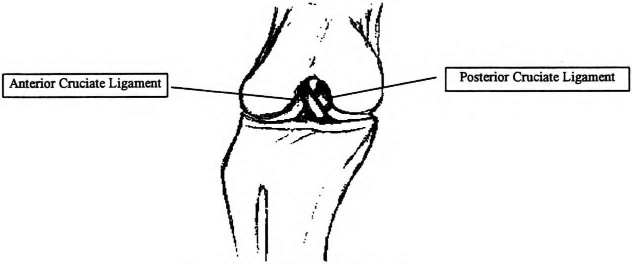


Fig. 2b: Right Knee - Anterior View

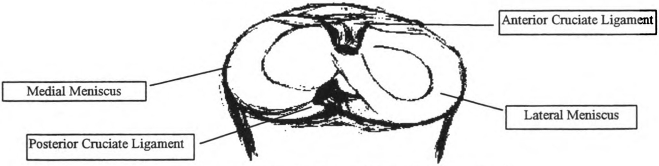


Fig. 3c: Right Knee - Transverse View of Tibial Plateau

Figure 2: Anatomy of the Right Knee Joint

One of the principal issues that needs to be addressed when analyzing gait of individuals with ACL deficiency, is the functional level of the knee joint. The role of the anterior cruciate ligament is to stabilize the knee joint. The ACL is the main ligament responsible for the prevention of anterior displacement of the tibia over the femur, and it also aids in restricting hyperextension. As a result of an injury to this major stabilizer, the knee joint is subject to rotational and shear moments which can further jeopardize the joint's functional capabilities (Shiavi, Zhang, Limbird, and Edmondstone, 1992). An injury to soft tissue in the joint results in mechanical adaptations to compensate for functional losses (Peltier, 1985).

An injury to the anterior cruciate ligament produces adaptations in the function of the knee joint. To be able to isolate these adaptations, the normal behavior of the ligament must be observed under different conditions. The following measurements were obtained during a study done on cadavers, with the purpose to measure strain in the anterior cruciate ligament during different positions of the knee joint. The tool used for this purpose was a Hall effect transducer. Full extension was defined as a flexion angle of 0 degrees. The minimum strain of the anterior cruciate ligament during a normal flexion/extension pattern was found at 30 to 35 degrees; under a passive motion, the strain level decreased. Flexion greater than 35 up to 120 degrees, increased the strain level. At about 15 to 45 degrees of flexion, the anterior cruciate ligament showed its greatest strain during a passive varus moment, defined as a rotation towards the midline of the body. The passive valgus moment, causing a rotation of the tibia away from the midline of the body, also produced increased levels of anterior cruciate ligament strain, except during full knee extension. When passive internal and external rotations were compared for the anterior cruciate ligament strain increase, the internal rotation was more significant, maximizing at 10 to 15 degrees of flexion and showing its minimum effect at 60 to 65 degrees. When an isometric quadriceps contraction was simulated, the anterior cruciate ligament strain increased significantly over the first 45 degrees of flexion and the strain decreased when the angle of flexion was more than 60 degrees. An eccentric contraction of the quadriceps also showed a significant strain increase for the first 45 degrees of flexion, and appeared to be similar to the normal pattern when the angle was greater than 70 degrees. When movement of external rotation combined with an eccentric contraction which lowered the leg, decreased levels of anterior cruciate ligament strain were demonstrated. The measurements show that the ACL tightens when the knee is in maximum flexion to approximately 35 degrees of flexion, and relaxes from 35 degrees of flexion to full extension (Arms, Pope, Johnson, Fischer, Arvidsson, and Eriksson, 1984).

In addition to the information on knee structure, the fundamental characteristics of knee mechanics also need to be included in the discussion. When the body is resting in a supine position, the only stress on the knees is produced by the generation of muscular force. On the other hand, when the body is in a standing position, the knee joint is supporting a large percentage of the body weight. If the line of action of this force does not pass directly through the center of the knee, the muscle forces must compensate to keep the body balanced. As the subject engages in walking, the center of gravity is rarely over the supporting foot during the single support phase of the gait cycle. To overcome this imbalance, dynamic equilibrium is achieved through the forces of inertia generated by accelerations and decelerations of the body (Maquet, 1984). The muscles of the three joints of the lower extremity act to maintain the body balance during the support phase. When one joint opposes the supporting action of the limb, the other two joints perform compensating actions to maintain stability and support (Winter, 1980).

II. GAIT SEQUENCE AND TIMING

Understanding the sequence and timing of the gait phases is critical for a correct analysis. Walking is characterized by a series of repetitive actions which constitute the gait cycle. A cycle starts with heel contact of one limb and ends with heel contact of the same limb (see Figure 3). The gait cycle can be divided into two major phases: the stance and the swing phases. The stance phase begins with heel strike, then the body weight progressively shifts over the support limb until the end of stance phase at toe off. The stance phase lasts for approximately 60% of the gait cycle. During both the initial and final portions of the stance phase there is a period of double support, in which both limbs are in contact with the ground. The swing phase begins at toe off and continues until heel contact. The swing phase makes up the other 40% of the gait cycle (Harris and Wertsch, 1994; Valmassy, 1995).

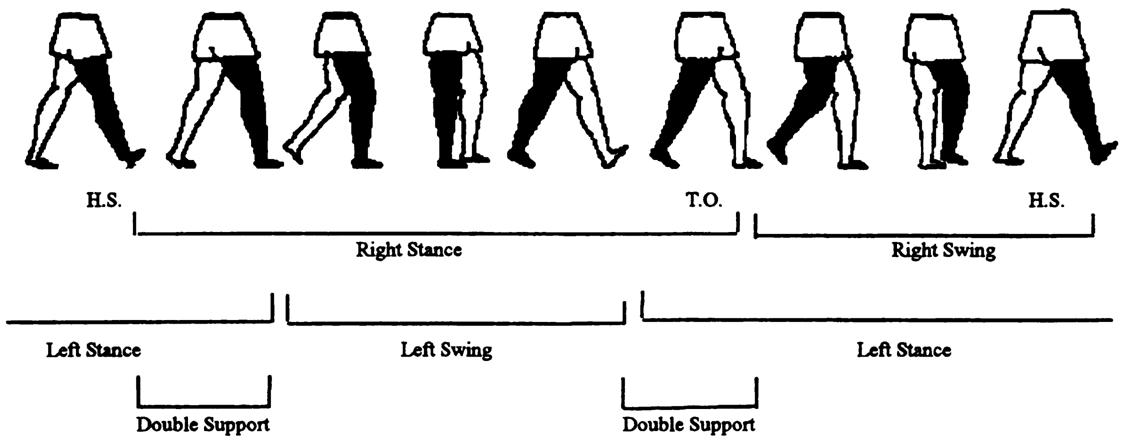


Figure 3: Temporal Gait Cycle Components

The period in which the limb is in contact with the ground is the most important for gait analysis because of the variability in gait patterns observed during this phase. Valmassy (1995) further subdivided the stance phase into four functional phases. The first phase, defined as the contact phase, is initiated as the heel contacts the ground and continues until the opposite limb's toe off. The midstance phase, starts with toe off of the opposite limb and continues until the heel lifts from the support limb. The third phase is the active propulsion phase, in which the opposite limb completes the stride by contacting the ground. And the last phase is passive lift off, which starts from heel contact of the opposite limb until toe off of the support foot. Both the first and fourth phases are double support phases. The contact phase is a period of deceleration, as the limb completes the swing phase and prepares to support the body. During the passive lift off, the body decelerates transferring the weight to the opposite limb which has made contact with the ground.

As the heel makes contact with the ground and throughout the contact phase, specific actions occur in the lower extremity which are characteristic of normal walking. During this period the hip extends, while the knee flexes and the ankle rapidly plantar flexes. The ankle plantar flexion results in a rapid forward displacement of the tibia and a slower movement of the femur, creating knee flexion. At heel contact, the knee is fully extended and it progressively flexes approximately 15 to 18 degrees throughout this phase. Both the actions at the knee and ankle joints are responsible for most of the shock absorption that occurs at contact. The quadriceps and hamstrings are active during this period to aid in knee stability. The quadriceps act to prevent excessive and rapid flexion, while the hamstrings resist hyperextension (Valmassy, 1995).

The functional changes observed during the contact phase due to an injured ACL were explained by Andriacchi (1993) and Ciccotti, Kerlan, Perry, and Pink (1994). Quadriceps avoidance gait was found to be a common adaptation during the initial part of the stance phase among individuals with ACL reconstruction and ACL deficiency. Quadriceps avoidance gait is initially observed during the contact phase when knee flexion is absent. Since knee extension is predominant throughout the support phase of the gait cycle, the demand on the quadriceps is reduced by the lack of knee flexion. Under normal circumstances, the quadriceps eccentrically contract during the contact phase, to balance gravitational forces at the knee. With the loss of the normal function of the ACL, a quadriceps contraction when the knee is near full extension would cause an anterior displacement of the tibia over the femur. Due to the prolonged knee extension observed in the injured knee group, the quadriceps produced a lower than normal net moment, to prevent the tibia from displacing anteriorly (Andriacchi, 1993).

A similar comparison of the gait patterns of subjects with an injury to the ACL, and a control group was performed by Timoney, Inman, Quesada, Sharkey, Barrack, Skinner, and Alexander (1993). The objective of the study was to identify the adaptations which resulted from the injury and subsequent reconstruction. The injured group demonstrated an external flexion moment during the contact and midstance phases of the gait cycle. This external flexion moment would lead to an internal extension moment produced by the quadriceps, conflicting with the theory of quadriceps avoidance gait mentioned earlier. The external flexion moment was found to be lower when compared to the control group, which still demonstrated an adaptation during the contact phase of the gait cycle. Another change experienced by the injured group was a slower loading rate during heel strike. This adaptation was interpreted by the authors as the care taken by the patient when stress was applied to the knee.

Similar adaptations were found in a study performed by Schipplein and Andriacchi (1991), in which they analyzed the interaction of the passive and active stabilizers during walking. The population which took part in this study consisted of patients with varus deformity. The authors calculated joint reaction forces, soft tissue tension, and muscle forces using a mathematical model. Dynamic lateral stability was observed during walking. Stability during walking was assisted by antagonistic muscle action, which helped maintain balance and prevent lateral opening of the joint. The patient group demonstrated a period of either complete flexion or extension during the stance phase, as opposed to two clear periods of flexion demonstrated in the control group. Pretension of the lateral ligaments of the knee was observed during the stance phase in the patient population. The stance phase was shorter in the patient population than in the control group, as the soft tissue maintained joint stability through tension increasing the strain in the soft tissues. The results of this study showed that a minimum muscle force, without antagonistic action, was not enough to stabilize the knee joint during walking. The joint needed to be stabilized dynamically by cocontraction of antagonistic muscles, and or pretension of the lateral soft tissues.

When the single support phase is initiated, the swing leg advances from a posterior to an anterior position relative to the support limb, requiring an increase in stability of the contact leg as well as the trunk. During the midstance phase, extension at the hip is continued from the previous phase, while the knee starts to extend. Extension at the knee is produced by the slower forward displacement of the proximal tibia in comparison to the more rapidly advancing femur. The quadriceps remain active for the initial portion of the midstance, as they prevent flexion at the knee joint and assist in the extension action. Extension of the leg is also assisted by eccentric contractions of the gastrocnemius and soleus muscles (Valmassy, 1995).

Active propulsion is the second portion of the single support period. During this phase the body advances beyond the supporting limb. An external rotation of the lower extremity is produced as the swing leg advances beyond the support leg. The knee and hip are externally rotated during this period. As the weight of the body shifts forward, the hip continues its extending action until hyperextension is reached. Knee extension initiated during midstance is continued until near full extension prior to toe off. This action of extension at the knee controls the lowering of the body's center of gravity and the stride length (Valmassy, 1995).

The second period of double support starts with the opposite limb's heel strike, which initiates the passive lift off phase. As the support limb prepares to enter in the swing phase, momentum of the opposite limb assists in forward advancement. Maximum external rotation of the knee and hip are observed during this phase, as both joints continue the rotation initiated during the active propulsion phase. The hip flexes during this period, reaching a neutral position at toe off. The knee flexes in preparation for leg swing. The quadriceps and hamstrings are active during this period (Valmassy, 1995).

Instability in the knee joint leads to adaptations throughout functional activities. The adaptations made during walking maintain a symmetrical gait pattern (Tibone, Antich, Fanton, Moynes, and Perry, 1986). However, when electromyographic (EMG) data were analyzed, different firing responses were evident in the involved and uninvolved limbs. Electromyographic data complement the kinematic analysis and allow further understanding of the changes in muscle activity as a result of the injury. Muscle contractions can be very important to the stabilization of the knee joint after an ACL injury. Muscles can actively stabilize the knee, and similarly protect it from the shear and rotational forces that can be experienced as a result of the injury.

The use of EMG can assist in the understanding of gait characteristics. A major concern with this procedure is the limited knowledge in the interpretation of the signals.

Basmajian and Deluca (1985), made an effort to explain the neurophysiological components which relate to the field of electromyography. The authors defined the electromyographic signal as "the electrical manifestation of the neuromuscular activation associated with a contracting muscle" (p. 65). The signal is influenced by a set of components from muscle properties and characteristics, to the instrumentation used for recording and reading. An electrode located in the vicinity of the muscle fiber will capture the action potential generated by the movement of ions and depolarization of the post-synaptic membrane.

Several factors are important in the reading and interpretation of an electromyographic signal. The action potential amplitude demonstrates a correlation with the diameter of the muscle fiber, as well as a dependency on position of the recording electrode in relation to the active muscle fiber. A greater amplitude is due to a larger radius of the muscle fiber, and a shorter distance between the recording site and the active muscle fiber. The tissue present between the recording site and the active muscle fiber will create a filter effect, this phenomenon is more obvious for surface electrodes than for indwelling electrodes. Another important factor is the action potential duration, which demonstrates a negative correlation with the conduction velocity of the muscle fiber. Since the recording electrode cannot detect the action potential of one muscle fiber, the signal is a result of combined individual action potentials, commonly defined as the motor unit action potential (Basmajian and Deluca, 1985).

One of the most difficult tasks in any biomechanical analysis is being able to determine if movement falls outside the normal range pattern. A typical pattern will have to be defined in order to highlight the differences observed as a consequence of an external factor; in this case, an ACL injury. Researchers have made an effort to clearly differentiate between typical and atypical gait patterns. Studies conducted by Ciccotti, Kerlan, Perry, and Pink (1994a); Shiavi, Zhang, Limbird, and Edmondstone (1992), described a normal pattern of muscular activity prior to the analysis of the experimental population.

A typical kinematic gait analysis was synchronized with EMG data of the lower leg muscles. Fine wire electrodes were used to test the vastus lateralis, vastus medialis oblique, rectus femoris, semimembranosus, biceps femoris, tibialis anterior, gastrocnemius, and soleus. Subjects were asked to performed seven different everyday activities. Onset time of contraction and peak activity were described for each of the muscles studied. Differences in muscle activities were explained, as they related to the specific demands of the task (Ciccotti, et al., 1994a). Only the results of the walking portion of the study will be discussed in this review. Knee extension, thigh deceleration, and knee hyperextension prevention were all controlled by the combined action of the hamstrings and quadriceps during the terminal part of the swing phase. The quadriceps contracted only during the latter portion of knee extension, since the initial movement of the swing phase is passive. The onset of hamstrings contraction occurred earlier than the quadriceps contraction during the gait cycle. During the terminal part of the swing phase, the tibialis anterior caused the foot to dorsiflexe in order to clear the ground. Basmajian and Deluca (1985) described the muscular activity of the quadriceps and hamstrings as they are synchronized with the gait phases. The quadriceps continued to contract through the contact and midstance phases, as the knee is flexed and extended again. To overcome a tendency to flex the knee during the passive lift off phase, the quadriceps increase in activity. The hamstrings group also contracts during the passive lift off phase, to facilitate the translation of the body over the support leg.

The results of this first study were used by Ciccotti et al. (1994b) as a control group for the second part, in which the subjects participating in the analysis had an ACL deficiency or had undergone ACL reconstruction. Two groups were formed for the study, the first group underwent rehabilitation after the injury, without any surgical intervention; and the second group underwent reconstruction of the joint and rehabilitation after the reconstruction. Results of this study showed that the rehabilitation only group experienced an increase in muscular activity of the vastus lateralis. The increased activity of the vastus lateralis was in response to a protective mechanism to reduce the internal rotation of the tibia, which can be accentuated due to the ACL injury. During the passive lift off phase, the rehabilitation only group, had greater activity in the rectus femoris. The increase of activity of the rectus femoris was a result of a lesser degree of knee flexion attempting to reduce the stress caused by anterior shear forces. In the contact phase of the gait cycle, the first group demonstrated an increase in hamstrings activity. This increase in hamstrings activity is a mechanism to prevent anterior translation of the tibia caused by the contraction of the quadriceps. The longer extension period of the knee during the stance phase acted to prevent the anterior shear of the tibia caused by knee flexion. Greater activity of the soleus is present as a result of a longer extension period of the knee joint. The authors concluded that the differences in muscle activity were most significant in the group that had had rehabilitation only. This conclusion was reached since the EMG profiles of the group that had reconstruction and rehabilitation of the knee joint were similar to the normal population profiles.

With an injured ACL, a contraction of the quadriceps during a position of full extension of the knee would produce an anterior tibial displacement over the femur (Andriacchi, 1993). The anterior tibial translation as a result of an injury to the anterior cruciate ligament was studied by Yack, Riley and Whieldon (1994). The authors tested weight bearing and nonweight bearing isometric exercises, to determine the amount of anterior tibial translation at a 20 degree angle of knee flexion, and at three different force levels. At 50, 75, and 100 percent of body weight, the nonweight bearing exercises demonstrated a greater translation than weight bearing exercises. During the gait cycle, the only time the knee reached an angle of 20 degrees was during the contact phase as it flexed after heel strike (Valmassy, 1995). With a deficient ACL, the anterior tibial translation is limited by secondary structures. In the nonweight bearing conditions, the strain is applied to the primary and secondary structures that control tibial translation. In the case of an injured ACL, the hamstrings control the tibial translation. This increased activity in the hamstrings group was evident in the contact phase, as observed by Ciccotti et al. (1994b). In the study performed by Yack et al. (1994), the hamstrings activity was maintained at less than 10% of its maximum. This restriction in the hamstrings activity allowed the anterior tibial translation to be measured, as it was not being restricted by primary or secondary structures.

Shiavi et al. (1992), employed the method of pattern analysis to determine the differences in muscular activity between two groups of subjects: those with knee injuries and a control group. Electromyography and foot switches were used to determine the temporal variables in the gait cycle. The data from the subjects were combined into clusters as they demonstrated similarities in muscular activity. A template was developed to identify the normal pattern of activity for each muscle studied. The greater number of atypical patterns were found among the injured population. The templates developed from the data of the rectus femoris, vastus lateralis, and vastus medialis, demonstrated a biphasic pattern. For the uninjured group, both the vastus lateralis and vastus medialis had a stronger second phase of activity. Within the injured population, the quadricep muscles lacked a second phase of activity during the end of the stance phase. For this same group, a decrease in magnitude was experienced in the activity of the rectus femoris and vastus lateralis during the contact phase. The injured group also experienced less activity associated with the hamstrings during single leg support. A delay in the onset of activity of the biceps femoris and semitendinosus from the end of the swing phase to the beginning of the stance phase was observed, as well as irregularities in the activity pattern of both muscles. The deviations experienced from the typical pattern by the injured group, were usually characterized by less muscular activity, delay in onset time, and less intense activity during the gait cycle phases.

The analysis of muscular activity done by Shiavi et al. (1992) provided an understanding of how muscles behave during different phases of the gait cycle. With this information in mind, it is important to review how muscular contractions affect the ACL throughout the range of motion. The ACL undergoes significant stress as a result of quadriceps and hamstrings contraction during parts of the flexion-extension range (Renstrom, Arms, Stanwyck, Johnson, and Pope, 1986). The stress applied to the ACL is of particular interest in the rehabilitation phase that takes place after the repair of the ligament. A simultaneous contraction of the flexor and extensor muscle groups may reduce the stress on the ligament (O'Connor, 1993). In an effort to identify the safe zones in which muscle contraction would not put excessive strain on the ACL (Aune, Nordsletten, Skjeldal, Madsen, and Ekeland, 1995; Draganich, Jaeger, and Kralj, 1989), a combination of agonist and antagonist muscle groups have been studied. In addition, flexion angle is of great importance to the load of the ligament due to the biomechanical advantage in the contraction of specific muscles.

To measure the strain transmitted to the ACL by muscle contraction, a strain transducer was attached to the ACL of cadaver specimens and the joint was left to move freely in response to the load applied to the muscle. Renstrom et al. (1986) tested the effect of quadriceps contraction, hamstrings contraction, and simultaneous contraction of both groups. The testing protocol consisted of isometric contractions at 15 degree intervals from full extension to 120 degrees of flexion. The load of the hamstrings was set at 250 N, and the load of the quadriceps was set at 400 N. For the simultaneous contraction test, the quadriceps were set to be in equilibrium with 250 N of load applied to the hamstrings. The results of the study showed the lower strain level range, at a passive state, between 30 and 40 degrees of flexion; with the highest levels found at 105 to 120 degrees, full extension was shown by 0 degrees of flexion. During the quadriceps contraction, the ACL strain was increased 5 % over the passive normal, between 0 and 45 degrees of flexion. Quadriceps contraction decreased the strain on the ACL when the joint position exceeded 75 degrees of flexion. The only significant difference found during the hamstrings contraction, was a lower strain on the ACL produced between the angles of 75 to 105 degrees. A simultaneous contraction produced a higher strain between full extension and 50 degrees of flexion. Simultaneous contraction of the quadriceps and hamstrings becomes a factor during the first 20 percent of the gait cycle, as well as for a short period before and after toe off. Over 50 degrees of flexion, the strain decreased below the passive normal, with a significant difference found at 90 to 120 degrees of flexion. A synergistic contraction of the quadriceps and hamstrings muscles significantly reduced the strain produced by the quadriceps alone by an average of 2 %, with knee flexion ocurring over 30 degrees.

Similar results were found by O'Connor (1993), as he tested the effects of cocontraction of ACL loading. A knee model was develop to facilitate the analysis for the study. When the extensor and flexor muscles contract simultaneously, the joint does not move. However, the muscle forces applied are not necessarily equal and opposite, which implies that the difference has to be made up by ligament forces. If the resultant muscle force acts forward in relation to the tibia, the ACL is placed in tension, and if the resultant force pulls back, the posterior ligament will balance the pull. O'Connor (1993) identified the 22 degree angle of flexion as the critical angle at which the two muscle forces are balanced without any ligament intervention. A simultaneous contraction of the quadriceps and gastrocnemius produces a forward pull, which puts the ACL in tension.

The addition of hamstrings contraction to the contraction of the quadriceps and gastrocnemius unloads the ACL and PCL, especially beyond the critical flexion angle.

Renstrom et al. (1986) concluded that simultaneous contraction of the quadriceps and the hamstrings reduced the load on the ACL. However, the question remained as to the occurrence of simultaneous contraction during flexion and extension. The existence of coactivation of the hamstrings and quadriceps was tested using electromyography. The researchers found that a cocontraction existed between the flexor and extensor groups (Draganich et al., 1989). In agreement with previous studies (Renstrom et al., 1986; O'Connor, 1993), it was shown that the cocontraction of the hamstrings and quadriceps groups reduced the strain on the anterior cruciate ligament and stabilized the knee.

The analysis of muscle activation during different activities gives an indication of synergistic action of muscles during different conditions (McNair and Wood, 1993). In the case of an injured ACL, the cocontraction and antagonistic actions of specific muscles become critical to the understanding of functional stability in the knee joint. The comparison of injured and uninjured ACL's and how the muscle function changes was studied by Solomonow, Baratta, Zhou, Shoji, Bose, Beck and D'Ambrosia (1987), as they investigated the stability component of the joint during flexion and extension exercises. Surface electrodes were used to measure the activity of the quadriceps and hamstrings during maximum voluntary contractions. The pattern found in the ACL deficient group, demonstrated the occurrence of subluxation at about 46 degrees of flexion. This subluxation could be seen by the temporary failure in the torque curve, and the decrease in quadriceps activity which was accompanied by an increase in hamstrings activity. This pattern was observed in most of the patients who did not have any post injury muscle rehabilitation. The authors concluded that muscular strength contributed to the stabilization of the knee joint. The importance of muscular strength could be explained in that two of the patients who had maintained a high level of physical activity after the injury, did not demonstrate a torque failure, but exhibited the same EMG patterns as the injured group.

Contraction of the quadriceps produces an anterior displacement and rotation of the tibia which results in subluxation of the knee joint when the ACL is not able to stabilize the joint (Hirokawa, Solomonow, Lu, Lou, and D'Ambrosia, 1992). The stress transmitted to the ACL by quadriceps contraction depends on the angle of knee flexion (Draganich et al., 1989). A load by the quadriceps between full extension and 80 degrees of flexion produced a significant anterior displacement of the tibia. As the angle of flexion exceeded 80 degrees, the displacement became posterior. The maximum anterior displacement was observed at 30 degrees of flexion. The posterior tibial displacement reflected a minimal strain to the ACL during flexion angles of 80 to 120 degrees (Hirokawa et al., 1992). Similar conclusions were reached by Renstrom et al. (1986).

III. MUSCLE FATIGUE

To this point, the importance of muscular activity in the role of knee stabilization has been explained in relation to injured and uninjured conditions. One factor that can affect the ability of the muscle to stabilize the knee, in the absence of the ACL, is muscle fatigue. Muscle fatigue has neither been studied in relation to injury, nor how it may further affect the functionality of the knee joint. Gerdle and Karlsson (1994), conducted a study to find the relationship between torque levels and the perception of fatigue, and the mean frequency of the extensors of the knee. The test consisted of contraction of the knee extensors at 70%, 25% and 10% of each individual's maximum voluntary contraction. The subjects were instructed to perform the contractions until exhaustion. A rating scale was used to determine the level of muscle fatigue, as it was perceived by the subject. A negative correlation was found between endurance time and torque. The mean frequency of the vastus lateralis and rectus femoris decreased at a greater rate than the vastus medialis with increasing torque levels throughout the endurance time. After the analysis of the results, Gerdle et al. (1994) concluded that mean frequency values were not adequate to determine peripheral fatigue, when working with low torque levels. The detection of muscle fatigue can be a very complicated task, as many factors influence its onset.

Muscle fatigue is believed to be controlled and reduced during learned endurance activities. Repetitive contractions are required to sustain an endurance activity for long periods of time. However, the neuromuscular system reduces the load on the muscles by sharing contraction time with different muscles to delay the onset of fatigue by any single muscle. This sharing of contraction time is true for endurance activities in which relatively slow and sustain contractions take place. Dul, Johnson, Shiavi, and Townsend (1984) defined fatigue as "a continuous process during muscular contractions that culminates in failure to maintain the required mechanical output" (p. 676). As muscle force requirements increase, the endurance time of the muscle contractions decrease.

During muscular load sharing, the fiber type composition plays an important role. Muscles with slow twitch fiber dominance generate more force during the synergistic muscle action, delaying fatigue (Dul et al., 1984).

The information presented in this chapter can be categorized by its relevance to the main topic. Understanding the range of normal characteristics of gait is the first crucial issue. With this basic information, the adaptations due to the anterior cruciate ligament injury can be recognized. A comprehensive examination of cinematographical data, as well as data derived from the use of electromyographic analyses, were necessary to develop a more complete understanding of the subject. The topic of fatigue was introduced, but information which relates to the current study is limited. Due to the lack of understanding of the phenomenon of muscle fatigue by those in the field, as well as its complexity, muscular fatigue has not been widely studied. Due to an injury to the ACL, the muscles of the lower extremity become responsible for joint stabilization. Muscle fatigue can limit the ability to fully stabilize the joint. The combined limitations of the ACL and of the muscles of the lower extremity under a fatigued state may lead to adaptations in the gait pattern. A set of hypotheses were formulated based on results of studies mentioned in this chapter:

Hypothesis 1: individuals with ACL reconstruction will exhibit increased muscular activity during the stance phase of the gait cycle, when compared with the control group.

Hypothesis 2: under a fatigued state, individuals with ACL reconstruction will have lower angular and linear displacements, velocities and accelerations occurring at the knee joint.

Hypothesis 3: subjects with ACL reconstruction will experience adaptations of greater magnitudes at the knee joint under the fatigued state when compared to the control group.

CHAPTER III

METHODS

The purpose of this study was to perform a three dimensional kinematic analysis utilizing EMG in order to examine the effects of muscle fatigue on gait characteristics of individuals with a reconstructed anterior cruciate ligament, compared to uninjured individuals. Subjects were tested under unfatigued and fatigued conditions. The muscles of the lower extremity were fatigued utilizing a submaximal bicycle ergometer test.

I. SUBJECTS

Four male and one female volunteers (mean age of 26.6 years) took part in the study. All subjects were involved in physical activity at the recreational level, at least once a week, at the time of the study. Subject's activity level was taken into consideration during the selection process. Similar levels of activity among subjects were preferred, to reduce variability within groups. Two subjects were part of the treatment (ACL injured) group, while three subjects were in the control (non injured) group. Selection of the injured subjects was based on the following prerequisites: (1) a diagnosed isolated injury of the anterior cruciate ligament, (2) similar ACL reconstructions within 15 months prior to the study's testing date (3) similar strength test results in each knee, and (4) subjects demonstrating results of laxity tests within the normal range (+/- 1 on the Lachman's scale) in the non injured extremity. Subjects in the treatment group were tested for leg strength in both legs prior to the acceptance into the study. Leg strength was tested using a Cybex II dynamometer, to determine the percent of strength recovery as compared with the contralateral leg. A right vs. left strength percentage greater than 15 would excuse the subject from the study. Subjects with no previous history of lower extremity injuries formed the control group. Laxity was tested using the Lachman test, which was performed by a certified athletic trainer. The laxity test was performed on the control group to assure normal range (+/- 1 on the Lachman test) in both knees. All subjects were required to sign an informed consent form (see Appendix A1) and complete a personal data sheet (see Appendix E) prior to their participation in the study. Subjects being considered for the ACL group signed an additional informed consent form (see Appendix A2) to allow strength test screening prior to the actual kinematic data analysis.

Anthropometric measurements were taken by the investigator prior to the first trial, the protocol followed guidelines by Lohman, Roche, and Martorell (1988). These measurements allowed for a quantitative description of the segments of the lower extremity and to check for asymmetries of leg segments. The measurements consisted of weight, standing height, leg length, upper thigh girth, calf girth, and femur breadth; measurements were taken from both right and left sides (see Appendix B). The subjects was asked to identify their dominant side defined as their preferred kicking leg. The subjects were marked with spherical reflective joint markers to facilitate the identification of the joint during the digitization process. Subjects were asked to wear tight fitting dark shorts to allow for a better contrast of the markers. The joints were first marked with water resistant ink markers, to allow for consistant placement of reflective joint markers in case they fell off. Markers were placed over the following bony landmarks on the injured limb: fifth metatarsal, calcaneus, lateral malleolus, lateral femoral epicondyle, greater trochanter, and anterior superior illiac spine (ASIS). For the uninjured side, markers were placed over the fifth metatarsal, calcaneus, medial malleolus, and medial femoral epicondyle. The marked joints were used for the input of position data for analysis of displacement, and calculation and analysis of velocity and acceleration parameters.

Subject preparation for electromyographic data collection consisted of shaving the area over the belly of the biceps femoris and the rectus femoris muscles, cleaning and slightly abrading the skin with a dry wash cloth to reduce resistance to signal uptake. A water based gel was applied to the electrodes to reduce electrical resistance. Hypoallergenic tape was used to adhere the electrodes to the skin. Electromyographic data were collected during all trials, as well as during the fatiguing period.

Each subject was asked to complete three successful trials of walking in both the unfatigued and fatigued states. A trial was considered successful when a complete stance phase of the injured side was within the confinements of the calibrated space. Walking under the unfatigued condition was filmed first, followed by a procedure to induce muscle fatigue. The subjects were then filmed while under a fatigued state. Subjects were given three practice trials in the filming area prior to the initiation of testing. Subjects were asked to walk at what they considered a normal, comfortable speed. All subjects walked with the injured side closer to the cameras, while subjects in the control group walked with their right side closer to the cameras. During the practice trials, the researcher observed if the stance phase occured in the calibrated space. The starting point was marked for each subject, to allow consistency in obtaining the stance phase in the center of the calibrated space. Subjects performed all trials barefooted, to allow for natural movements of the joints during walking.

After the subjects had completed the first three trials, they proceeded to the second phase of the test: fatigue of the lower extremity muscles. To reach a fatigued state, the subjects rode a bicycle ergometer. The resistance was increased every two minutes following the protocol for the submaximal bicycle ergometer test, until the onset of fatigue (see Appendix C for complete protocol). The subjects were asked to rate their perceived muscular fatigue every two minutes prior to the increase in work load. Muscle fatigue was determined by each subject's rating of perceived exertion (RPE). A rating of 8 on Borg's 10 point RPE scale indicated a fatigued state, and the conclusion of the fatiguing process. In conjunction with the Borg scale, electromyographic signals were used to aid in the detection of fatigue. When the integrated EMG signal demonstrated an increase of 200-300% in activity level, muscle fatigue was assumed. After the detection of fatigue by either the perceived exertion scale or electromyographic data, the subjects repeated the three gait collection trials, following the same protocol as was described earlier. After the completion of each trial, the subjects were asked to rate their perceived muscular fatigue. If their rating of perceived exertion dropped more than two points they were asked to ride on the bicycle ergometer starting at the workload at which fatigue was previously detected.

II. DATA COLLECTION PROCEDURES

Testing took place in the Intramural (IM) Sports Circle, at Michigan State University. Instrumentation and equipment were owned by the Department of Physical Education and Exercise Science. Surface electrodes were borrowed from the University of Missouri. Two S-VHS video cameras were used to film the subjects as they walked along a five meter walkway. The cameras were placed at a 60 degree angle to each other, allowing the best view of all markers as the subjects moved in the sagittal plane along the field of view (see Figure 4). The cameras filmed at a speed of 60 frames per second and the shutter speed was set at 1/300 second. Prior to subject filming, a calibration structure was filmed to define the volume over the walkway. The calibration structure consisted of 16 control points, four points at each corner of the rectangular volume. The structure defined a volume of 3 meters long by 1 meter wide by 1.125 meters high. The four corners defining the volume had a symmetrical point distribution; these points were placed at 13.5 centimeters, 47.25 centimeters, 81 centimeters, and 112.5 centimeters from the floor (see Figure 5).

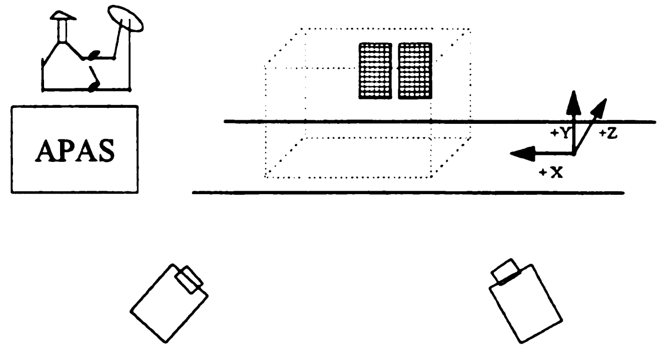


Figure 4: Testing Setup

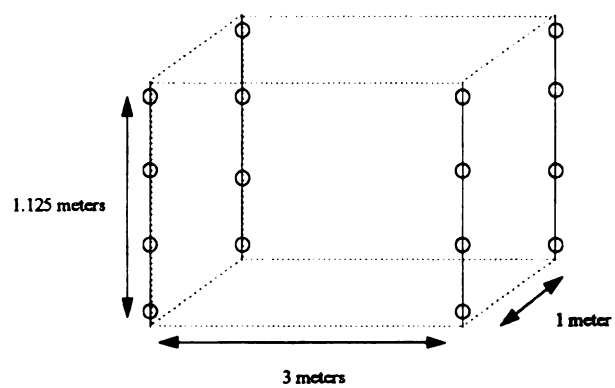


Figure 5: Calibration Structure

Timing lights were placed in the field of view of both cameras to allow synchronization of frames during data reduction. A line was drawn perpendicular to the walkway and EMG data collection was initiated manually as the subject crossed the line. These procedures allowed EMG data to be matched with the kinematic values during data analysis. Surface electrodes were placed over the belly of the rectus femoris and biceps femoris of the injured limb and the leg closest to the cameras of subjects in the control group. The surface electrodes transmited the muscles' signals directly to the analog module of the Ariel Performance Analysis System. Electromyographic data were recorded for three seconds during each trial of gait analysis. During the fatiguing process, EMG signals were recorded every two minutes to monitor muscular activity, prior to each increase of resistance, for a period of three seconds. The investigator read the electromyographic data, if a 200-300% increase in the integrated EMG signal was observed and the subject indicated a rating of 8 on the RPE scale, fatigue was assumed.

III. DATA REDUCTION PROCEDURES

The film data from both cameras was analyzed using the Ariel Performance Analysis System (APAS). The calibrated volume was defined following APAS's requirements, refer to Angeli (1995) (see Appendix D). For the purpose of error reduction, the same fixed point was used for all trials and the first digitized calibration structure frame was used for all trials. In the case of subjects with injuries to the left knee, the calibration coefficients needed to be rotated due to a limitation in the APAS analysis software. APAS software does not allow the changes to be made post digitization. This rotation allowed for the (0,0,0) point in the coordinate system to be the same for all subjects, as they walked in a positive direction according to the defined X axis. This procedure was used to facilitate the comparison of results.

All trials were previewed prior to the initiation of the analysis. The best two trials for each subject under each condition were analyzed. The best trials were selected on the basis of visibility of the markers and location of full stance within the calibrated volume.

Under the grabbing module of the APAS system, the parameters were set to grab frame, starting ten frames prior to heelstrike of the injured limb, through the complete cycle, and ending ten frames following the heelstrike marking the completion of the cycle. The order of joint digitization was the same for all subjects whose right side was closer to the camera: right toe, right heel, right ankle, right knee, right hip, left toe, left heel, left ankle, left knee, left hip, and ASIS (see Figure 6). For subjects whose left side was closer to the cameras, the process started on the left leg, followed by the right leg and ASIS. The transforming module (utilizing the direct linear transformation procedure) combined the two digitized views into three dimensional. The cubic spline technique was used to smooth the data.

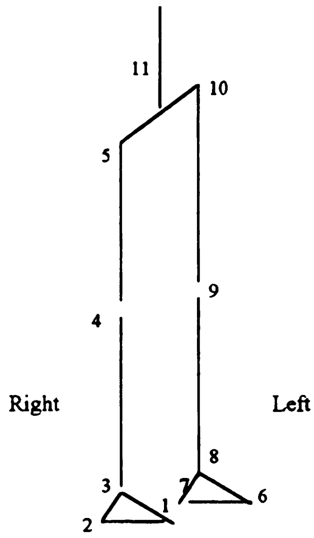


Figure 6: Digitization Order

IV. DATA ANALYSIS

The two best trials for each subject were included in the data analysis. The trials were chosen based on location of the stance phase in the calibrated volume, and the ability to see all the joint markers throughout most of the gait cycle. Linear and angular displacements, velocities and accelerations were calculated and analyzed for each joint digitized. Electromyographic data were synchronized with kinematic data, using temporal parameters, for completion of the analysis. Results were compared between the unfatigued and fatigued conditions for both the injured and uninjured groups. A quantitative description was performed to highlight differences observed within one group under the two different conditions. These observed differences were then compared against the other group. A descriptive analysis best served the purposes of the study, as the adaptations may not be significant to be detected through statistical analyses or the data collection system (APAS).

CHAPTER IV

RESULTS AND DISCUSSION

The results of the three dimensional gait analysis consisted of a comparison of linear and angular parameters under unfatigued and fatigued conditions. Linear and angular displacements, velocities, and accelerations were calculated for three control subjects, and two subjects with ACL reconstruction. Electromyographic (EMG) data were used to determine muscular activity patterns for the quadriceps and hamstrings muscle groups during the walk.

All the data were normalized to percent gait cycle. Angular parameters were expressed as the rotation of the distal segment relative to the proximal segment. Linear and angular parameters were calculated for the side closer to the camera. Angular analysis of the data were presented for sagittal and frontal planes of motion; rotations about the X and Z axes, respectively.

EMG data were analyzed under the unfatigued and fatigued conditions to determine muscle activity characteristics. EMG data were collected for the involved side in subjects of the ACL reconstructed group and on the primarily targeted side for the subjects in the control group. The EMG signal was full wave rectified, and filtered using a 0.3 sec window envelop. The data were synchronized with the kinematic parameters to determine muscular activity throughout the complete gait cycle. Synchronization was performed by identifying foot contact for both EMG and kinematic data. All the signals were amplified and filtered under the same conditions. EMG data for Subject 1 of the control group and Subject 5 of the injured group were not included in the description due to equipment failure during data collection.

The data analysis consisted of a general description of gait patterns observed during two trials for both tested conditions. The results are grouped in two main parts. First, results from angular parameters and muscular activity data for the control group under both unfatigued and fatigued conditions are presented, followed by angular parameters and muscular activity data for the treatment group under these same conditions. A description of normal angular characteristics will precede the report of the results obtained in the study. Secondly, results of linear parameters for the control group under unfatigued and fatigued states, followed by the results of linear parameters for the treatment group under both tested conditions are presented. Normal linear data are presented prior to the results of this study for the control and treatment groups.

I. ANGULAR MOTION

A. Normal Angular Displacement Data

In gait analysis, the concept of normal data needs to be carefully examined. For the purpose of this paper, normal data were identified as a general pattern which was representative of healthy individuals in the "average" population. A note should be made pertaining the definition of normal data: normal data should be inclusive of all the patterns observed in individuals with no history of pathological conditions which might cause adaptations in their gait patterns. Because of the wide range of gait characteristics observed within the "normal" population, the trend is to use the average pattern as "normal". Deviations outside the "normal" pattern may or may not be an indication of pathological conditions affecting the individual's gait.

Angular displacements for the knee joint were presented for flexion/extension and abduction/adduction data. Researchers have shown that a normal flexion/extension angular displacement curve presents two defined periods (Valmassy, 1995). At foot contact the knee is in full extension. Knee flexion of 2 to 3 degrees at foot contact is also considered normal. During the initial 20% of the gait cycle the knee flexes to approximately 15 degrees. This period is normally referred as the initial flexion period. Knee flexion acts to absorb shock and decelerate the body following foot contact. During midstance, from 20% to 40% of the gait cycle, the knee extends and returns to the position of full extension presented at foot contact. Knee extension assists in the forward propulsion of the body as the trunk is carried over the supporting leg. Knee extension continues from midstance into the active propulsion phase. A second and larger flexion/extension period is presented during the swing phase of the gait cycle. The knee initiates the second period of flexion in preparation for the swing phase. At toe off, the knee is flexed approximately 40 degrees. Knee flexion increases throughout the swing phase to provide toe clearance during advancement of the leg. Maximum flexion is reached at about 80% of the gait cycle (midswing), reaching approximately 60 degrees.

When the ankle joint has moved forward of the hip joint, the knee starts its second extension period in preparation for foot contact. (See Figure 7).

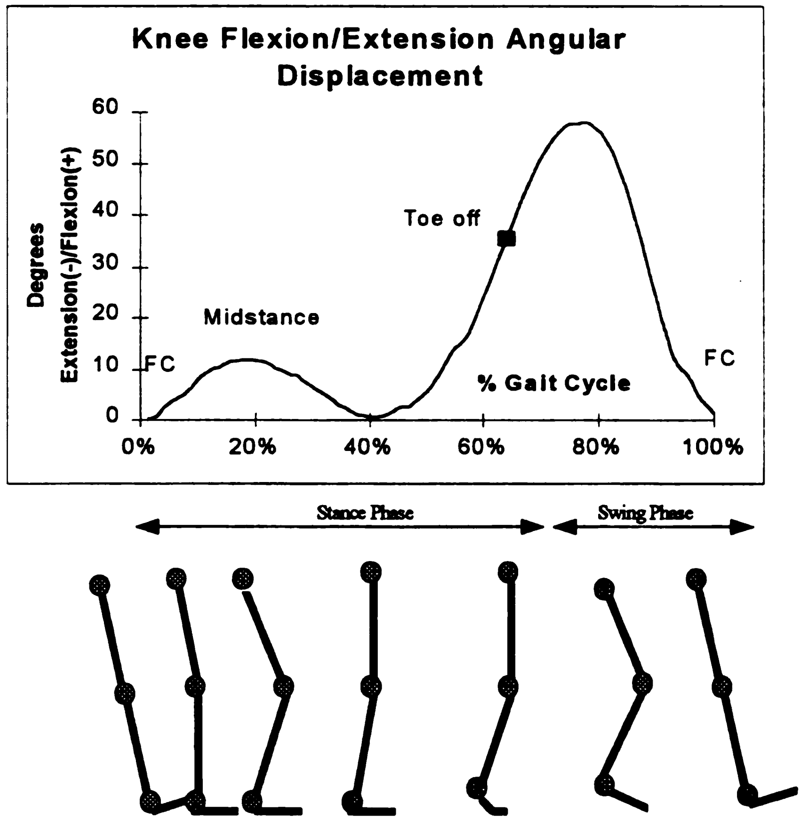


Figure 7: Knee Flexion/Extension Angular Displacement. Normal Pattern.

The normal abduction/adduction movement patterns for angular knee displacement exhibit smaller magnitudes when compared to the flexion/extension patterns. The leg is involved mainly with flexion and extension as it moves primarily in the sagittal plane. Therefore, the knee presents greater displacements for flexion/extension than abduction/adduction. Angular knee displacements along the frontal plane are also harder to detect, because of the smaller ranges of motion. The neutral position was defined as 0 degrees of abduction, representing a perfect alignment of the femoral condyles over the tibial condyles. At foot contact the knee is in neutral position. Throughout the stance phase, the knee maintains this initial position, presenting only small fluctuations in the abduction/adduction displacement of the joint. These fluctuations range approximately 2 degrees in either an abduction or an adduction direction. At the initiation of the swing phase, the knee adducts to approximately 5 degrees, and then enters into an abduction period reaching maximum abduction of 5 degrees. During the terminal 10% of the gait cycle, adduction at the knee is performed to return the joint to neutral position at foot contact. (See Figure 8).

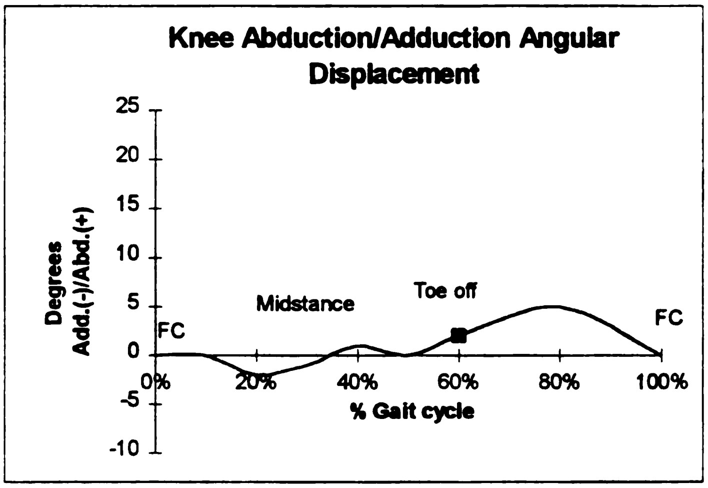


Figure 8: Knee Abduction/Adduction Angular Displacement. Normal Pattern.

Several authors have defined normal muscular activity patterns during gait (Basmajian et al., 1985; Ciccotti et al., 1994). Muscular activity during gait can be analyzed by targeting different muscles of the lower extremity. The muscles usually targeted for EMG data collection include: the rectus femoris, vastus lateralis, biceps femoris, semimembranosis, gastrocnemious, and tibialis anterior. For the purpose of this study, the rectus femoris and vastus lateralis will be combined and identified as the quadriceps group. The biceps femoris and semitendinosis will be identified as the hamstrings group. The gastrocnemious and tibialis anterior will not be included in the discussion. A normal EMG pattern of the quadriceps and hamstrings during a complete gait cycle is shown in Figure 9. Under normal conditions the quadriceps are active at foot contact and act to control knee flexion through eccentric contraction during the initial 20% of the gait cycle. The hamstrings group are also active at foot contact to approximately 15% of the gait cycle. Hamstrings contraction during the initial portion of the stance phase prevents hyperextension at the knee joint. The quadriceps group remains active during the initial portion of the knee extension phase. Quadriceps activity decreases after the initial 25% of the gait cycle. A second quadriceps contraction phase is observed from 50% to 60% of the gait cycle. During the passive lift off phase, the quadriceps act to flex the hip joint and to restrict excessive flexion at the knee joint. During this period, the rectus femoris is the only active muscle of the quadriceps group, since it is the only muscle of the quadriceps group that crosses the hip joint. At 75% of the gait cycle, the hamstrings contract to assist in knee flexion during the swing phase. The hamstrings remain active until the completion of the gait cycle. After midswing the hamstrings contract to control hip flexion as the limb is prepared for foot contact. During the later portion of the swing phase, the hamstrings again act to prevent hyperextension of the knee joint. The quadriceps contract during the last 5% of the gait cycle to assist in knee extension and hip flexion in preparation for second foot contact.

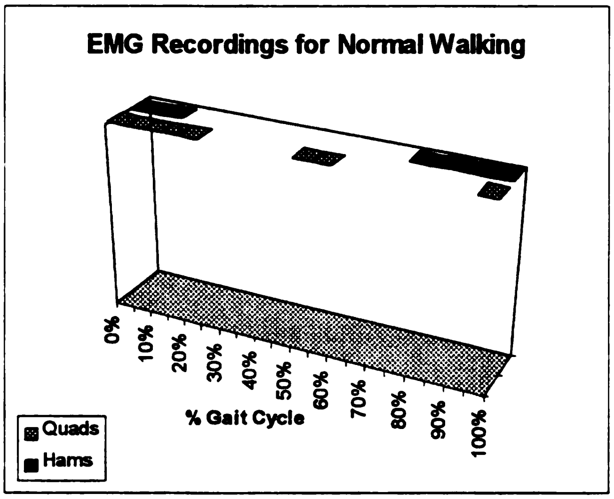


Figure 9: Muscular Activity Patterns for the Quadriceps and Hamstrings Muscle Groups. Normal Data.

Prior to filming, anthropometric measurements were taken of all subjects. No asymmetries were found in the leg segment lengths when anthropometric measurements were compared bilaterally for each subject in the study. In addition, subjects in the ACL reconstruction group were screened, prior to inclusion in the study, for leg strength using a CYBEX Dynamometer. The accepted three to two ratio, of the hamstrings to quadriceps muscle strength, was found in the reconstructed group. In addition, strength differences lower than 15 percent between left and right sides for both quadriceps and hamstrings groups were found. This strength recovery rate of the injured side of more than 85 percent when compared to the contralateral side strength value, is accepted by the National Athletic Training Association (NATA) as the minimal recovery percentage for reintegration of the athlete into practice. Finally, the Lachman's knee laxity test was given to subjects in both groups by a Certified Athletic Trainer. The results of this test were that all subjects in both the control group and the ACL reconstruction group had a range in the Lachman's test of +/- 1, which indicated "stable knees".

B. Control Group

The angular displacement data of the control group were compared to the norms already presented. The data for the unfatigued condition were presented first and a second comparison was made between the data for the fatigued condition and the unfatigued condition. The graph representing the data obtained during the trials are presented in Appendix F, only a representative graph of the common pattern presented by the control group was discussed in this section. Under the unfatigued condition the control group demonstrated the characteristic flexion/extension pattern exhibited during normal gait, see Figure 10. However, none of the subjects experienced full extension at the knee joint during foot contact. The knee joint was flexed approximately 12 degrees at foot contact. This flexion angle was correspondent to full extension of the joint. The range of initial flexion was 15 degrees for both the control and normal groups. The initial flexion period occurred during the first 20% of the gait cycle. The control group, did not demonstrate full extension at the knee joint during midstance. However, the knee joint was extended to the initial 12 degrees of flexion, demonstrating a normal flexion/extension pattern during the stance phase. Subject 2 (CLM2) presented restricted knee extension during midstance, only extending to 18 degrees. The extension period was completed at 40% of the gait cycle for all subjects. During the swing phase maximum flexion of 65-75 degrees was observed at the knee joint. Second foot contact occurred with 12 degrees of flexion at the knee joint, demonstrating consistency and agreement in the amount of flexion presented at foot contact.

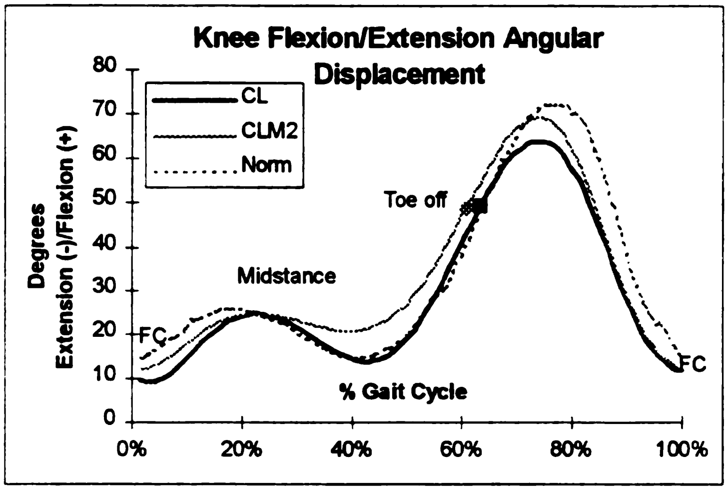


Figure 10: Knee Flexion/Extension Angular Displacement under the Unfatigued Condition. Control Group.

All the subjects in the control group presented the same knee angular velocity pattern. Figure 11 represents the angular velocity graph characteristics of the control group under the unfatigued condition. The knee angular velocity presented a negative slope for the initial 10% of the gait cycle. The angular velocity returned to 0 degrees/second at the conclusion of the knee flexion period. A positive slope of approximately the same magnitude as the previous period was observed between 20% and 30% of the cycle. An opposing negative slope in the velocity curve was observed from 30% to 41% of the cycle. This pattern was observed during the knee extension period. The velocity magnitudes presented during the flexion period were equal to those presented during the extension phase. A negative slope in the velocity curve was observed from 41% to 60% of the gait cycle. This decrease in velocity demonstrated a magnitude of approximately 250 degrees/second. Following toe off, the velocity at the knee joint increased until 90% of the gait cycle. During the last 10% of the gait cycle a negative slope was observed in knee velocity as the limb slowed down in preparation for foot contact.

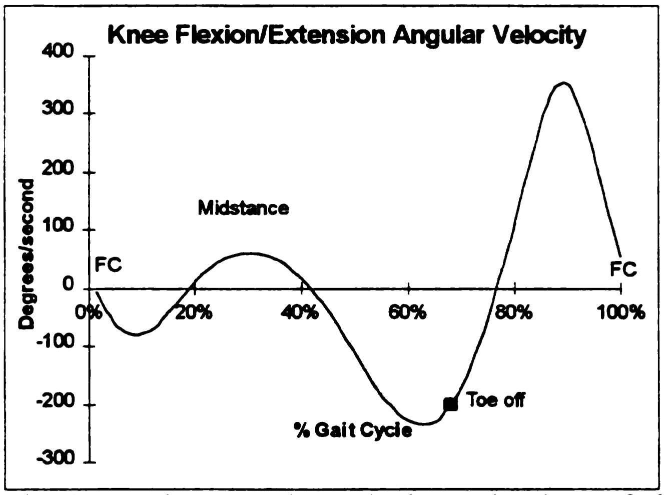


Figure 11: Knee Flexion/Extension Angular Velocity under the Unfatigued Condition. Control Group.

The angular acceleration pattern of the knee joint exhibited by the control group under the unfatigued condition is presented in Figure 12. At foot contact, the knee had reached an acceleration magnitude of -2000 degrees/second^2. An acceleration period was initiated at the knee joint to approximately 20% of the cycle. A deceleration period was observed from 20% to 50% of the gait cycle, corresponding to extension at the knee joint and the initiation of the second flexion period presented during late stance phase. A second larger acceleration period was presented from 50% to 80% of the gait cycle, as the leg moved forward in the swing phase. Past midswing, the knee joint movement decelerated for the purpose of foot contact, reaching the lowest acceleration magnitude prior to the completion of the gait cycle.

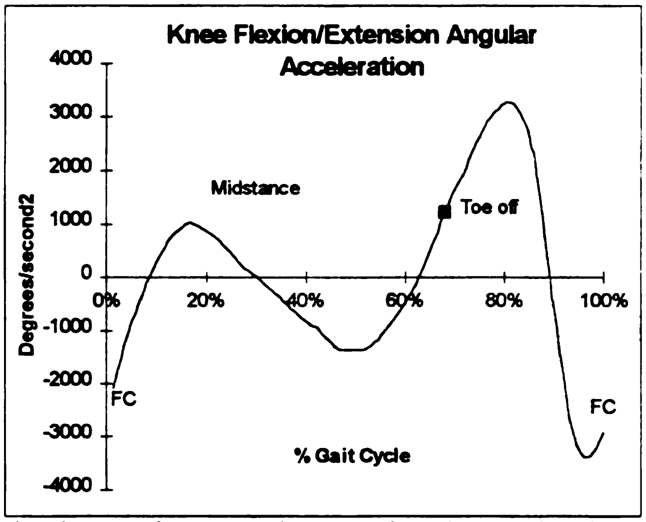


Figure 12: Knee Flexion/Extension Angular Acceleration under the Unfatigued Condition. Control Group.

Differences in the abduction/adduction angular displacements were found across subjects in the control group. Data for both Subjects 1 (CLM1) and 3 (CLF3) are presented in Figure 13, as these two subjects defined the extremes of the control group. At foot contact the knee joint was adducted approximately 2 degrees for all the control group subjects. For the initial 40% of the gait cycle, the knee maintained the adducted position by either further adducting, or abducting and adducting through a range of 2 degrees from the initial position. Differences between subjects were present during the last 60% of the gait cycle. Knee abduction was observed in all control subjects at 40% of the gait cycle. The magnitude of knee abduction varied from 20 to 7 degrees for Subjects 1 and 3, respectively. The time period in which the knee was abducting was different for all subjects in the control group. The knee joint was observed at its most abducted position from approximately 70 to 80% of the gait cycle, for these two subjects.

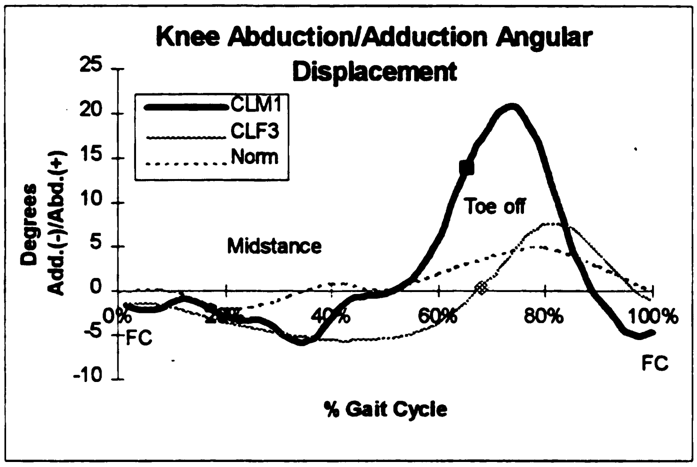


Figure 13: Knee Abduction/Adduction Angular Displacement under the Unfatigued Condition. Control Group.

Due to the differences observed between subjects on the abduction/adduction angular displacements, abduction/adduction angular velocity patterns are presented for Subjects 1 and 3, see Figure 14. Due to the fluctuation between abduction and adduction presented by Subject 1 during the initial 40% of the cycle, the velocity graph also exhibited various changes during this period. The angular velocity maintained a positive magnitude during most of the knee abduction phase, from 40% to 65% of the cycle. Negative velocity values were observed during the last portion of the swing phase, as the knee adducted to the foot contact position. Subject 3 demonstrated fewer fluctuations in the velocity graph for the initial 40% of the cycle. This corresponded to the adduction of the knee experienced after foot contact. After 40% of the gait cycle Subject 3 presented the same general velocity pattern as that presented by Subject 1. Positive angular velocities were observed during abduction, and negative velocities were observed after maximum abduction of the knee.

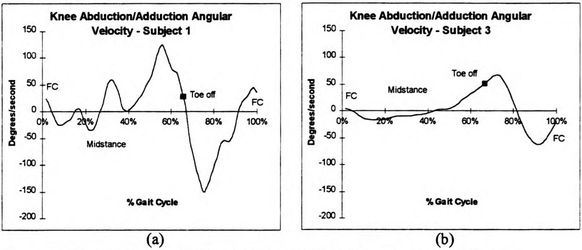


Figure 14: Knee Abduction/Adduction Angular Velocity under the Unfatigued Condition. (a) Subject 1 (b) Subject 3.

The acceleration graphs corresponding to knee abduction/adduction are presented in Figure 15. The acceleration graph exhibited by Subject 1 was indicative of the changes in velocity presented throughout the gait cycle. Abrupt changes were displayed on the acceleration graph describing fast and small changes in the velocity of the knee joint. The acceleration graph for Subject 3 did not demonstrate similar magnitudes changes, presenting a smoother curve. This subject experienced more controlled abduction and adduction of the knee joint throughout the gait cycle.

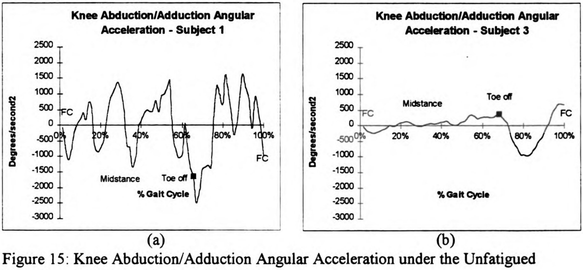


Figure 15: Knee Abduction/Adduction Angular Acceleration under the Unfatigued Condition. (a) Subject 1 (b) Subject 3

The knee angular displacements for the fatigued condition representative of the control group data are presented in Figure 16. The characteristic flexion/extension curve was presented by all subjects under the fatigued condition. Some differences in the range of motion were found when data were compared with the unfatigued state. The degree of knee flexion at foot contact was found to be the same under both conditions, 12 degrees. Initial flexion presented a range of approximately 18 degrees, peaking at 20% of the gait cycle. Under the fatigued condition all subjects presented a greater magnitude of knee flexion during the stance phase. Extension of the knee joint occurred over the expected 40% of the cycle. As the knee returned to the foot contact position, the slope of the displacement graph decreased, causing a prolonged extension period of 25% of the gait cycle. Subject 2 demonstrated limited extension of the knee joint during midstance. However, this reduction in knee extension was consistent with the reduction observed during the unfatigued state. An increase in maximum flexion of 5 degrees over the unfatigued state was demonstrated in the second flexion period of the knee joint. Maximum knee flexion was reached later in the swing phase, when compared to the unfatigued state, due to the longer extension period presented during the stance phase. At second foot contact, at the end of the gait cycle, the knee was flexed 12 degrees, matching the amount of flexion presented during the initial foot contact.

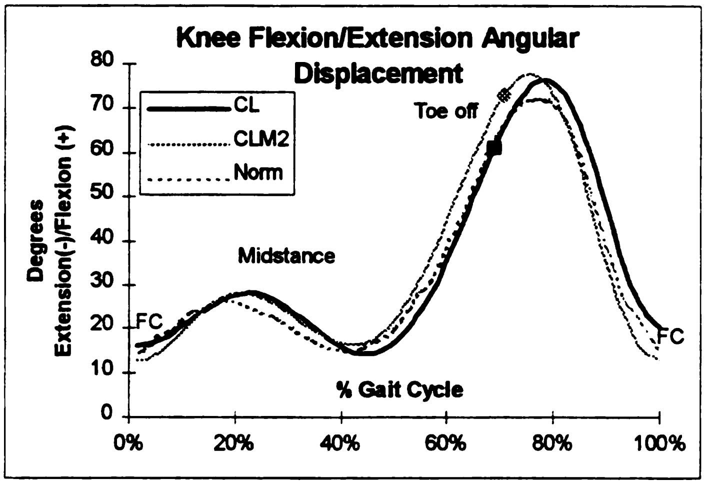


Figure 16: Knee Flexion/Extension Angular Displacement under the Fatigued Condition. Control Group.

Flexion/extension angular velocity patterns at the knee joint under the fatigued state are presented in Figure 17. The pattern presented by these subjects under the fatigued state compared closely to the patterns found under the unfatigued state. Due to the increased knee flexion shown during the stance phase, the negative-positive phase in the velocity graph occurred from foot contact to approximately 25% of the gait cycle. The longer pattern was due to an increase in knee flexion during the stance phase of the fatigued condition. The magnitude of the velocity was greater for the fatigued condition. The second period presented on the velocity graph corresponded to knee extension during midstance. Under the fatigued state, the pattern was shifted so that it occurred later during the gait cycle. The shift was due to the longer flexion phase previously discussed. The magnitude presented during this second phase was greater during the fatigued state when compared to the unfatigued trials. The knee joint extended at a faster velocity to be able to cover a greater range of motion under the same amount of time.

The reduction in knee velocity following knee extension was of greater magnitude under the fatigued state than the unfatigued state. During the swing phase, the knee angular velocity increased to about 90% of the cycle. The velocity of the knee joint reached a greater magnitude under the fatigued state, due to the larger flexion range presented under this state when compared with the unfatigued state.

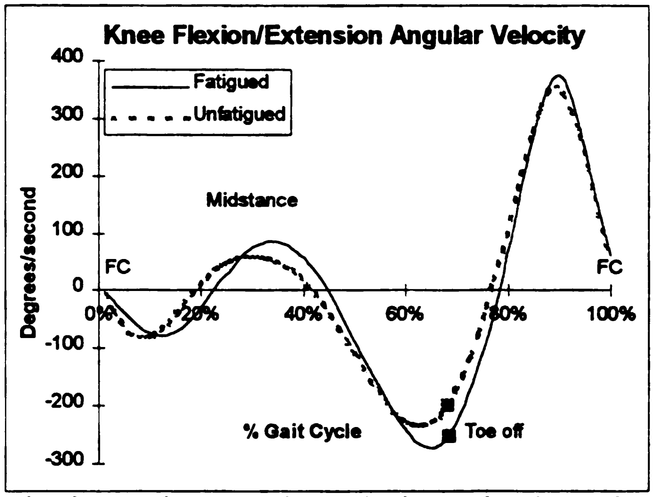


Figure 17: Knee Flexion/Extension Angular Velocity under the Fatigued Condition. Control Group.

The acceleration graph in Figure 18 corresponds to the velocity graph presented for the fatigued trial. The pattern of acceleration of the knee joint under the fatigued state closely followed the pattern presented under the unfatigued state. However, differences in magnitudes were apparent when both conditions were compared. At foot contact, under the fatigued state, the knee joint was decelerated to approximately -1000 degrees/second^2, while the magnitude under the unfatigued condition was almost doubled. During the swing phase the subjects showed greater acceleration at the knee joint under the fatigued condition than under the unfatigued condition. This was due to greater velocity magnitudes presented under the unfatigued state during this period.

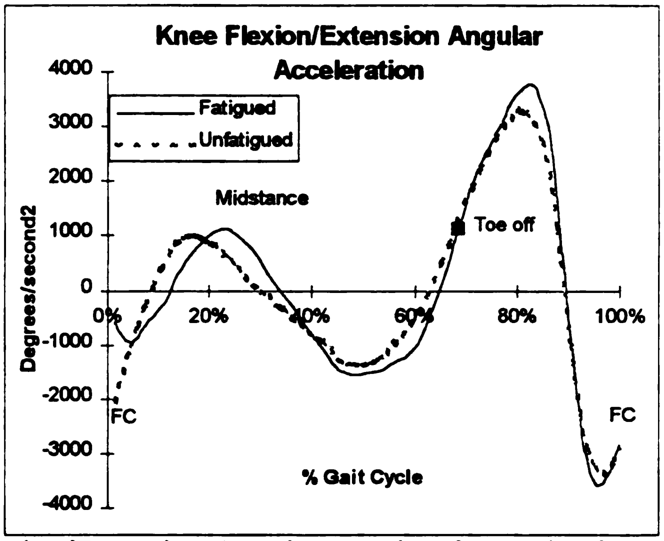


Figure 18: Knee Flexion/Extension Angular Acceleration under the Fatigued Condition. Control Group.

Differences were observed in the abduction and adduction angular displacement magnitudes, under the fatigued state. Subjects 1 and 3 were used for comparison of the abduction/adduction angular displacement characteristics under the fatigued condition, see Figure 19. At foot contact the knee joint was adducted 5 degrees for both subjects 1 and 3. An adducted position at the knee was maintained by abduction/adduction movements of 2 degrees throughout the initial 35% of the gait cycle. Subject 1, demonstrated 2 degrees adduction of the knee during the initial portion of the stance phase. Entering in the second marked period, the knee joint abducted to 20 degrees. Subject 3 demonstrated an abduction of 12 degrees, showing a less pronounced abduction of the knee joint when compared to Subject 1. During the unfatigued state Subject 3 had 7 degrees of abduction. Maximum abduction occurred at approximately 60% of the gait cycle. The largest abduction magnitude was reached earlier during the gait cycle under the fatigued state. However, the time period of abduction of the knee joint was found to be equal under both unfatigued and fatigued conditions.

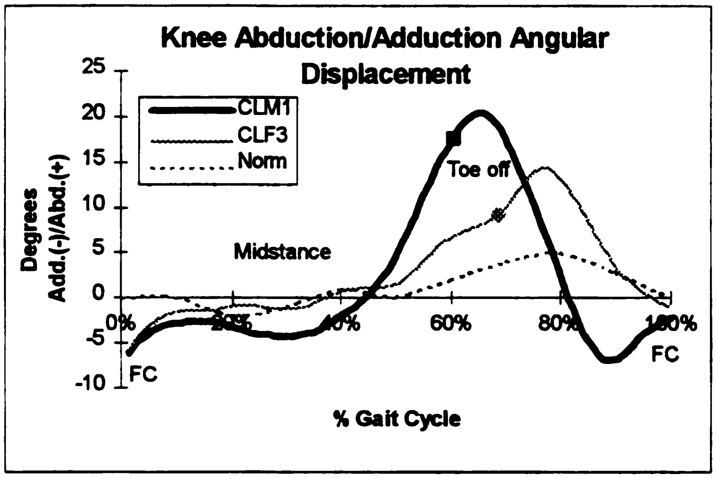
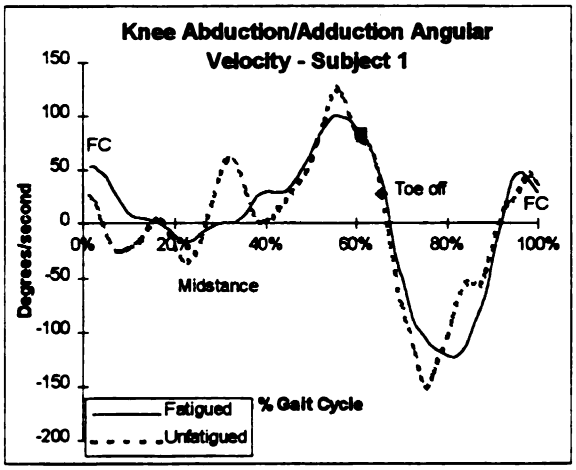
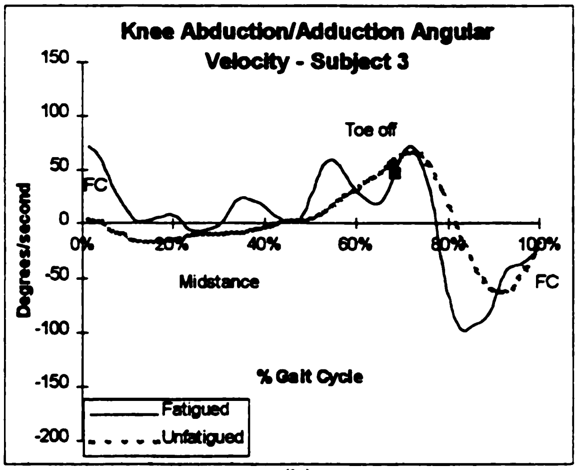


Figure 19: Knee Abduction/Adduction Angular Displacement under the Fatigued Condition. Control Group.

The knee's abduction/adduction angular velocity graphs for the fatigued state are presented in Figure 20. Data for Subjects 1 and 3 were presented due to the differences observed in the displacement data. Under the fatigued condition, Subject 1 demonstrated a more controlled movement pattern when compared to Subject 3. The differences between the two subjects observed in the velocity patterns, suggest less abrupt changes presented by Subject 1. For the initial 35% of the gait cycle, the knee adducted, therefore presenting a velocity graph with no abrupt changes over small periods of time. As the knee was adducted over the initial 20% of the cycle, a negative slope is seen on the velocity graph, due to a slower rate of change of abduction/adduction of the knee joint. A positive abduction slope in the velocity graph was observed from 20% to approximately 60% of the cycle. During the swing phase the velocity was mainly negative until the completion of the cycle. The velocity pattern presented by Subject 3 illustrated the fast changes occurring throughout the gait cycle. Constant changes in the velocity graph were presented for the initial 80% of the cycle. This subject presented the smallest abduction/adduction range of motion throughout the gait cycle, which demonstrated small velocity magnitudes.



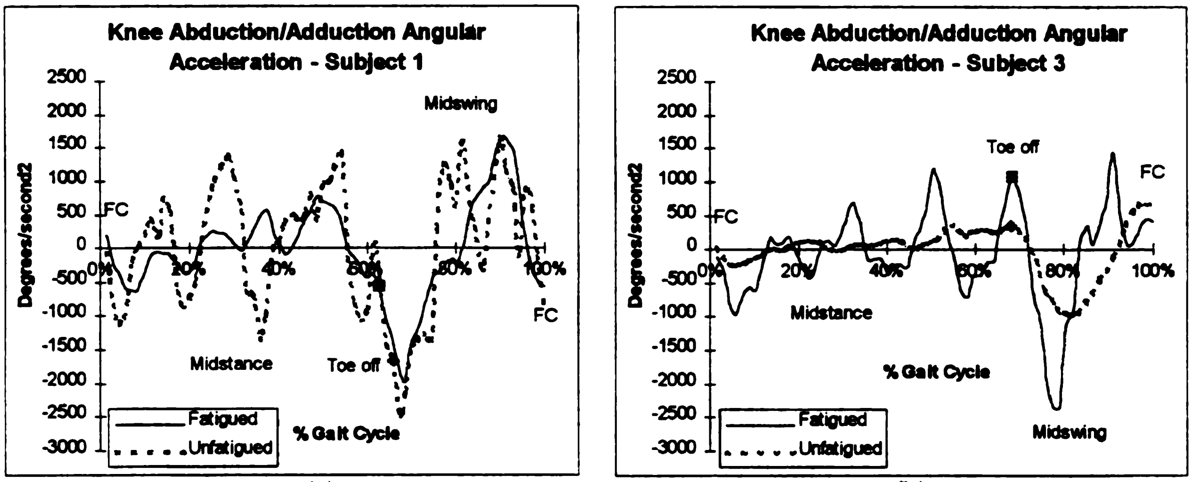
(a)



(b)

Figure 20: Knee Abduction/Adduction Angular Velocity under the Fatigued Condition. (a) Subject 1 (b) Subject 3.

Abduction/adduction angular acceleration graphs for the fatigued condition are presented in Figure 21. The acceleration graphs correspond to the velocity graphs described above. The acceleration pattern for Subject 1 demonstrated changes in magnitude and direction for the initial 50% of the gait cycle. A negative acceleration peak followed the initiation of swing, and a positive peak was experienced after midswing. Subject 1 presented similar angular acceleration patterns for both unfatigued and fatigued conditions. Differences were observed only on acceleration magnitudes. When the acceleration graph for Subject 3 was analyzed, the pattern indicated constant changes in velocity over small periods of time. There was no defined pattern as to the movement of the knee joint throughout the gait cycle. Under the fatigued condition, the acceleration pattern for Subject 3 resembled that of Subject 1 under the same condition. In contrast with the smoother curve presented by Subject 3 under the unfatigued condition.

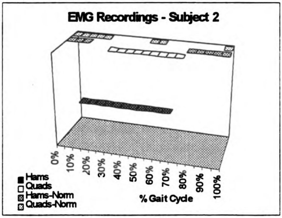


(a) (b)

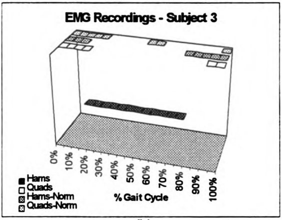
Figure 21: Knee Abduction/Adduction Angular Acceleration under the Fatigued Condition. (a) Subject 1 (b) Subject 3.

Data collected through electromyography for the quadriceps and hamstrings muscles groups were used to determine temporal parameters of muscular activity specific to the two muscle groups during gait. Reference to intensity of contraction was only used for comparison between the activity patterns presented by both muscle groups. Intensity was not used as a measurement of the force of contraction. The muscle activity patterns presented by Subjects 2 and 3 under the unfatigued condition differed from the normal pattern presented earlier. In Figures 22 a and b, the normal activity patterns have been inserted with each subject's pattern to facilitate comparison . The activity pattern presented by the quadriceps group was different for both subjects, however similar activity patterns were seen for the hamstrings group for these two subjects. Subject 2 (Figure 22a) demonstrated quadriceps activity between 30% and 75% of the gait cycle.

This activity corresponded to the extension of the knee during midstance. This subject did not demonstrate any quadriceps activity during the initial flexion period, where the quadriceps are expected to eccentricity contract to control the action at the knee joint. Subject 3 showed quadriceps activity during the initial 20% of the gait cycle and also on the last 10% of the cycle. The quadriceps firing pattern demonstrated by this subject more closely matched the normal EMG pattern described earlier. The quadriceps were active to control knee flexion during the stance phase, and to extend the knee joint during the last portion of the swing phase. Subject 2 demonstrated activity by the hamstrings from 15% to 70% of the gait cycle, while Subject 3 showed hamstrings activity from 20% to 80% of the cycle. During this period the hamstrings acted at both the knee and hip joints. Quadriceps contraction was of greater intensity when compared to the hamstrings.



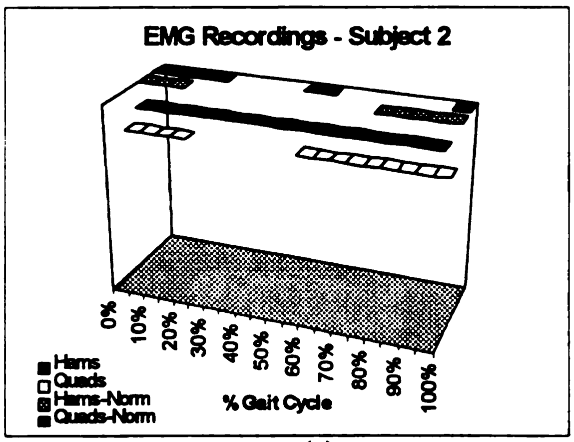
(a)



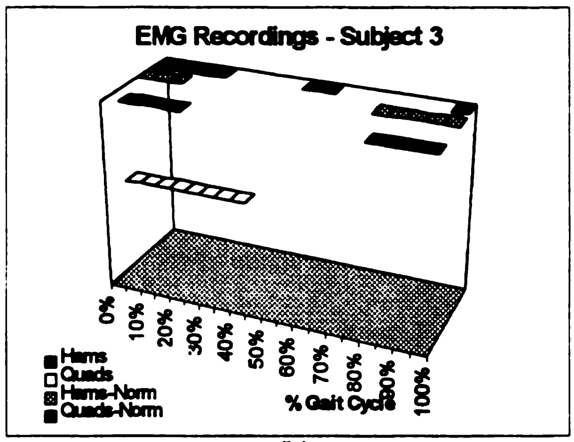
(b)

Figure 22: Muscular Activity Patterns for the Quadriceps and Hamstrings Muscle Groups under the Unfatigued Condition. (a) Subject 2 (b) Subject 3.

During the fatigued trials the muscle activity presented by both muscle groups differed from the pattern demonstrated during the unfatigued condition. (See Figure 23). Although both subjects presented different adaptations on their muscular activity patterns, under the fatigued state, subjects demonstrated a more intense hamstrings contraction when compared to the intensity of the contraction of the quadriceps group. Subject 2 demonstrated quadriceps activity during the initial 20% of the cycle, and during the last 45% of the gait cycle. Subject 3 demonstrated quadriceps activity for the initial 40% of the gait cycle. Subject 2 demonstrated constant activity of the hamstrings group throughout the gait cycle. While Subject 3 demonstrated hamstrings activity from 5% to 25% and the last 20% of the gait cycle.



(a)



(b)

Figure 23: Muscular Activity Patterns for the Quadriceps and Hamstrings Muscle Groups under the Fatigued Condition. (a) Subject 2 (b) Subject 3.

C. ACL Reconstructed group

Both subjects (Subjects 4 and 5) forming part of the treatment group were targeted on the left side. Figure 24 is representative of the knee angular displacements observed for both subjects under the unfatigued condition. Differences were observed between the gait characteristics exhibited by the ACL-reconstructed and control groups under the fatigued state. At foot contact both subjects presented 10 degrees of knee flexion. Subject 4 (ACLM4) demonstrated the characteristic 15 degrees of initial flexion peaking at 20% of the gait cycle. Subject 5 (ACLM5) demonstrated restricted flexion during the stance phase, ranging approximately 5 degrees. The limited knee flexion demonstrated by Subject 5 can be explained by quadriceps avoidance gait. The subject avoided full use of the quadriceps as these muscles eccentrically contract to control the lowering of the body during the initial flexion period. Quadriceps avoidance resulted in limited knee flexion during this phase. A prolonged period of restricted flexion was observed, as the action occurred during the initial 25% of the gait cycle. The longer initial flexion period compensated for the limited flexion of the knee joint, extending the time allowed for the deceleration of the limb. Differences between the two subjects were also found during the extension period of the stance phase. Subject 4 demonstrated limited extension at the knee joint of 5 and 10 degrees for trials one and two respectively, while during the preceding period the knee had flexed 15 degrees. This limited extension seen at the knee joint might be an adaptation to restrict any movement of the femur over the tibia which is primarily controlled by the anterior cruciate ligament. Subject 5 performed "full extension" of the knee joint, since the knee joint returned to 10 degrees of flexion. Maximum flexion reached during the swing phase was 70 and 60 degrees of knee flexion for Subjects 4 and 5 respectively. At the completion of the gait cycle, marked by second foot contact, the knee was flexed approximately the same amount as at the initiation of the cycle analyzed. Subject 5 demonstrated a greater knee extension range during the second foot contact of the cycle.

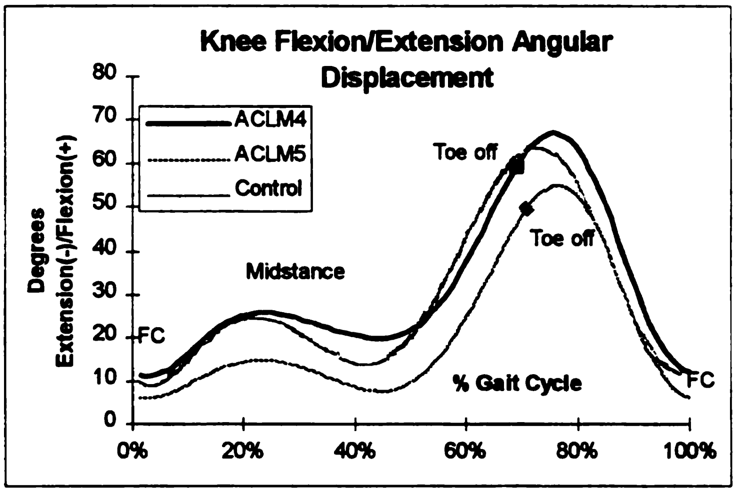


Figure 24: Knee Flexion/Extension Angular Displacement under the Unfatigued Condition. ACL-reconstructed Group.

Due to differences in the angular displacement between both subjects, the angular velocity graphs will be analyzed separately (see Figure 25). Differences in the timing, and in the velocity magnitudes were observed between the treatment and the control groups data. Subject 4 presented the same timing and magnitude in the velocity of the knee during the initial 20% of the gait cycle, as the control group. Subject 5 had a smaller velocity magnitude over a larger period, which was due to the restricted knee flexion presented during the initial 20% of the cycle. From 20% to 40% of the gait cycle, Subject 4 demonstrated the same velocity pattern as the control group. However, the velocity values remained close to the 0 range as compared to the control group. This difference in magnitude was due to the restricted knee extension presented by the subject from 20% to 40% of the gait cycle. Although a normal displacement pattern was presented by Subject 5, differences in velocity with the control group were due to the limited range and prolonged flexion/extension phase during midstance. Although this subject presented restricted flexion, the extension range matched the flexion range during the initial part of the stance phase. Both subjects presented lower velocity magnitudes prior to toe-off, and during the swing phase than the control group. Subject 5 exhibited a slower rate of deceleration of the knee joint prior to foot contact, when compared to the other subjects in the study.

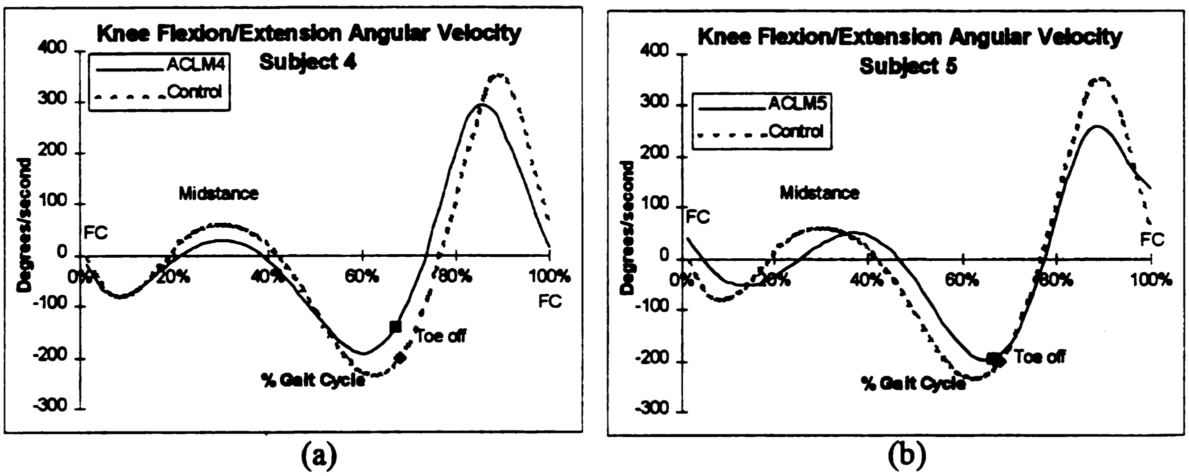
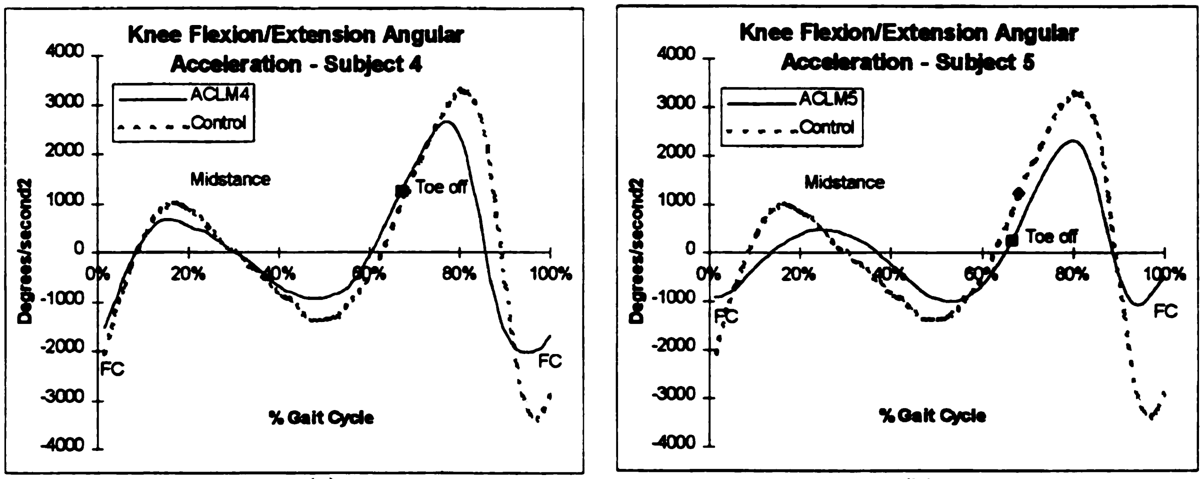


Figure 25: Knee Flexion/Extension Angular Velocity under the Unfatigued Condition. (a) Subject 4 (b) Subject 5.

The angular acceleration graphs of Subjects 4 and 5, under the unfatigued condition are presented in Figure 26. The same acceleration pattern as the one presented by the control group was seen in Subject 4. The acceleration graph for Subject 4 when compared to the control group exhibited smaller magnitudes during the gait cycle. Subject 4 had smaller velocity magnitudes between 20% and 40% of the cycle. This same relative difference in magnitude was observed between 25% and 60% of the cycle in the acceleration graph.

Subject 5 presented differences in both timing and magnitudes when compared to the control group. The magnitude of the first acceleration period was smaller than the one presented by the control group. This lower magnitude of acceleration was seen during a prolonged period of the stance phase. The deceleration period had smaller magnitudes over the same period of time as the control group. Extension at the knee joint during the stance phase was performed at a slower rate than the control group, due to a smaller range of motion exhibited during the same time period. The acceleration values during the swing phase were also lower in magnitudes than those seen by the control group.



(a) (b)

Figure 26: Knee Flexion/Extension Angular Acceleration under the Unfatigued Condition. (a) Subject 4 (b) Subject 5.

Abduction/adduction angular displacement graphs for the unfatigued condition are presented in Figure 27. Both subjects in the ACL-reconstructed group presented a similar pattern of the frontal motion of the knee, which closely resembled the normal pattern. At foot contact both subjects exhibited a neutral position at the knee joint. Throughout the stance phase, Subject 4 demonstrated a 5 degree adduction at the knee. Subject 5, experienced fluctuations of +/- 2 degrees during the initial 50% of the gait cycle. Both subjects abducted at knee during the first half of the swing phase. Subject 4 reached maximum knee abduction of 15 degrees, while Subject 5 had a maximum value of 10 degrees. Knee adduction was initiated past midswing to return the knee to the neutral position at the completion of the gait cycle.

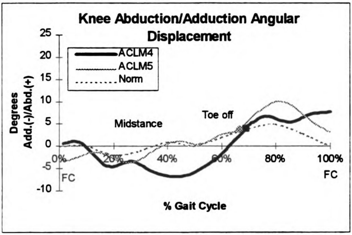
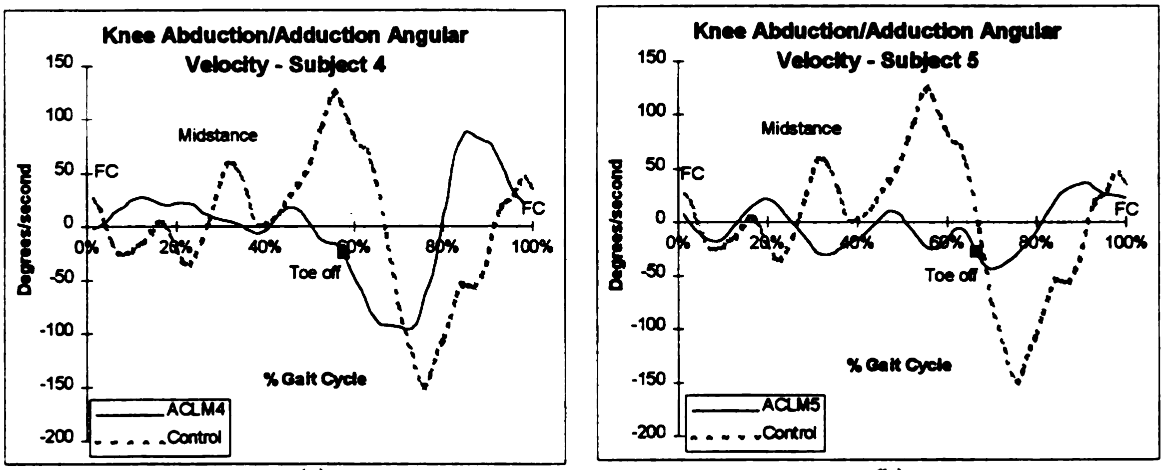


Figure 27: Knee Abduction/Adduction Angular Displacement under the Unfatigued Condition. ACL-reconstructed Group.

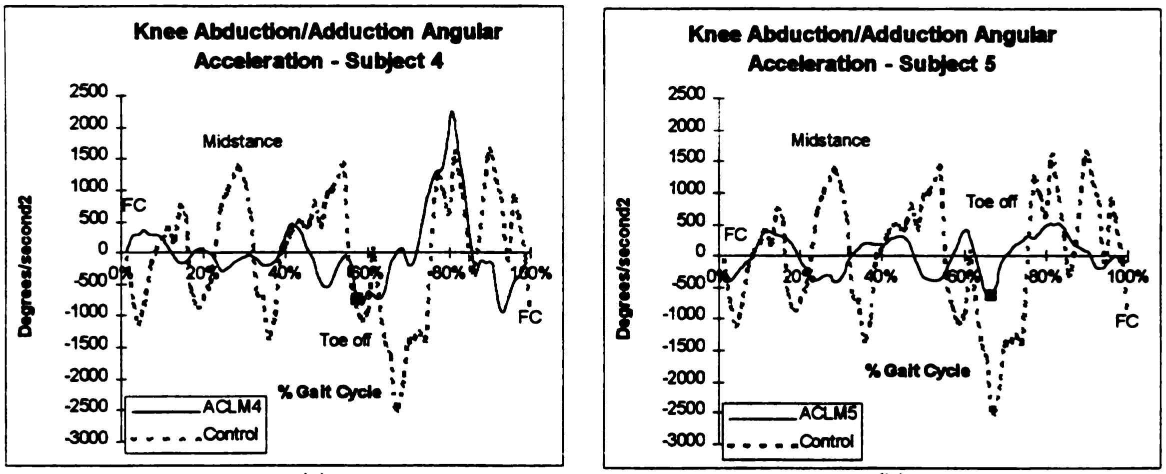
Angular velocity and acceleration graphs for the unfatigued condition are presented in Figures 28 and 29, respectively. The velocity pattern presented by Subject 4 during the stance phase, was smoother when compared to the pattern presented by Subject 5. The angular velocity at the knee joint, for Subject 5, was determined by changes in magnitude of displacement throughout the stance phase. Adduction at the knee was illustrated by a smoother graph for Subject 4. When analyzing the acceleration graphs presented by both subjects, some differences in patterns were identified. The abduction/adduction angular acceleration of the knee presented rapid changes in magnitude, for Subject 4. When the velocity graph, for the same subject, was carefully analyzed, small and abrupt changes were observed during the initial 70% of the cycle.

These changes lead to the high number of magnitude changes presented on the acceleration graph. However, the acceleration magnitudes were tight around zero. The pattern presented by Subject 5, showed lower variability in the magnitudes, when compared to Subject 4. During midswing, Subject 4 exhibited the highest acceleration magnitude of the knee. This acceleration peak was not demonstrated by Subject 5. The difference in the patterns was due to the difference in displacement magnitudes presented during the swing phase, by each subject.



(a) (b)

Figure 28: Knee Abduction/Adduction Angular Velocity under the Unfatigued Condition. (a) Subject 4 (b) Subject 5.



(a) (b)

Figure 29: Knee Abduction/Adduction Angular Acceleration under the Unfatigued Condition. (a) Subject 4 (b) Subject 5.

Under the fatigued condition, both subjects forming part of the treatment group demonstrated adaptations to the flexion/extension pattern experienced during the unfatigued state. At foot contact the knee was flexed approximately 10 degrees, showing a consistent pattern with the amount of flexion demonstrated during the unfatigued state. During initial flexion both subjects increased the range of movement. However, Subject 5 demonstrated limited ranges in flexion when compared with the normal angular displacement, see Figure 30. The timing of peak initial flexion during the fatigued state was consistent with that of the unfatigued state for Subject 4. Subject 5 showed a 5% of gait cycle delay in peak flexion during the fatigued state. The flexion/extension range for Subject 4 was close to normal values. During the swing phase Subject 5 demonstrated a decrease in the flexion magnitude during the fatigued state, when compared to the unfatigued state.

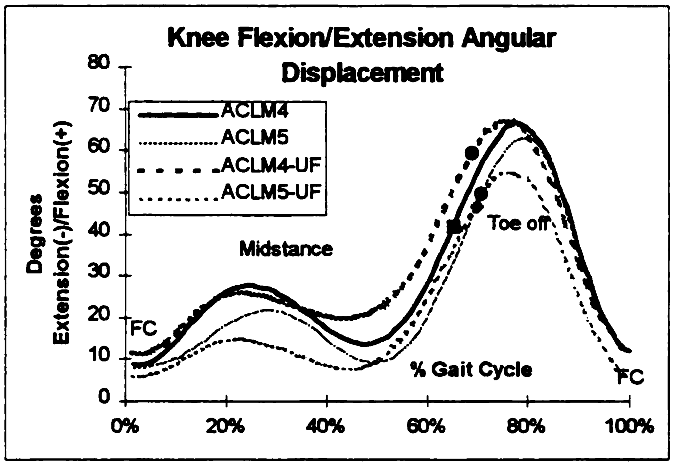
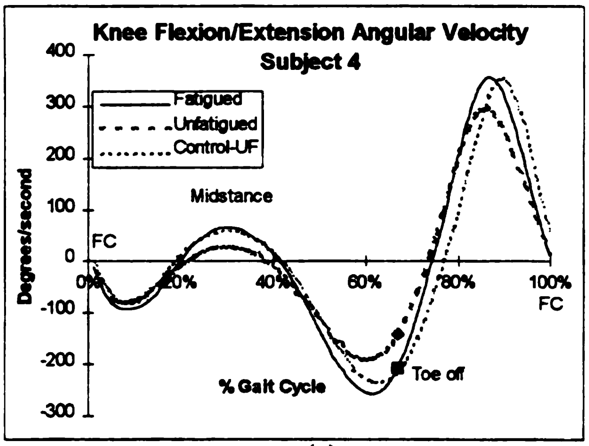
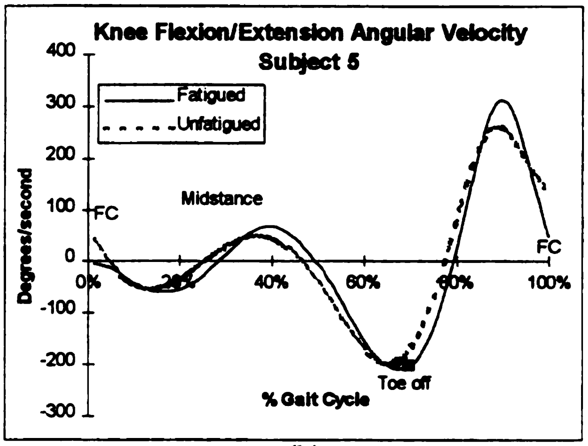


Figure 30: Knee Flexion/Extension Angular Displacement under the Fatigued Condition. ACL-reconstructed Group.

The angular velocity graphs of the knee joint under the fatigued condition are presented in Figure 31. Subject 4 had a velocity pattern in the fatigued condition very similar to that shown by the control group under the unfatigued condition. During the fatigued state Subject 4 demonstrated normal flexion/extension of the knee joint during the stance phase. The increase in the flexion/extension range during the fatigued state led to a normal velocity graph. The lower velocity values seen in the graph of the unfatigued state during the extension period, were not present during the fatigued state. Greater velocity magnitudes were seen prior to toe off, since the knee had to flex over a greater range to prepare the limb for the swing phase under the fatigued state when compared to the unfatigued condition. Subject 5 had similar angular velocity patterns between the unfatigued and fatigued states. Some timing differences were observed due to the changes presented on the angular displacement of the knee joint during midstance, under the fatigued state.



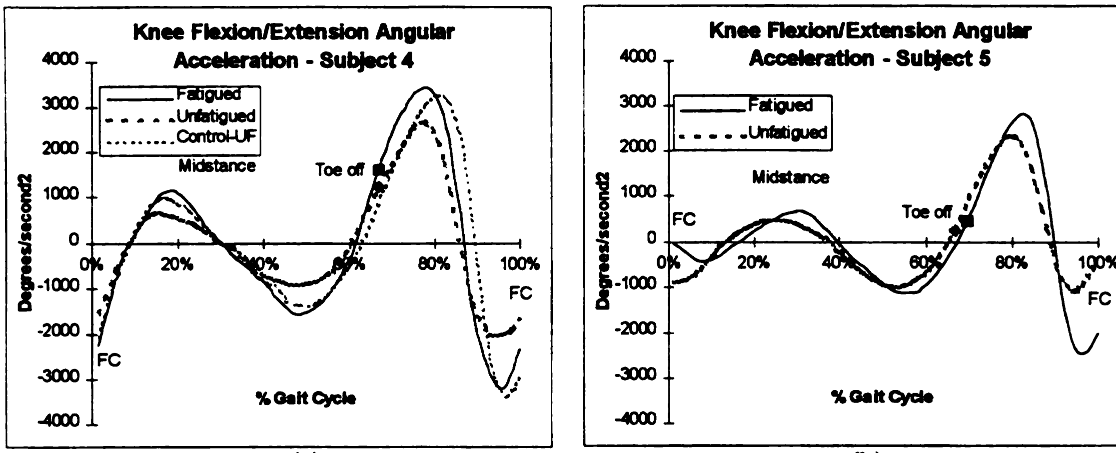
(a)



(b)

Figure 31: Knee Flexion/Extension Angular Velocity under the Fatigued Condition. (a) Subject 4 (b) Subject 5.

The angular acceleration graphs for the fatigued condition of Subjects 4 and 5 are presented in Figure 32. As expected, the acceleration graph for Subject 4 under the fatigued condition displayed the same pattern as the graph of the unfatigued condition for the control group. The pattern presented by Subject 5 resembled the pattern presented by the same subject under the unfatigued condition. Similar acceleration magnitude was exhibited during midstance, however, under the unfatigued condition a steeper slope was observed on the acceleration curve. This change in the slope was due to a shorter period of knee flexion. The knee flexed a greater range during the fatigued condition for Subject 5, leading to a higher acceleration magnitude during the stance phase. In addition, a greater range of knee flexion during the swing phase, for the fatigued state, was shown by a higher acceleration magnitude for this condition when compared to the unfatigued condition.



(a) (b)

Figure 32: Knee Flexion/Extension Angular Acceleration under the Fatigued Condition. (a) Subject 4 (b) Subject 5.

Several differences were seen in the abduction/adduction angular displacement pattern in the subjects of the ACL-reconstructed group when both tested conditions were compared. The abduction/adduction angular displacement graphs for the fatigued condition for both Subjects 4 and 5 are illustrated in Figure 33. Subject 4 maintained the same displacement pattern under both the unfatigued and fatigued conditions, showing differences in magnitude and timing during the swing phase. During the swing phase of the fatigued state, the maximum knee abduction magnitude for Subject 4 was reduced. Magnitude of maximum knee abduction was 10 degrees, and this knee position was maintained until the completion of the gait cycle. At foot contact, Subject 5, exhibited 5 degrees of adduction at the knee joint. During the stance phase, Subject 5 abducted the knee joint and reached the neutral knee position at 40% of the gait cycle. The knee maintained a relatively constant adduction displacement throughout the stance phase. Subject 5 demonstrated the same angular displacement pattern, during the swing phase, under both tested conditions.

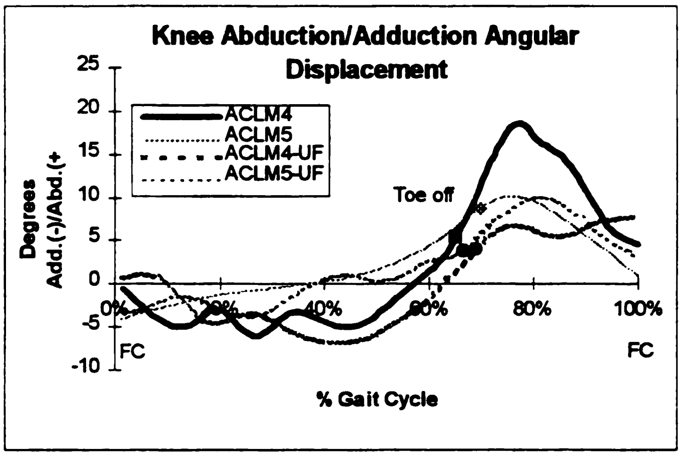
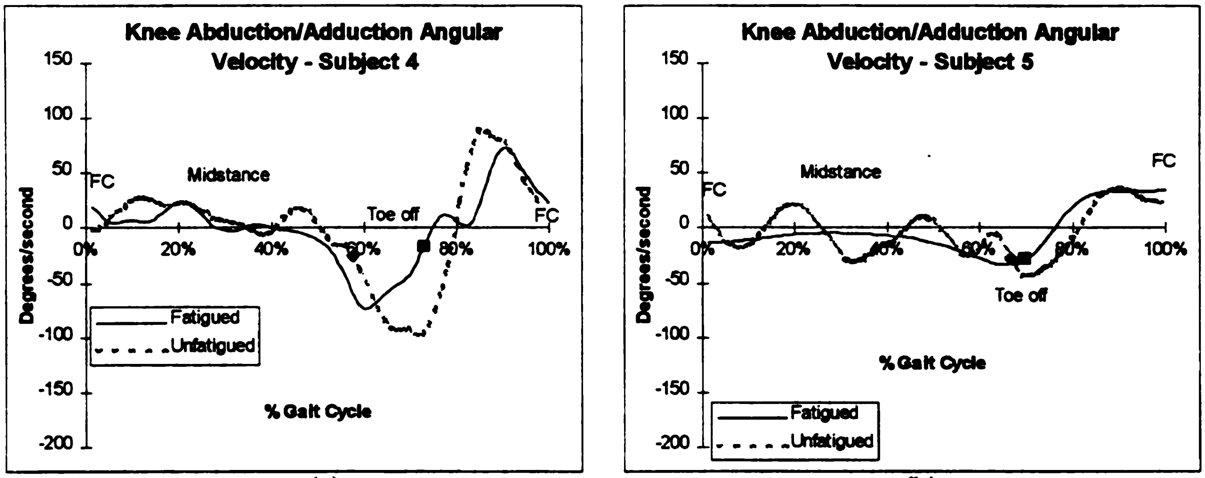


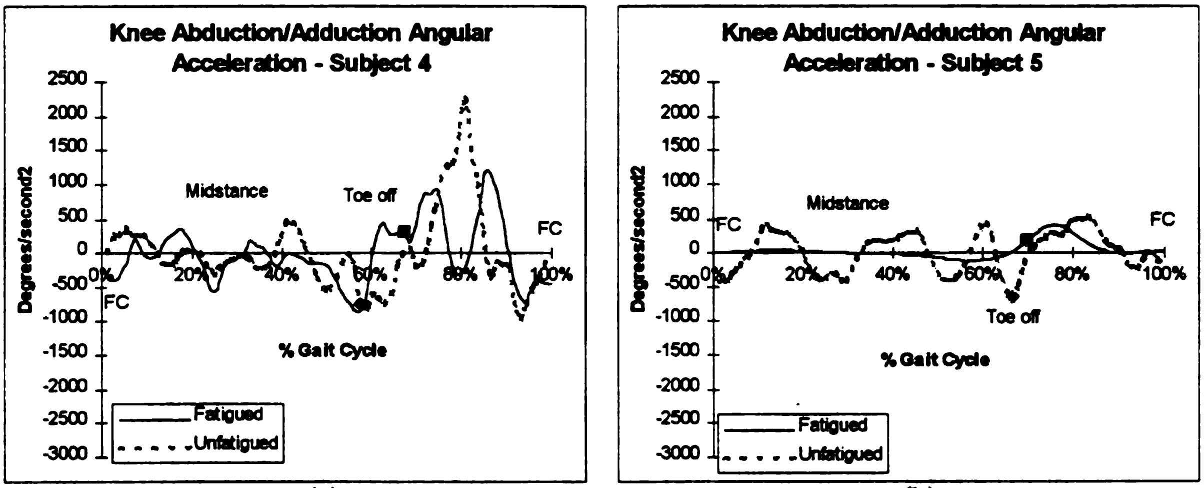
Figure 33: Knee Abduction/Adduction Angular Displacement under the Fatigued Condition. ACL-reconstructed Group.

Abduction/adduction angular velocity and acceleration graphs for the fatigued condition for Subjects 4 and 5 are presented in Figures 34 and 35. Magnitude changes were observed for Subject 4 when the velocity and acceleration patterns under the fatigued state were compared to those presented under the unfatigued state. Lower velocity magnitudes were observed during the fatigued state. The acceleration pattern presented by Subject 4 under the fatigued state, compared to that presented by Subject 5 under the unfatigued condition. The peak acceleration value was smaller in magnitude during the fatigued condition than the unfatigued condition, due to a smaller abduction displacement which occurred during the swing phase for Subject 4. Under the fatigued condition, Subject 5 displayed a constant velocity value throughout the stance phase. Knee acceleration was 0 degrees/second^2 for the initial 60% of the gait cycle. This acceleration was due to the uniform abduction displacement presented by the subject during the initial portion of the gait cycle.



(a) (b)

Figure 34: Knee Abduction/Adduction Angular Velocity under the Fatigued Condition. (a) Subject 4 (b) Subject 5.

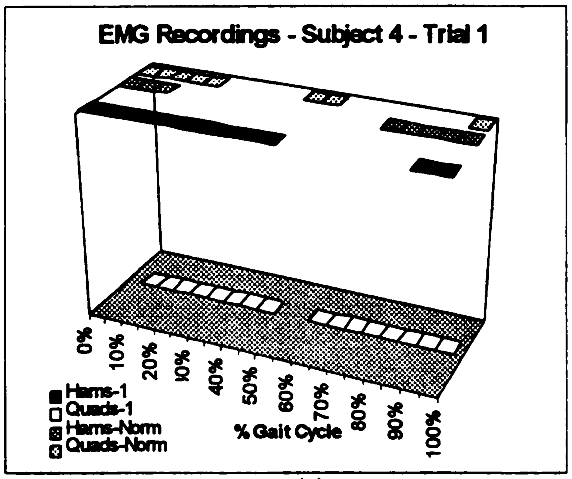


(a) (b)

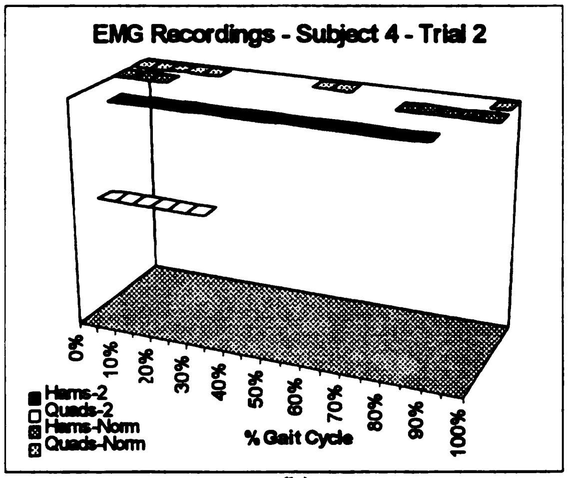
Figure 35: Knee Abduction/Adduction Angular Acceleration under the Fatigued Condition. (a) Subject 4 (b) Subject 5.

Muscular activity characteristics under the unfatigued condition for Subject 4 are presented in Figure 36. The subject demonstrated greater intensity of muscular activity for the hamstrings group when compared to the intensity level of the quadriceps group. The quadriceps were active from 10% to 50% of the gait cycle. This portion of the gait cycle corresponds with knee extension during midstance and the passive lift off phase.

Even though the quadriceps were not contracting during the initial 20% of the gait cycle, indicating quadriceps avoidance, the gait characteristics were not representative of quadriceps avoidance gait. In quadriceps avoidance gait, limited flexion occurs over the initial 20% of the gait cycle. However, for Subject 4 a full flexion range was seen during the stance phase. The quadriceps were active during the initial 30% of the cycle for the second trial tested. The activity pattern observed for the hamstrings group was also different for the two trials analyzed. During the first trial, the subject presented an active hamstrings group during the initial 55% of the gait cycle, and a second activity phase for the last 10% of the gait cycle. In the case of the second trial, the same subject exhibited activity on the hamstrings between 10% and 90% of the gait cycle.



(a)



(b)

Figure 36: Muscular Activity Patterns for the Quadriceps and Hamstrings Muscle Groups under the Unfatigued Condition. Subject 4. (a) Trial 1 (b) Trial 2.

The muscular activity pattern presented during the fatigued state for Subject 4 is illustrated in Figure 37. Under the fatigued condition, Subject 4 demonstrated greater intensity of contraction for the quadriceps group when compared to the intensity of contraction demonstrated by the hamstrings group. Quadriceps activity was present during the initial 20% of the cycle, and from 50% to 90% of the gait cycle. The quadriceps acted to restrict knee flexion during the initial 20% of the gait cycle. The activity of the quadriceps muscles recorded during the latter part of the cycle was due to flexion at the hip joint and extension of the knee joint prior to second foot contact. Hamstrings activity was exhibited for the initial 30% of the cycle and 40% of the gait cycle prior to second foot contact.

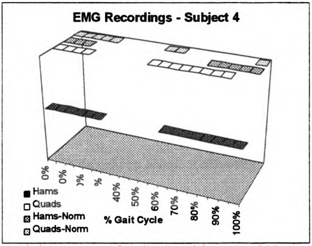


Figure 37: Muscular Activity Patterns for the Quadriceps and Hamstrings Muscle Groups under the Fatigued Condition. Subject 4.

II. LINEAR MOTION

A. Normal Data

Linear parameters of gait were analyzed with the purpose to obtain information about the relative movement of the knee joint with respect to the ankle and hip joints. Hip, knee and ankle joint markers defined the two segments of the lower extremity. The relative movements of these segments were used to define the knee angle for angular data analysis. All subjects walked in the positive X direction, according to the coordinate system used in the study. An increase in joint displacement along the X-axis would indicate the advancement of the joint along the walkway. Linear displacement data for the hip, knee and ankle joints were analyzed in relation to the alignment of the joints at specific periods in the gait cycle. Joint alignment was exhibited when the position of two or all three joints was the same along the X-axis, see Figure 38. The same relative alignment of two adjacent joints along the X-axis, positioned the defined body segment perpendicular to the ground.

Six aligned positions of the three joints occurred during the gait cycle. At foot contact, the ankle joint is the joint furthest from the origin, therefore demonstrating the greatest value along the X-axis. Once the flat foot position is obtained, movement at the ankle joint is limited throughout the stance phase. Following foot contact, knee flexion is performed. Flexion at the knee resulted from the shank segment moving faster than the thigh. This forward movement of the shank caused the first ankle-knee alignment early in the stance phase. At approximately 20% of the gait cycle, as the knee reached maximum flexion, the position of the knee joint along the X-axis was furthest away from the origin. At this point an ankle-hip alignment is observed. Following maximum flexion of the knee, the forward displacement of the knee joint decreased. While the hip joint maintained the same rate of displacement, the faster moving hip joint caught up with the knee allowing for the first hip-knee alignment during midstance. The hip joint continued to move at a faster rate than the knee joint until the initiation of the second knee flexion

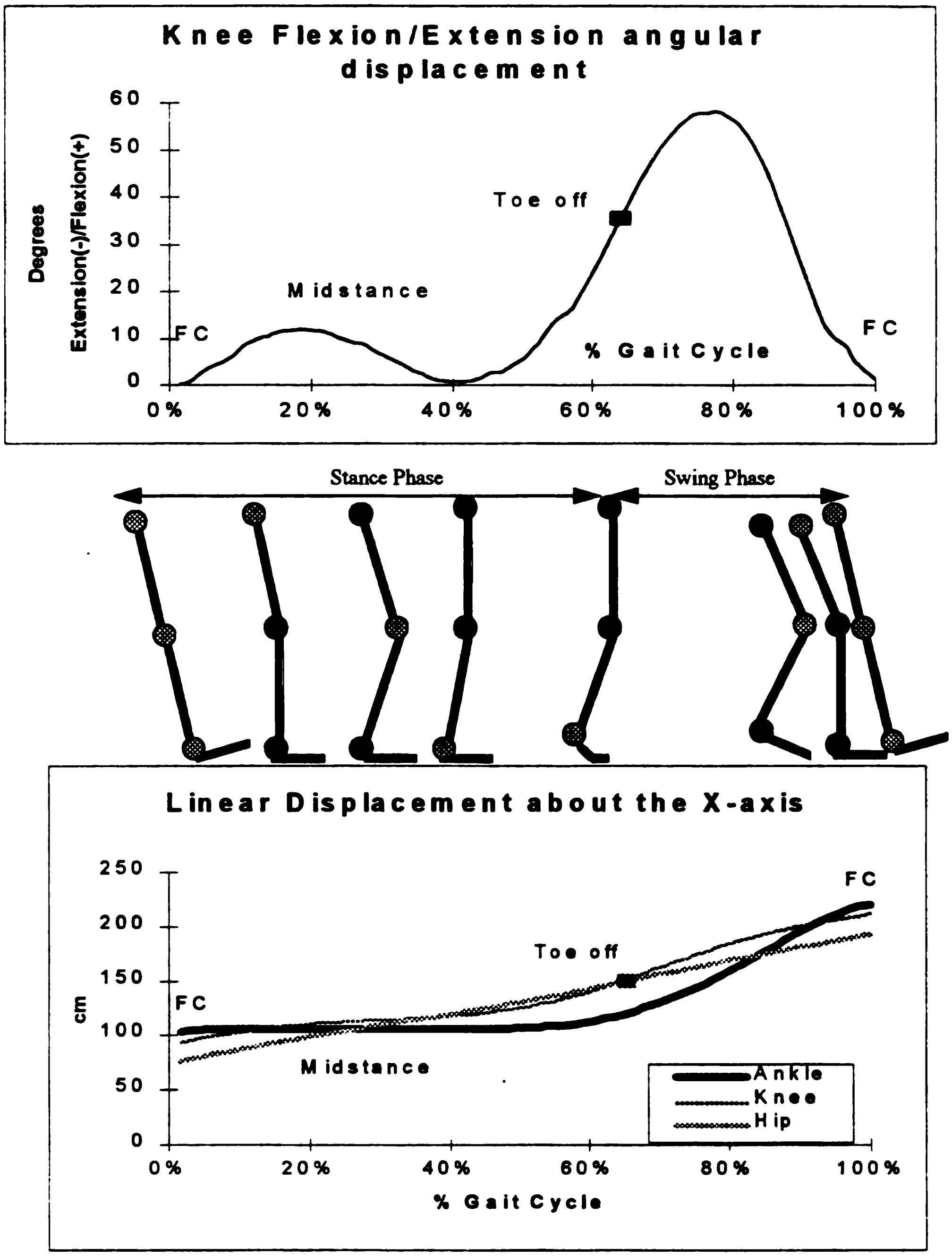


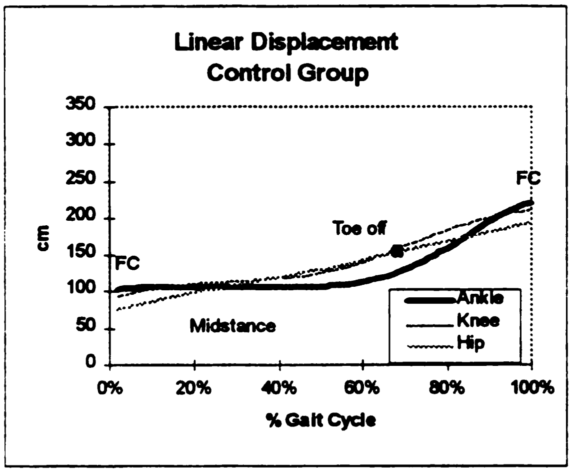
Figure 38: Joint Alignment

period. Flexion of the knee joint increased the rate of displacement of the knee joint, allowing for the second alignment of the hip and knee joints prior to toe off. As the limb prepared to enter into the swing phase, the ankle joint initiated its forward movement.

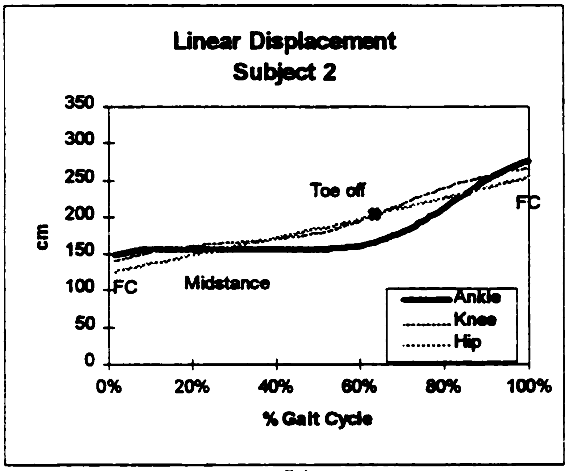
During the swing phase the ankle joint demonstrated the highest velocity of all three joints. Flexion at the knee joint caused the knee to be the furthest point from the origin, and as it occurs during the stance phase, an alignment of the hip and ankle joints is observed at the point of maximum knee flexion. Following midswing, the highest velocity is seen in the shank segment, allowing the ankle joint to catch up with the knee joint. The ankle-knee alignment occurs late in the swing phase and is the last alignment seen during the cycle.

B. Control Group

As illustrated in Figure 39, all the subjects in the control group demonstrated similar linear displacement patterns of the lower extremity joints during the unfatigued state. In addition, all the subjects demonstrated the six normal alignments of the joints throughout the gait cycle. The analysis of the linear displacement of the joints was focused primarily on the stance phase, since most of the adaptations in gait characteristics occur during this period. The first alignment of the ankle and knee joints was found at approximately 10% of the gait cycle. As flexion at the knee joint continued, the hip and ankle joints aligned at approximately 25% of the cycle. This second alignment was observed past the maximum flexion point which occurred at 20% of the cycle. Prior to the point of "full extension" of the knee during midstance, the first alignment of the hip and knee joints was observed, occurring at 37% of the gait cycle. The second alignment between the hip and knee joints was observed prior to toe off at approximately 62% of the gait cycle. The differences shown by Subject 2 during the angular displacement of the knee joint, were also observed when analyzing the linear displacement pattern of the knee joint for this subject. The linear displacement graph for the unfatigued condition for Subject 2 (Figure 39b) illustrated differences in the slope of the knee's linear displacement graph during the stance phase when compared to the other subjects in the control group. Specifically, Subject 2 had limited extension of the knee joint during the stance phase and following maximum flexion, the knee's linear displacement plateaued. This plateau was longer for Subject 2 with respect to the other subjects, indicating a prolonged flexion period as there was limited extension at the knee joint.



(a)



(b)

Figure 39: Linear Displacement under the Unfatigued Condition. (a) Control group (b) Subject 2.

When the linear velocities of the three joints under the unfatigued condition were examined several characteristics were found and are illustrated in Figure 40. First, velocity at the ankle joint for the initial 10% of the gait cycle had a negative slope. The ankle joint was decelerated to 0 cm/sec, and maintained this magnitude until 40% of the cycle. This velocity magnitude was due to the limited displacement presented at the ankle joint during the first portion of the stance phase. Following this initial period of the stance phase, the velocity of the ankle joint increased until midswing, and then decreased as the joint slowed down prior to second foot contact.

At the knee joint, the velocity decreased during the initial flexion period and increased from the maximum flexion point until toe off. This velocity pattern corresponds to the advancement of the body over the supporting leg. During the second flexion period, the slope of the velocity curve was reduced past midswing as the velocity at the knee joint decreased.

The linear velocity of the hip joint was of a smaller magnitude than the other two joints which indicated that the rate of the hip's linear displacement was relatively constant throughout the gait cycle. A slight negative slope was observed for the initial 40% of the cycle. However, the hip's linear velocity pattern did not present any specific characteristics that could be identified during the gait cycle.

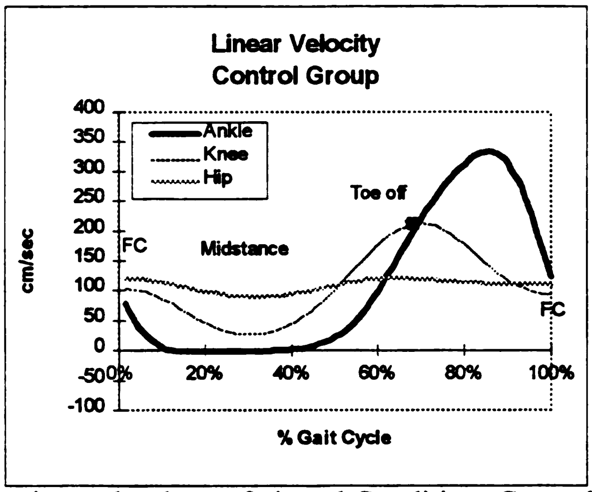


Figure 40: Linear Velocity under the Unfatigued Condition. Control Group.

Linear accelerations of the ankle, knee and hip joints under the unfatigued state, are presented in Figure 41. For the initial 15% of the gait cycle, the slope of the acceleration graph was positive. From 15% to 40% of the gait the acceleration values approximated 0 cm/sec^2 due to the constant velocity presented at the ankle joint. An acceleration period of the ankle was observed from 40% to 65% of the cycle due to preparation for toe off, and decelerated following toe off until midswing. The ankle joint accelerated following midswing, to move the foot in front of the knee joint as the limb prepared for foot contact.

At the knee joint, a deceleration period was observed for the initial 15% of the gait cycle. The joint decelerated as the foot struck the ground and flexion at the knee joint controlled the forward motion of the body. The knee accelerated from 15% to approximately 55% of the gait cycle, decelerated during the initial portion of the swing phase, and then accelerated from midswing to foot contact. Due to the relatively constant velocity observed at the hip joint throughout the gait cycle, the acceleration magnitudes were close to 0 cm/sec^2.

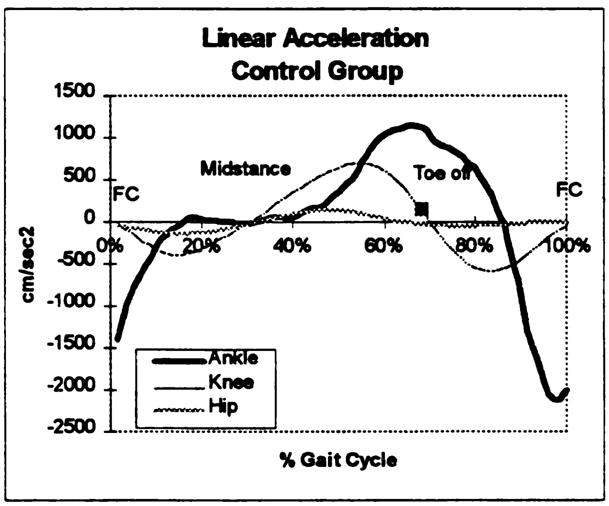
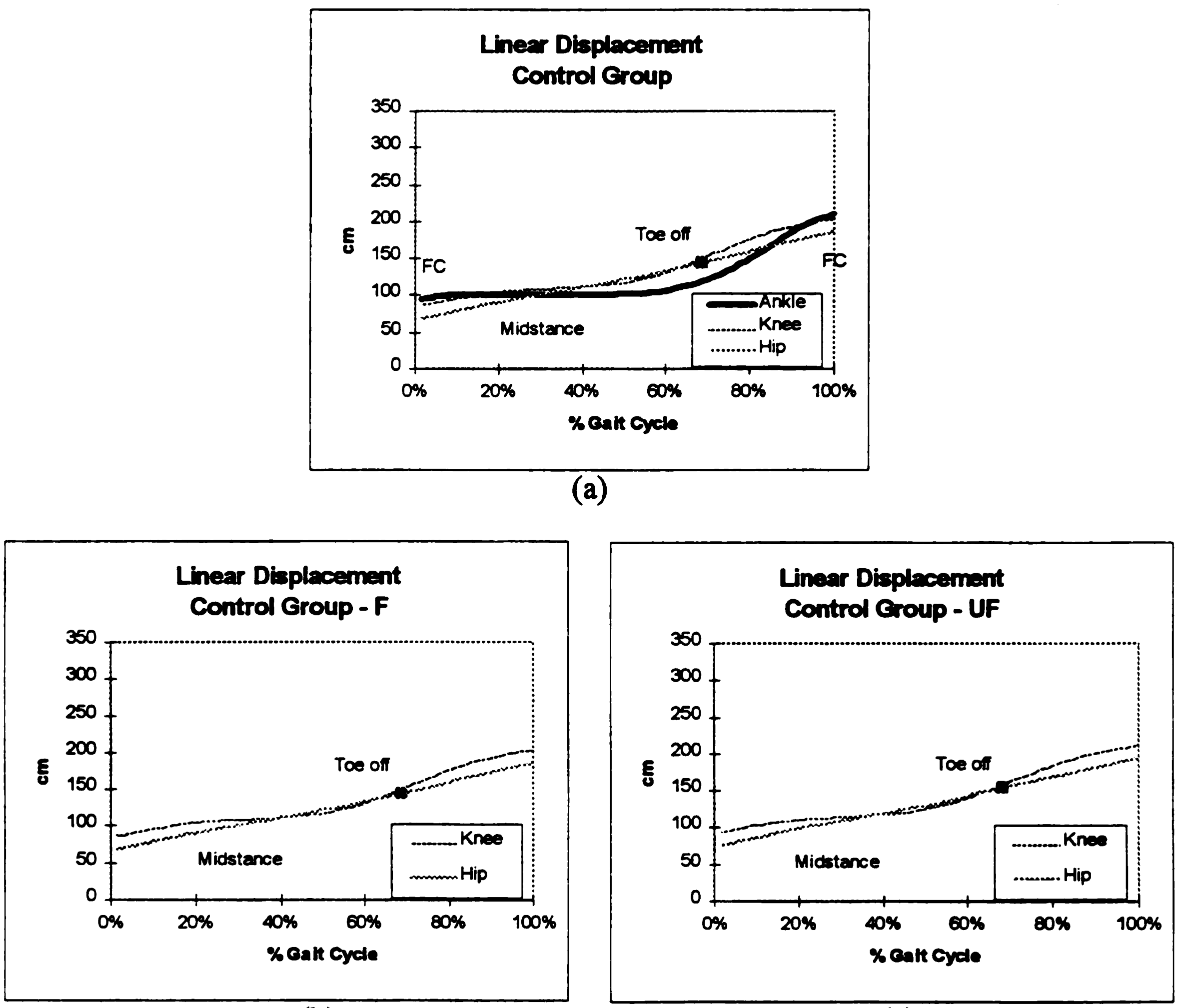


Figure 41: Linear Acceleration under the Unfatigued Condition. Control Group.

Under the fatigued condition, the linear displacements of the hip, knee and ankle joints followed their respective similar pattern as that pattern presented under the unfatigued condition. However, timing differences were found throughout the gait cycle under the fatigued condition. The linear displacements of the hip, knee and ankle joints are shown for the control group under both conditions in Figure 42. A greater range of



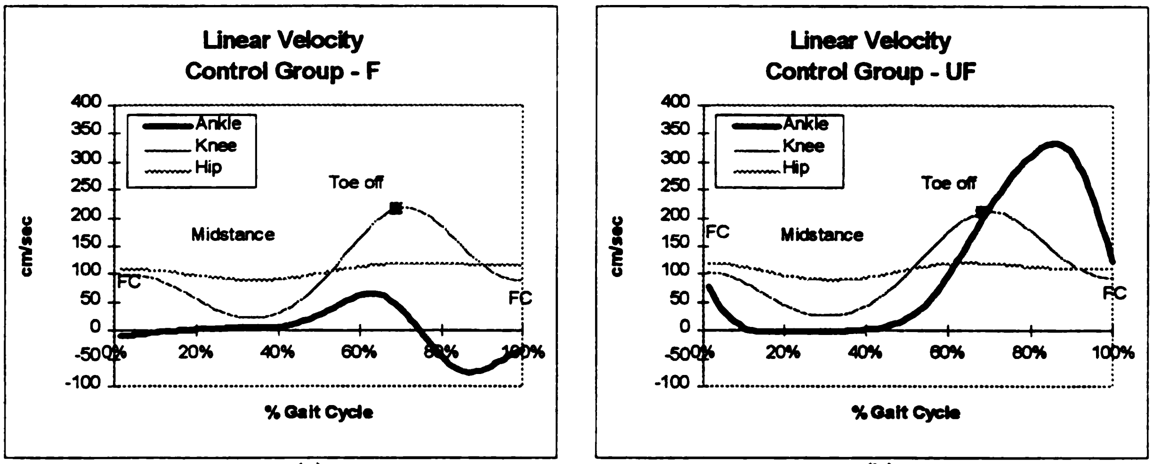
(a)

(b) (c)

Figure 42: Linear Displacement under the Fatigued Condition. (a) Control group (b) Knee-hip Comparison under the Fatigued Condition (c) Knee-hip Comparison under the Unfatigued Condition.

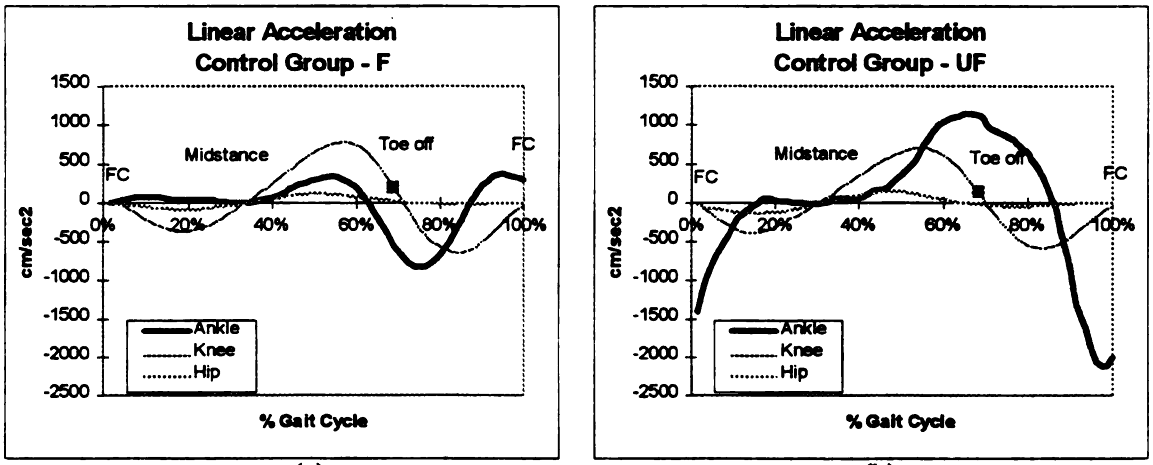
knee flexion/extension was demonstrated during the fatigued state when compared to the unfatigued condition. Under the fatigued condition, the first hip-knee alignment occurred at 42% of the gait cycle instead of 37% of the cycle. This delay was due to the longer period taken to flex and extend the knee joint. The second hip-knee alignment was observed at approximately the same time during the stance phase for both conditions, suggesting greater extension of the knee joint at toe off when comparing the fatigued to the unfatigued conditions.

The velocity and acceleration graphs for the fatigued condition are presented in Figures 43 and 44. Both the velocity and acceleration patterns showed similar characteristics at the knee and hip joints as those observed under the unfatigued state. During the fatigued condition the first hip-knee alignment occurred later during the stance phase when compared to the unfatigued state. This difference in timing was shown as a longer negative slope on the velocity graph under the fatigued condition. The knee was decelerated over a longer period of time due to the longer flexion/extension phase presented at the knee joint, from foot contact to 35% of the cycle. Since the second alignment observed for the hip and knee joints occurred at the same percent of the gait cycle under both conditions, the magnitude of the velocity graph increased. Under the fatigued condition, the knee joint had a shorter period in which to recover and align with the hip joint. Therefore, the velocity graph presented a positive slope reaching a greater magnitude. This faster rate of change of the displacement of the knee joint can also be observed on the acceleration graph. The knee was accelerated to a greater magnitude over a longer period of time during the fatigued condition, when compared to the unfatigued state.



(a) (b)

Figure 43: Linear Velocity under the Fatigued and Unfatigued Conditions. Control Group. (a) Fatigued (b) Unfatigued.



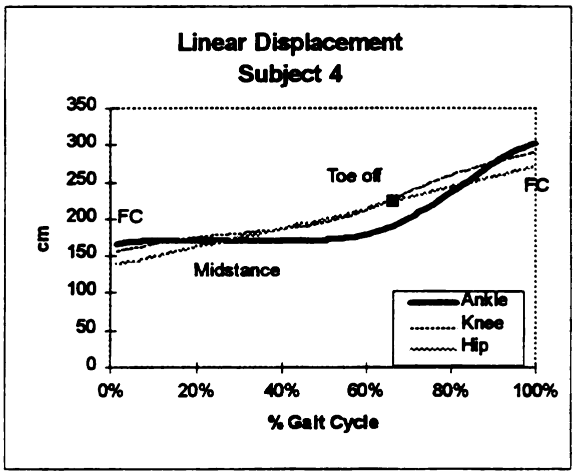
(a) (b)

Figure 44: Linear Acceleration under the Fatigued and Unfatigued Conditions. Control Group. (a) Fatigued (b) Unfatigued.

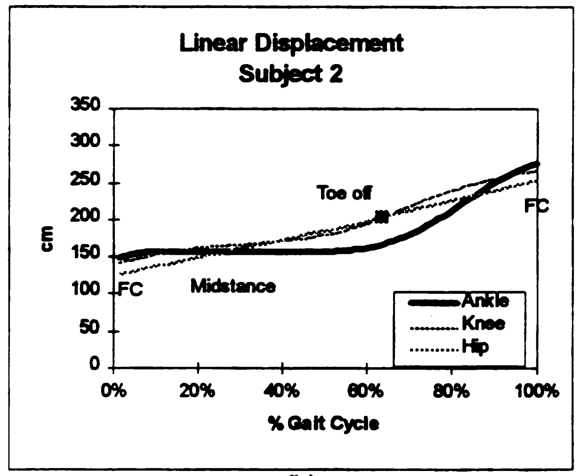
As illustrated in Figures 43 and 44, the ankle joint showed differences in both the velocity and acceleration patterns under the fatigued condition, when compared to the unfatigued condition. A relatively constant velocity of 0 cm/sec was presented for the initial 40% of the gait cycle. As the ankle joint prepared for toe off the velocity of the joint was increased. However, the magnitude of the increase was restricted during the fatigued condition when compared to the unfatigued state. During the fatigued condition the ankle joint demonstrated a negative slope in the velocity graph past the point of toe off. Under the unfatigued condition the changes in the direction of the slope did not occur until midswing. The acceleration graph for the control group under the fatigued condition, illustrates the reduced magnitude of the acceleration of the joint form 40% to 60% of the gait cycle. A deceleration period is presented through toe off, and at 75% of the gait cycle the joint accelerates to prepare the joint for foot contact.

C. ACL - Reconstructed Group

Linear displacement graphs for Subjects 4 and 5 under the unfatigued condition are presented in Figure 45. Due to the differences observed in the linear displacement graphs, data for both subjects will be presented and discussed. The knee angular characteristics presented by Subject 4 under the unfatigued condition were similar to those presented by Subject 2 under the same condition. (See Figures 45 a and b). As expected, the linear displacement graph for the hip, knee and ankle joints were similar for both subjects. During midstance the forward displacement of the knee joint was limited, while Subject 2 demonstrated normal initial flexion of the knee joint, extension of the joint during midstance was limited. At the point of maximum knee flexion (20% of the gait cycle) in the stance phase, the rate of change of the knee's displacement decreased with respect to the hip joint. Due to the limited extension of the knee joint during the stance phase, the joint maintained a slower rate of displacement to allow for the second hip-knee alignment. Timing of the rest of the alignments that occurred throughout the gait cycle were within the normal ranges of percent of gait cycle.



(a)



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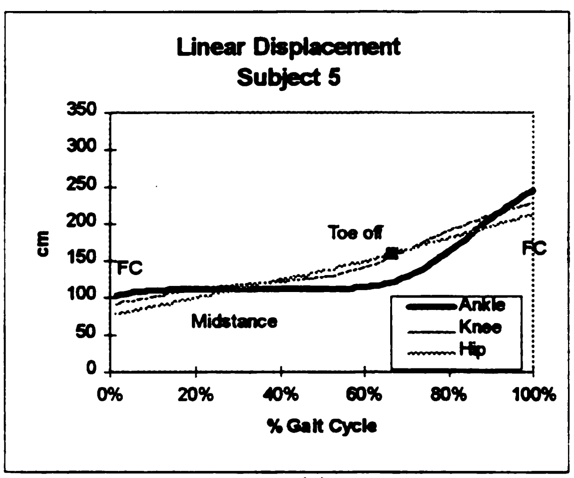
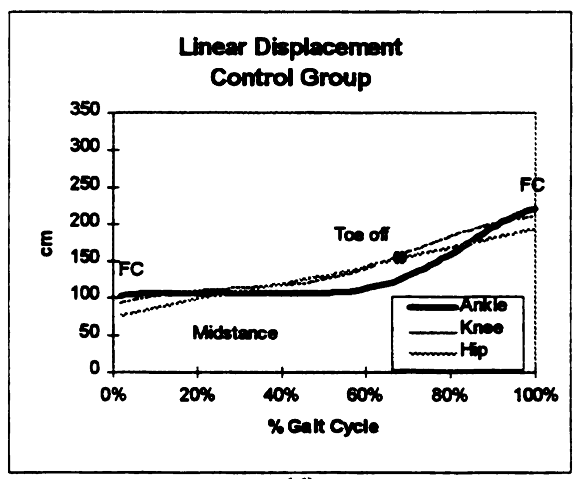


(c)



(d)

Figure 45: Linear Displacement under the Unfatigued Condition (a) Subject 4 (b) Subject 2 - Control Group (c) Subject 5 (d) Control Group.

Linear displacement patterns presented by Subject 5 under the unfatigued condition presented differences when compared to the control group patterns of the same condition (see Figures 45 c and d). Subject 5 showed limited flexion of the knee joint during the stance phase. In addition, the linear displacement characteristics had later alignments of the ankle-knee joints during the stance phase than the control group. The first ankle-knee alignment occurred at 21% of the gait cycle. This delay in the first alignment did not cause a delay for the rest of the alignments. The second ankle-hip alignment was observed at about the same period on the stance phase as the control group. A steeper slope was observed in displacement for the knee joint of Subject 5, when compared to the control group. Flexion of the knee joint aids in the forward advancement of the knee joint. The limited flexion experienced at the knee joint caused the displacement of the hip and the knee to be parallel for the initial portion of the stance phase. Flexion at the knee advanced the knee joint further in front of the hip and ankle. However, the time period during the stance phase between the ankle-hip alignment and the first hip-knee alignment was reduced as a result of the limited flexion. A longer percent of the cycle elapsed between the first and second hip-knee alignment. This time difference was due to the longer extension period demonstrated by the knee joint during the stance phase. The prolonged knee extension caused a delay in the initiation of the second knee flexion period, therefore, requiring more time for the knee joint to catch up with the hip. The pattern of the ankle joint displacement for Subject 5 was different to the pattern presented by the other subjects in the study. During the last 10% of the gait cycle the slope of the displacement graph was reduced for the subject in the control group and Subject 4 of the treatment group. In the case of Subject 5, the slope of the ankle joint displacement was maintained from the point of toe off to the completion of the cycle.

The linear velocity for the three joints under the unfatigued condition are presented in Figures 46. The linear velocity of the knee joint presented a negative slope following foot contact, for both subjects in the treatment group. Due to the restricted knee flexion experienced by Subject 5 the slope of the velocity graph for the knee was not as steep as the one presented by Subject 4. Subject 5 had a longer period of deceleration of the knee joint to 40% of the cycle, due to the reduced ability of the joint to decelerate the limb after foot contact. At 40% of the gait cycle the slope of the knee velocity became positive. The velocity of the knee joint increased until the point of the second hip-knee alignment. The velocity graphs of the hip and ankle joints were found to be similar for both subjects of the treatment group. The velocity of the ankle joint for Subject 5 showed a different pattern after midswing, than that presented by the other subjects. From midswing to the completion of the cycle, the velocity of the joint decreased to the same magnitude presented at foot contact, as presented by the control group and Subject 4. Subject 5 demonstrated a decrease in the velocity after midswing, however, the magnitude from toe off to second foot contact was only reduced by half the expected value. This difference in the velocity pattern was due to the observed rate of change of the displacement graph during the last 10% of the cycle.

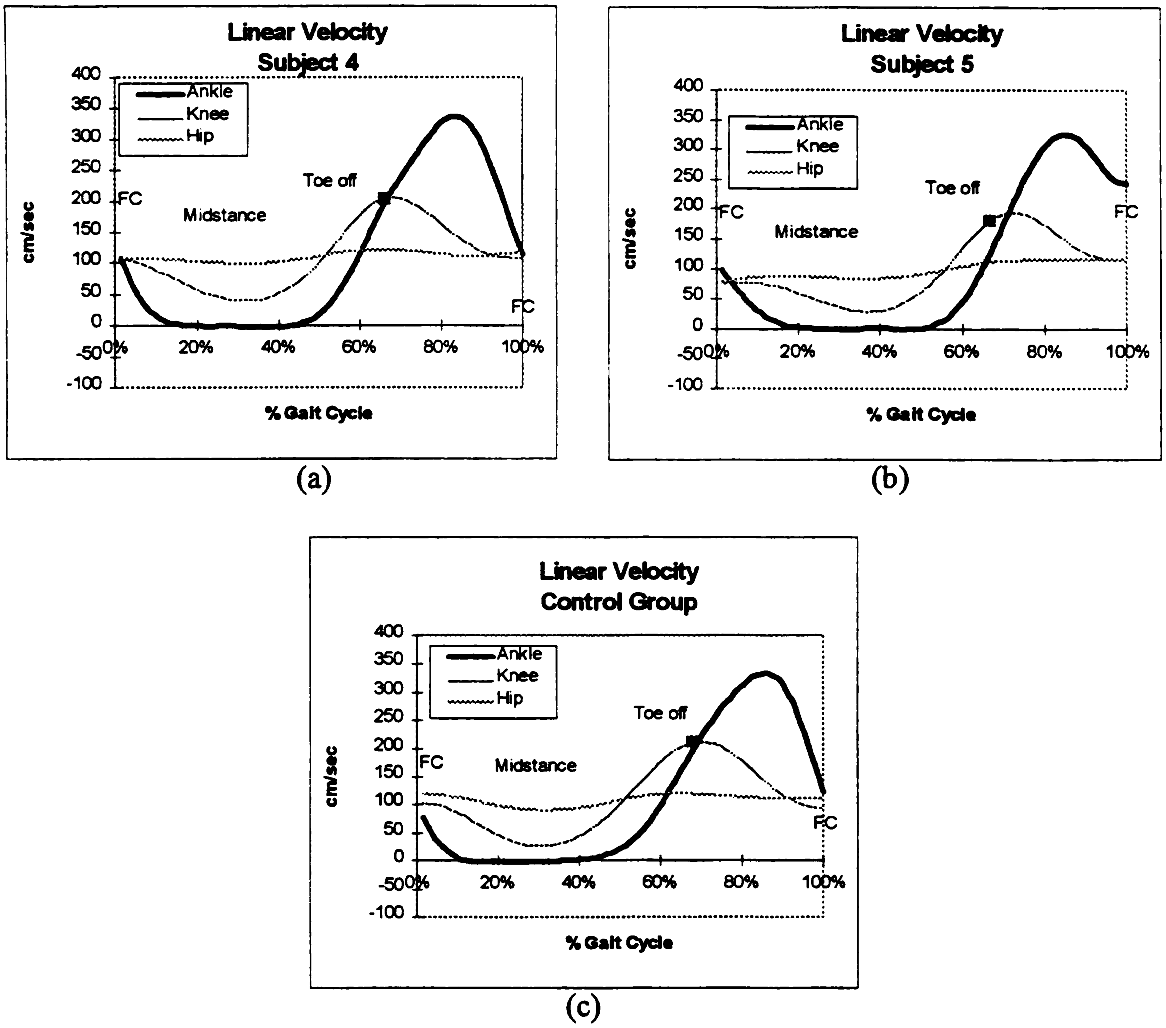
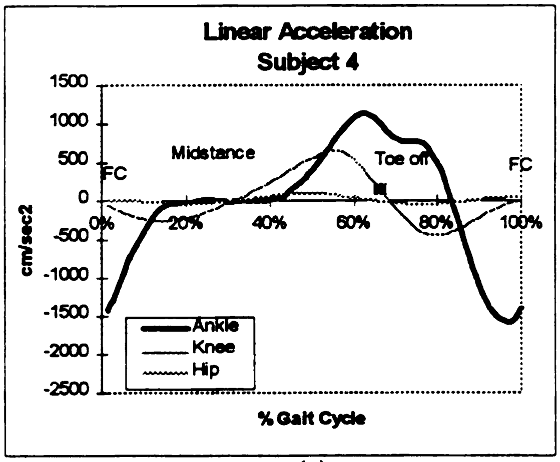
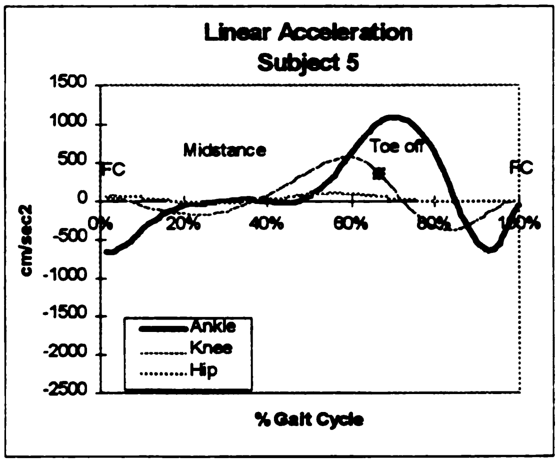


Figure 46: Linear Velocity under the Unfatigued Condition. (a) Subject 4 (b) Subject 5 (c) Control Group.

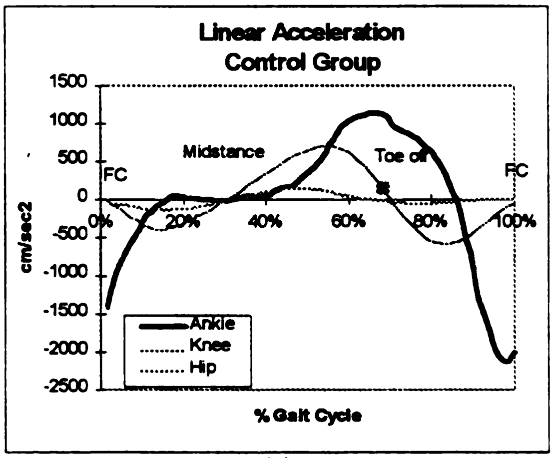
The acceleration graphs, illustrated in Figures 47 a and b, showed differences in magnitudes at the knee joint between Subjects 4 and 5. The acceleration and deceleration magnitudes were smaller for Subject 5. However, the acceleration magnitudes for Subject 4 were smaller than those presented by the control group. At the ankle joint, Subject 4 had a similar acceleration pattern as that demonstrated by the control group. After the point of toe off the slope of the acceleration graph for Subject 4 showed some differences when compared to the control group. From toe off to midswing, the slope the acceleration graph was 0 for Subject 4, indicating a constant rate of change in the velocity of the joint. Subject 4 also presented a smaller deceleration magnitude prior to the completion of the cycle. The acceleration magnitudes for the ankle joint for Subject 5 were smaller when compared to those presented by the rest of the subjects in the study. The acceleration/deceleration period was shorter for Subject 5, due to a later initiation of the forward displacement at the ankle joint.



(a)



(b)

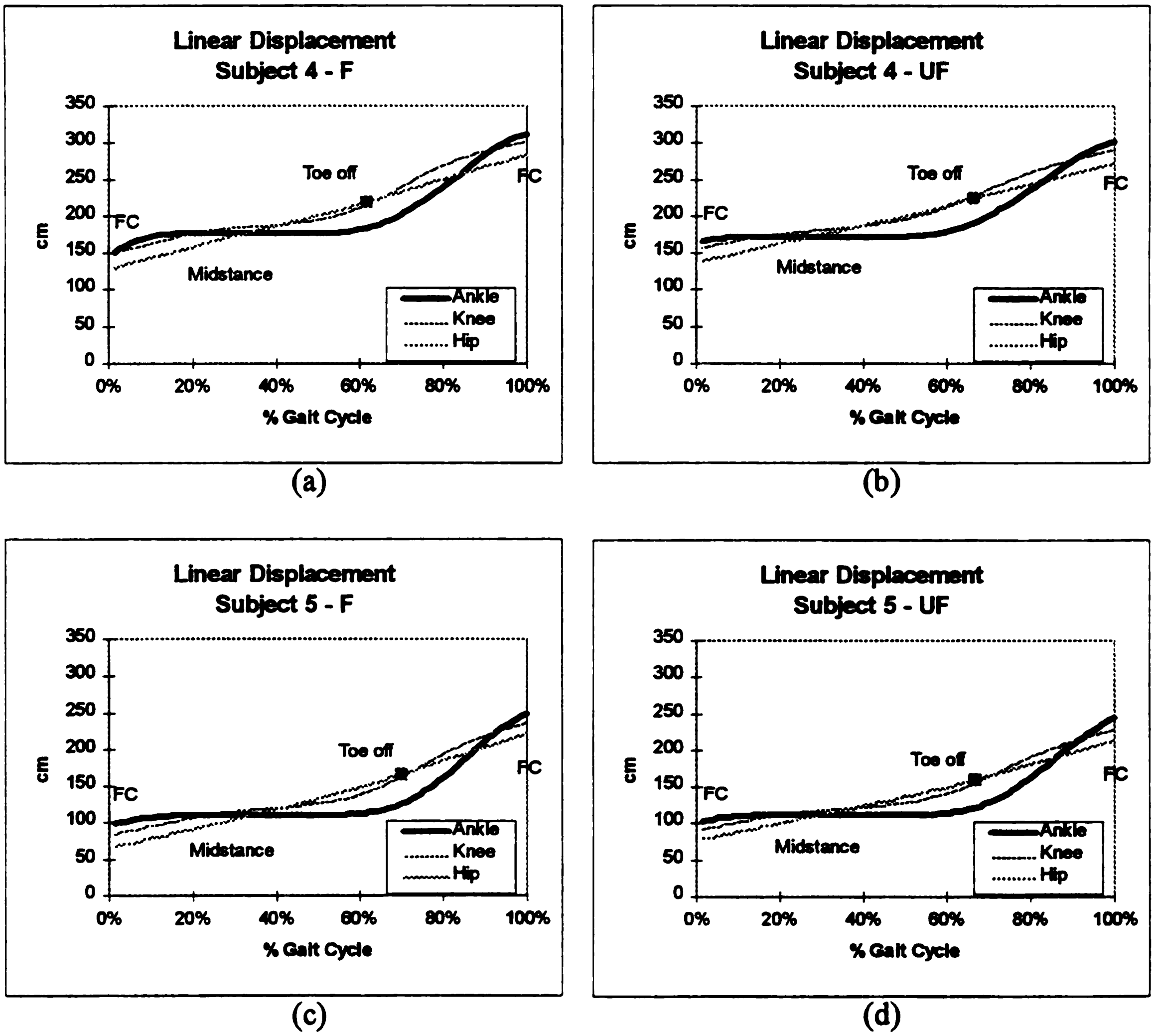


(c)

Figure 47: Linear Acceleration under the Unfatigued Condition. (a) Subject 4 (b) Subject 5 (c) Control Group.

Under the fatigued condition, shown in Figure 48, the linear displacements of the hip, knee, and ankle joints showed some temporal differences when compared to the unfatigued condition. Subject 4 had a later alignment than under the unfatigued condition between the ankle and knee joints occurring at approximately 20% of the gait cycle. In addition, a shorter period was observed between the first and second hip-knee alignment, when compared with the unfatigued condition. Subject 5 demonstrated an earlier ankle-knee alignment during the stance phase under the fatigued condition than the unfatigued condition, which led to a longer percent of cycle between the ankle-hip and the first hip-knee alignments. The slope of the knee's displacement graph was steeper for the fatigued condition when compared to the displacement under the unfatigued condition for Subject 5. This change in the slope characteristics was due to greater knee flexion experienced under the fatigued condition, allowing the knee joint to displace faster than the hip joint during the initial portion of the stance phase.

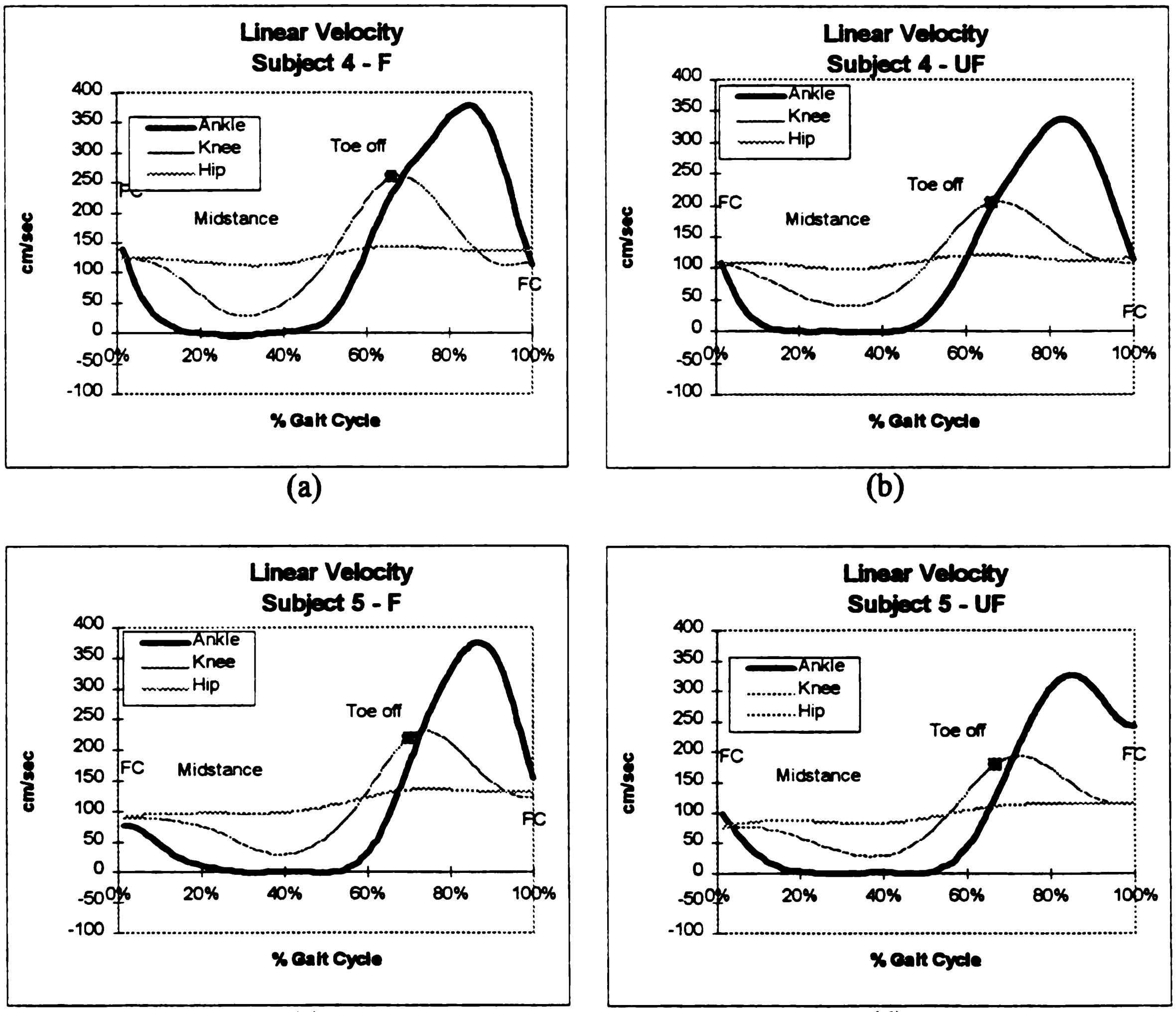
Linear velocity and acceleration graphs for the fatigued condition are presented for both subjects in the ACL reconstructed group. The velocity and acceleration graphs obtained under the fatigued condition were compared to the graphs for the unfatigued condition for the same subject. These patterns are illustrated in Figures 49 and 50, respectively. Subject 4 demonstrated a steeper negative velocity slope occurring from foot contact to approximately 30% of the gait cycle. This change in the slope of the velocity graph was due to a greater flexion range presented by the subject during the stance phase. The larger flexion at the knee joint led to a higher rate of change on the



(a) (b) (c) (d)

Figure 48: Linear Displacement under the Fatigued and Unfatigued Conditions. (a) Subject 4 - Fatigued (b) Subject 4 - Unfatigued (c) Subject 5 - Fatigued (d) Subject 5 - Unfatigued.

displacement of the joint. The magnitude of the knee's velocity displayed at toe off was greater under the fatigued condition than under the unfatigued condition. The ankle joint also displayed a steeper slope on the velocity graph during the swing phase for the fatigued condition when compared to the unfatigued state. This subject's acceleration graph illustrated the difference found in the slope of the velocity of the knee joint by a decrease in acceleration magnitudes presented at the knee joint. The ankle joint presented higher acceleration values under the fatigued condition than under the unfatigued condition. Additionally, the ankle joint had a more pronounced acceleration change from toe off to midswing of the fatigued condition, when compared to the unfatigued condition.



(a) (b) (c) (d)

Figure 49: Linear Velocity under the Fatigued and Unfatigued Conditions. (a) Subject 4 - Fatigued (b) Subject 4 - Unfatigued (c) Subject 5 - Fatigued (d) Subject 5 - Unfatigued.

The differences in the linear velocity between the two tested conditions were less pronounced for Subject 5. The only apparent difference presented in the knee velocity pattern was a greater velocity magnitude found at toe off during the fatigued state. The velocity pattern exhibited at the ankle joint was comparable to normal values, as opposed to the difference in magnitude presented under the unfatigued condition. The acceleration graph for the knee joint under both conditions presented similar patterns, for Subject 5. Under the fatigued condition, the slope of the acceleration of the knee joint from 30% to 60% of the cycle was steeper than under the unfatigued condition. This was due to the difference found on the magnitude of the velocity of the knee prior to the swing phase between the two conditions.

III. SUMMARY

Angular displacement data were presented only for the knee joint. The most relevant differences in this study were those experienced during the stance phase of the gait cycle. The control group demonstrated normal flexion/extension characteristics under the unfatigued condition. An exception to this finding was Subject 2 of this group, who showed limited extension during midstance. Under the fatigued condition, all subjects in the control group had a greater flexion/extension range during the stance phase, than under the unfatigued condition. The flexion/extension period lasted an additional 5% of the gait cycle longer when compared to the unfatigued condition. Increases in velocity and acceleration magnitudes over those in the unfatigued condition were present during the fatigued condition.

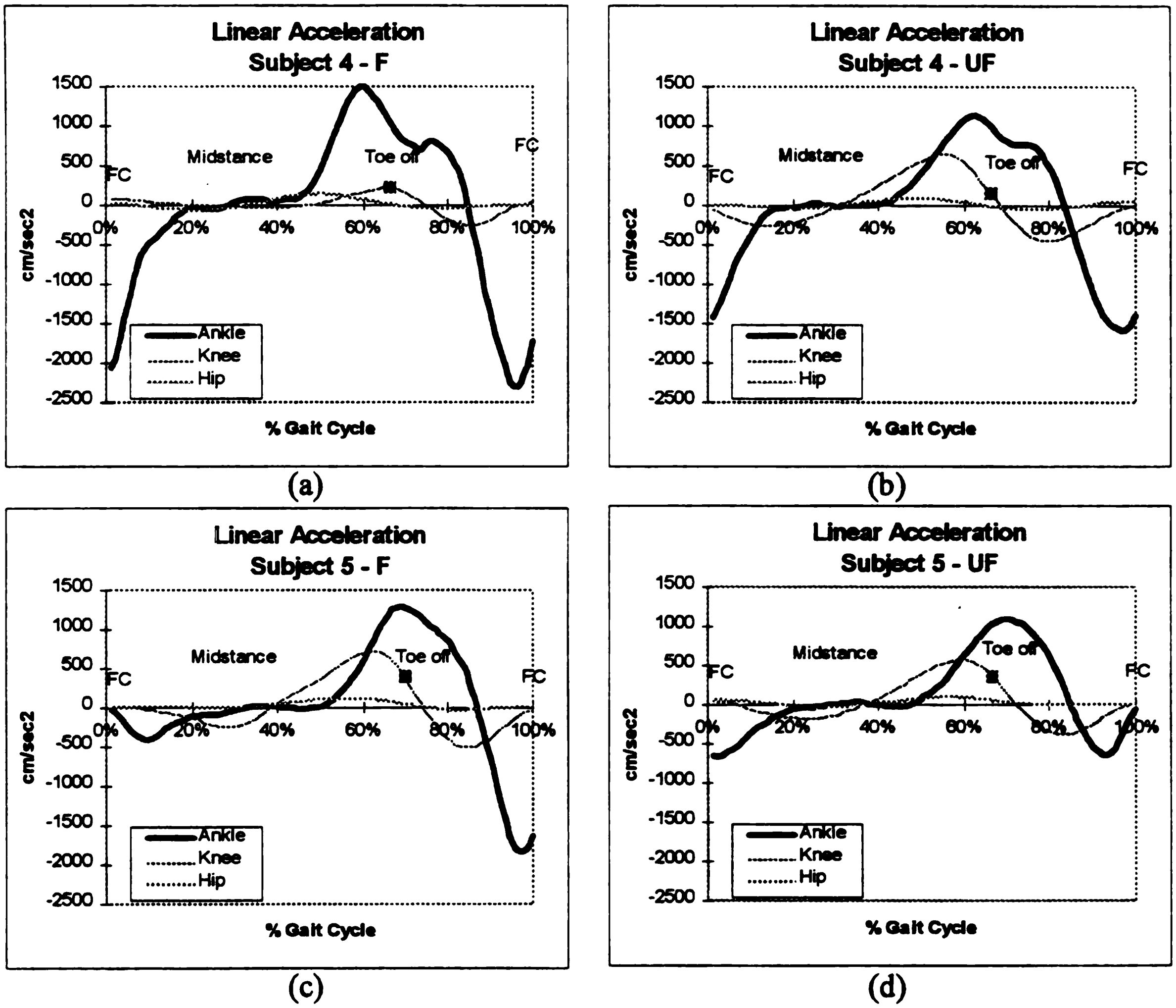


Figure 50: Linear Acceleration under the Fatigued and Unfatigued Conditions. (a) Subject 4 - Fatigued (b) Subject 4 - Unfatigued (c) Subject 5 - Fatigued (d) Subject 5 - Unfatigued.

The subjects forming the ACL-reconstruction group demonstrated adaptations in the flexion/extension angular displacements of the knee joint. Under the unfatigued condition, Subject 4 showed limited extension during midstance, while Subject 5 experienced limited flexion/extension during the entire stance phase. These gait adaptations indicated similar characteristics as those presented during quadriceps avoidance gait. With quadriceps avoidance gait, the subject limits the motion at the knee, as a protective mechanism for the injured joint. Due to the differences experienced in the angular displacement of the knee joint, the angular velocities and accelerations were of lower magnitudes than those magnitudes found for the control group. A delay in the flexion/extension pattern during the stance phase was also experienced by the treatment group when compared to the control group.

During the fatigued state, Subject 4 demonstrated a normal flexion/extension pattern during the stance phase. Subject 5 had a 5 degree increase in the flexion/extension range when compared to the unfatigued condition. However, the range of motion was lower than normal expected values. Under the fatigued condition, all subjects, except Subject 1, experienced an increase in flexion/extension magnitudes during the stance phase. The velocity and acceleration patterns, under the fatigued condition for Subject 4, were similar in magnitude and timing to the patterns found for the unfatigued condition for the control group. The flexion/extension pattern for Subject 5 was similar during both tested conditions, leading to similar magnitudes in the velocity and acceleration patterns under both conditions.

The results of the abduction/adduction angular displacements, velocities and accelerations indicated less consistency among the groups and conditions tested, than for flexion/extension. It is believed that this increase in variation was due to the small range of motion experienced by the knee joint movement in the frontal plane, in combination with the sagittal targeting. A wide range of abduction/adduction angular displacement was experienced by the subjects in the control group. However, the greatest differences between subjects were experienced during the swing phase. The velocity and acceleration patterns showed numerous magnitude changes throughout the stance phase. When comparing the unfatigued and fatigued conditions for the control group, the general patterns were found to be the same under both conditions. Adaptations experienced by the control group during the swing phase of the fatigued condition were shown by an increase in magnitude of maximum abduction. Differences in velocity and acceleration magnitudes were apparent when comparing both tested conditions. Although a similar discrepancy in patterns was presented by the two subjects forming the treatment group, the displacement patterns were more characteristic of displacements traditionally presented as normal. The ACL reconstructed group's changes in displacement for abduction/adduction were less in number and magnitude when compared to the unfatigued condition of the control group. As a result, both subjects experienced velocities and accelerations of lesser magnitude than the control group.

Under the fatigued condition, Subject 4 demonstrated the greatest change in pattern when compared to the unfatigued condition. Subject 4 had an increase in the maximum abduction value during the swing phase, demonstrating a similar range of motion as that presented by the control group under both conditions. Subject 5 maintained the same general pattern under both unfatigued and fatigued conditions. The velocity and acceleration patterns of Subject 5 indicated a more controlled movement during the fatigued state, due to the uniform motion characteristics presented under the fatigued condition.

The muscular activity characteristics found for both conditions differed for all the subjects tested. Under the unfatigued state, subjects in the control group showed a higher intensity of contraction for the quadriceps group than the treatment group. However, quadriceps firing patterns were different in timing for both tested subjects in this group. In addition, hamstrings activity was observed during the later portion of the stance phase, and the initial part of the swing phase for both subjects, instead of the expected hamstrings activity for the initial 15% of the cycle, and from 75% to completion of the cycle. Under the fatigued condition, the hamstrings demonstrated a higher intensity of contraction when compared to the contraction of the quadriceps, under the same condition. The pattern of activity again was different for both subjects. Under the fatigued condition, Subject 2 demonstrated hamstrings activity during the full gait cycle. This increase in muscle activity might be interpreted as a secondary mechanism for knee stabilization. Subject 4, one of the treatment group members, demonstrated a higher intensity of contraction for the hamstrings when compared to the quadriceps under the unfatigued state. The prolonged activity of the hamstrings over the stance phase under the fatigued condition indicated the subject's reliability on the hamstrings group to assist in knee stability. During the fatigued state, the hamstrings were active over the first half of the stance phase, and during the swing phase. However, the intensity of contraction demonstrated by the hamstrings was lower than that of the quadriceps group.

Characteristics presented in the linear motion of the lower extremity joints were similar for all subjects tested in the study. Adaptations in the linear displacements were directly related to adaptations observed in the angular motion. Limited extension at the knee joint during the stance phase was related to slower horizontal displacement of the knee joint, while differences in the slopes of the joint displacements were related to changes in the angular displacement patterns, under both tested conditions. The velocity and acceleration magnitudes were smaller for the treatment group than for the control group under the unfatigued condition. Finally, both groups demonstrated a decrease in the velocity and acceleration magnitudes during the fatigued state when compared to the unfatigued state.

Both groups in this study demonstrated similar gait patterns under both tested conditions. However, differences in range and timing were found to be characteristic adaptations under the fatigued state. Greater variability in the gait patterns seemed characteristic of the fatigued condition. Both groups experienced greater magnitudes in the angular velocity and acceleration values at the knee joint, under the fatigued condition than for the unfatigued condition. These higher velocity and acceleration magnitudes are known to increase the stresses on soft tissue in the joint. If the mechanics of the joint are not fully restored after an injury, these increases in magnitude can result in reinjury. Joint stability provided by muscular contraction was compromised during the fatigued condition, as was shown by changes in EMG patterns under a lower magnitude of contraction. Therefore, adaptations in the knee kinematics during gait under the fatigued condition are of greater concern for the injured subject than for the uninjured subjects, due to the restricted ability of the ligament to stabilize the knee joint

CHAPTER V

CONCLUSIONS

The role of the anterior cruciate ligament (ACL) as primary stabilizer of the knee joint has been recognized by several authors (Jackson et al., 1987; Schipplein et al., 1991). When an injury to the ACL occurs, the knee joint's stability is compromised. Strengthening of the thigh muscles has been shown to restore some level of stability to the ACL-deficient knee joint (Schipplein et al., 1991). Gait analysis has been widely used in an attempt to recognize adaptations of gait pattern caused by injuries to the ACL (Andriacchi, 1990; Andriacchi, 1993; Timoney et al., 1993; Berchuck et al., 1990). Quadriceps avoidance gait has been identified in subjects with ACL deficient and reconstructed knees (Berchuck et al., 1990). This adaptation is characterized by a reduction in the knee flexion angle during the initial 20% of the gait cycle. Although kinematic and electromyographic research has been conducted to analyze the muscular activity patterns presented by individuals with ACL reconstruction during walking (Shiavi et al., 1992), there have been no studies reported to analyze the influence of lower extremity muscular fatigue on gait characteristics of ACL reconstructed, or ACL deficient individuals. It is a widely accepted fact that muscle fatigue can jeopardize the level of joint stability during movement (Dul et al.,1984).

Research on the effects of muscular fatigue during gait for individuals with ACL reconstruction is necessary to understand the implications for athletes' reintegration into practice. The effects of muscle fatigue on knee joint stability may be detrimental to athletes returning to practice and competition, prior to full restoration of normal knee mechanics. Determination of adaptations in gait characteristics can be useful to the athletic trainers, athletes, and coaches in the decision making process of when to return the athlete to the field.

A three-dimensional analysis of gait characteristics of individuals with ACL reconstruction and uninjured individuals was performed under unfatigued and fatigued conditions. The purpose of the study was to determine what, if any, changes in gait characteristics occurred due to lower extremity muscular fatigue. In addition, EMG was used to investigate the relative sequencing of firing of the quadriceps and hamstrings muscle groups. No asymmetries were found in bilateral segment lengths for each subject, when measured anthropometrically. Cybex results for the ACL reconstructed group showed the accepted three to two ratio of the hamstrings to quadriceps muscle strength. In addition, strength differences lower than 15 percent between left and right sides for both quadriceps and hamstrings groups were found. This is the maximum strength difference accepted by the NATA prior to reintegration of the athlete to practice. All subjects had laxity tests within the +/- 1 normal range. A control group was tested under unfatigued and fatigued conditions to determine if possible differences presented by the ACL reconstruction group were also exhibited by uninjured individuals.

The three-dimensional analysis examined the knee joint's linear and angular motion during a full gait cycle. Kinematic data were collected on three control subjects, and two subjects with bilateral ACL reconstruction. All subjects performed three trials under unfatigued and fatigued conditions. A bicycle ergometer submaximal test was utilized to fatigue each subject. A 75 watts work load was used initially; with a 25 watts increaes every 2 minutes. The Borg scale was used to rate the level of fatigue every 2 minutes. After each walking trial, the subjects were asked to rank their level of fatigue on the Borg scale. If the results dropped more than two points on the scale the subjects were asked to repeat the bicycle ergometer test. Two trials under each condition were digitized as part of the analysis. The calculations of linear and angular kinematics were performed only on the side of the most recent injury for the treatment group, and the side closer to the cameras for the control group. Linear and angular parameters were described for each group under the unfatigued and fatigued conditions. The results of the two groups were summarized to identify any trends observed during the analysis.

I. LIMITATIONS

The following limitations of the study must be taken into consideration before generalizing the results. First, the low number of subjects and the characteristics of the data collected did not allow for statistical comparisons of the results. The results of the study were reported in a descriptive manner, without statistical conclusions. Second, some limitations were experienced due to the inability of the software to accurately calculate transverse rotations at the knee joint. Internal/external rotation data were not reported due to the magnitude of the error presented in the results. The kinematic analysis was performed only on flexion/extension and abduction/adduction movements. Third, during the test under the fatigued condition, the same level of fatigued was assumed throughout the three trials. The maintenance of fatigue was difficult to regulate, as each subject was asked to perform three trials. However, the level of muscle fatigue might have been jeopardized by unexpected delays, such as markers falling off, or system failures. The use of the Borg scale assisted in identifying the perceived fatigue of each subject. A better system to quantify muscular fatigue might be needed if the same level of fatigue is to be maintained throughout trials.

II. HYPOTHESES

Three hypotheses were tested for this study. The first hypothesis stated that individuals with ACL reconstruction would exhibit greater muscular activity during the stance phase under unfatigued conditions than the normal subjects. Because of problems experienced with EMG data collection for Subject 5, the examination of this hypothesis was based solely on the results presented by Subject 4 and is therefore limited in scope. During the stance phase of the unfatigued condition, Subject 4 presented greater muscular activity than the subjects in the control group. The hamstrings and quadriceps were active over a longer period of the stance phase, when compared to the muscle activity presented by both the control group and the expected normal under the unfatigued condition. During the unfatigued condition, Subject 4 showed hamstrings activity from foot contact to 55% of the cycle, in the first trial; and from 10% to 90% of the gait cycle during the second trial. Normal muscular activity by the hamstrings is expected for the initial 15% of the cycle, and from 75% to completion of the cycle. The quadriceps were active from 10% to 55% of stance and then again from 60% to completion of the cycle for the first trial of Subject 4. During the second trial, the quadriceps were active for the initial 30% of the gait cycle. The expected activity pattern of the quadriceps during the stance phase, has muscle activity for the initial 25% of the cycle, from 50% to 60% of the cycle, and then again for the last 5% of the cycle. The results of this study supported the first hypothesis.

The second hypothesis stated that individuals in the ACL-reconstruction group would demonstrate lower linear and angular parameters at the knee joint during the fatigued state, when compared to the unfatigued condition. Linear and angular displacements, velocities, and accelerations were shown to be lower in magnitude for portions of the stance and swing phases of the gait cycle under the fatigued condition. Angular velocities and accelerations for flexion/extension under the fatigued condition, were higher than those presented under the unfatigued condition. Angular velocities and acceleration for abduction/adduction were lower for both subjects in the treatment group under the fatigued condition, when compared to the unfatigued condition. The differences found in the gait patterns were not due solely to overall lower magnitudes presented at the knee joint. Therefore, this hypothesis was not supported by the results of the study.

The third and last hypothesis was not supported by the subjects forming the ACL-reconstruction group. The hypothesis stated that the subjects with ACL reconstruction would experience adaptations of greater magnitudes at the knee joint under the fatigued state than the control group. Changes in gait characteristics were found for all subjects in the study. Adaptations presented by the subjects in the treatment group were of the same magnitude as those presented by the control group, between the two tested conditions. A greater flexion/extension range of motion at the knee joint was seen under the fatigued condition for the treatment group. However, all subjects in the study demonstrated the same relative change in range of motion. Adaptations seen under the fatigued condition in abduction/adduction angular displacement were different for all subjects. Due to the variability in adaptations presented by all the subjects, the magnitude of changes cannot be quantified in group comparisons.

Timing differences in linear displacements were experienced by both groups under the fatigued condition, when compared to their respective unfatigued condition. Adaptations presented under the fatigued condition by the treatment group, were comparable in magnitude to those experienced by the control group. Both groups showed similar displacement patterns under both conditions.

III. SUGGESTIONS FOR FUTURE STUDIES

Further research in this area is recommended. Recommendations for future studies will be made based on necessary changes in order to obtain better quality of results. Due to the system's limitations, the subjects should be targeted using a triad system (see Figure 51). This targeting style would allow the system to identify a coordinate system about the joint, and therefore be able to calculate rotations about the three axes of rotations. A primarily sagittal targeting style only allows for calculations of rotations in the sagittal and frontal planes.

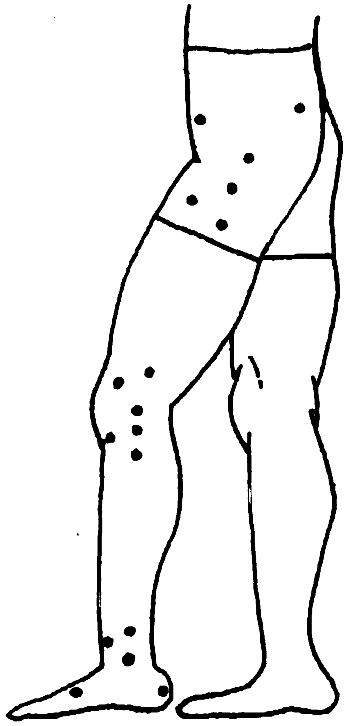


Figure 51: Placement of Targets Using Triads.

The problem of target blocking might be reduced by adding a third camera on the same side as the other two cameras (see Figure 52). Accurate digitization of the hip target was difficult due to consistent target blocking caused by the arm swing during the trials. Adding a third camera might increase the number of frames in which the hip target is seen by at least two cameras, and therefore increase the accuracy of digitization of this target.

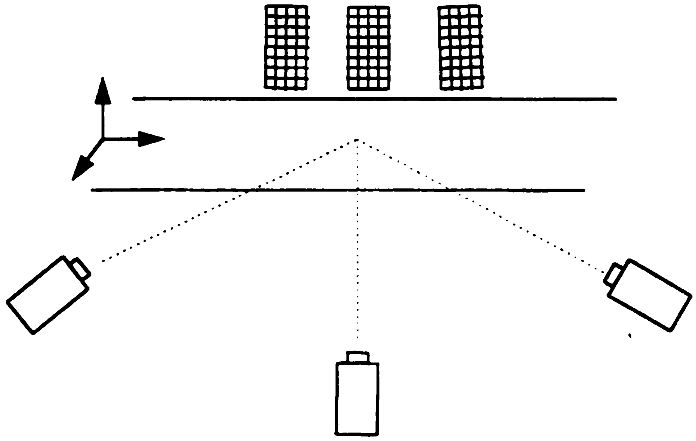


Figure 52: Suggested Testing Setting Using Three Cameras

Range of motion of both injured and uninjured knees should be measured as part of the pre-test administered to all subjects. This measurement would allow detection of limitations in range of motion that later might be exhibited during gait. Normal range of motion and strength testing results would allow the investigator to discard these two factors as the causes of any adaptations seen in gait.

The maintenance of fatigue was difficult to ensure, since the subjects were asked to perform three walking trials. To assist in maintaining fatigue, the Borg scale was used. A different fatiguing protocol might be used to more closely resemble the type of muscle fatigue experienced after an athletic practice. Bilateral muscular activity, as well as kinematic data, might be needed to understand any compensatory actions that might occur due to the fatiguing of the muscles of the lower extremity. Forces and moments are important kinetic parameters for determining gait abnormalities. A complete study which includes kinematic, kinetic, and electromyographic data might be of greater value to determine the overall changes that occur due to lower extremity muscular fatigue.

LIST OF REFERENCES

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Andriacchi, T. (1990). Dynamics of pathological motion: applied to the anterior cruciate deficient knee. Journal of Biomechanics, 23, 99-105.

Andriacchi, T. (1993). Functional analysis of pre and post-knee surgery: Total knee arthroplasty and ACL reconstruction. Journal of Biomechanical Engineering, 115, 575-581.

Angeli, C. (1995). User's guide to the Ariel Performance Analysis System. Unpublished manuscript.

Arms, S., Pope, M., Johnson, R., Fischer, R., Arvidsson, I., & Eriksson, E. (1984). The biomechanics of ACL rehabilitation and reconstruction. The American Journal of Sport Medicine, 12, 8-18.

Aune, A., Nordsletten, L., Skjeldal, S., Madsen, J., & Ekeland, A. (1995). Hamstrings and gastrocnemious co-contraction protects the anterior cruciate ligament against failure: An in vivo study in the rat. Journal of Orthopaedic Research, 13, 147-150.

Basmajian, J., & DeLuca, J. (1985). Muscles Alive. Their Functions Revealed by Electromyography. Baltimore: The Williams and Wilkins Company.

Berchuck, M., Andriacchi, T., Bach, B., & Reider, B. (1990). Gait adaptations be patients who have a deficient anterior cruciate ligament. The Journal of Bone and Joint Surgery, 72-A, 871-877.

Borg, G. (1982). Psychophysical bases of perceived exertion. Medicine and Science in Sports and Exercise, 14, 377-381.

Ciccotti, M., Kerlan, R., Perry, J., & Pink, M. (1994a). An electromyographic analysis of the knee during functional activities. I. The normal profile. The American Journal of Sports Medicine, 22, 645-650.

Ciccotti, M., Kerlan, R., Perry, J., Pink, M. (1994b). An electromyographic analysis of the knee during functional activities. II. The anterior cruciate ligament-deficient and reconstructed profiles. The American Journal of Sports Medicine, 22, 651-658.

Draganich, L., Jaeger, R., & Kralj, R. (1989). Coactivation of the hamstrings and quadriceps during extension of the knee. Journal of Bone and Joint Surgery, 71-A, 1075-1081.

Dul, J., Johnson, G., Shiavi, R., & Townsend, M. (1984). Muscular synergism - II. A minimum-fatigue criterion for load sharing between synergistic muscles. Journal of Biomechanics, 17, 675-684.

Feagin, J. (1988). The crucial ligaments. New York: Churchill Livingstone.

Gerdle, B., & Karlsson, S. (1994). The Mean frequency of the EMG of the knee extensors is torque dependent both in the unfatigue and the fatigued states. Clinical Physiology, 14, 419-432.

Gray, H. (1974). Gray's anatomy. Philadelphia: Running Press.

Guidelines for exercise testing and prescription. American College of Sports Medicine. (4th edition). (1991). Philadelphia, PA: Lea & Febiger.

Harris, G., & Wertsch, J. (1994). Procedures for gait analysis. Archives of Physical Medical Rehabilitation, 75, 216-225.

Hirokawa, S., Solomonow, M., Lu, Y., Lou, Z., & D'Ambrosia, R. (1992). Anterior-posterior and rotational displacement of the tibia elicited by quadriceps contraction. The American Journal of Sports Medicine, 20, 299-306.

Jackson, D. & Drez, D. (1987). The anterior cruciate deficient knee. New concepts in ligament repair. St. Louis: Mosby Company.

Lohman, T., Roche, A., & Martorell, R. (1988). Anthropometric standardization reference manual. Champaign, IL: Human Kinetics Books.

Macnicol, M. (1986). The problem knee - Diagnosis and management in the younger patient. Rockville: Aspen Publishers.

Maquet, P. (1984). Biomechanics of the knee. Berlin: Springer-Verlag.

McNair, P., & Wood, G. (1993). Frequency analysis of the EMG from the quadriceps of anterior cruciate ligament deficient individuals. Electromyography and Clinical Neurophysiology, 33, 43-48.

Nicholas, J. & Hershman, E. (1995). The lower extremity and spine in sports medicine Vol. 2. St. Louis: Mosby.

Noyes, F., Torvik, P., Hyde, W., & DeLucas, J. (1976). Biomechanics of ligament failure. The Journal of Bone and Joint Surgery, 56-A, 1406-1418.

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

Pecina, M. & Bojanic, I. (1993). Overuse injuries of the musculoskeletal system. Boca Raton: CRC Press.

Peltier, R. (1985). Effects of semi-rigid orthotics on abduction-adduction moments at the knee joint during walking. Unpublished master's thesis, Michigan State University, Lansing.

Renstrom, P., Arms, S., Stanwyck, T., Johnson, R., & Pope, M. (1986). Strain within the anterior cruciate ligament during hamstring and quadriceps activity. The American Journal of Sports Medicine, 14, 83-87.

Schipplein, O., & Andriacchi, T. (1991). Interaction between active and passive knee stabilizers during level walking. Journal of Orthopaedic Research, 9, 113-119.

Shiavi R., Zhang L., Limbird T., & Edmondstone M. (1992). Pattern analysis of electromyographic linear envelopes exhibited by subjects with uninjured and injured knees during free and fast walking. Journal of Orthopaedic Research, 10, 226-236.

Solomonow, M., Baratta, R., Zhou, H., Shoji, H., Bose, W., Beck, C., & D'Ambrosia, R. (1987) The synergistic action of the anterior cruciate ligament and thigh muscles in maintaining joint stability. The American Journal of Sports Medicine, 15, 207-213.

Tibone, J., Antich, T., Fanton, G., Moynes, D., & Perry, J. (1986). Functional analysis of anterior cruciate ligament instability. The American Journal of Sports Medicine, 14, 276-284.

Timoney, J., Inman, W., Quesada, P., Sharkey, P., Barrack, R., Skinner, H., & Alexander, H. (1993). Return to normal gait patterns after anterior cruciate ligament reconstruction. The American Journal of Sports Medicine, 21, 887-889.

Valmassy, R. (1995). Clinical biomechanics of the lower extremities. Boston: Mosby.

Winter, D. (1980). Overall principle of lower limb support during stance phase of gait. Journal of Biomechanics, 13, 923-927.

Yack, H., Riley, L., Whieldon, T. (1994). Anterior tibial translation during progressive loading of the ACL-deficient knee during weight-bearing and nonweight-bearing isometric exercise. Journal of Orthopaedic Sports Physical Therapy, 20, 247-253.

Zatsiorky, V., Werner, S., & Kaimin, M. (1994). Basic kinematics of walking. The Journal of Sports Medicine and Physical Fitness, 34, 109-131.

APPENDICES

APPENDIX A

APPENDIX A1

Informed Consent Form

Subject's Name: ____________________________________

The present study will consist of a biomechanical evaluation of gait characteristics of a group of individuals without injury to the lower extremity, and a group with anterior cruciate ligament reconstruction. On the test day prior to the initiation of testing, anthropometric measurements of the lower extremity will be taken. Knee joint laxity will be tested by a certified athletic trainer, the Lachman test will be used for this purpose. I will be asked to complete an information sheet. I understand that both groups will follow the same testing protocol. I understand that I will be asked to walk along a walkway while being filmed, and electromyographic activity (muscle activity) will be recorded. Following a set of three trials, I will be asked to ride a bicycle ergometer until the onset of muscle fatigue as determined by the Borg scale and EMG activity. Another set of three trials will be filmed, after the onset of fatigue.

I understand that all testing will only require a single day commitment of approximately three hours. I understand that in case of equipment failure I may be asked to return another day to complete testing.

I understand that participation in this study is voluntary, and that I am free to withdraw during any portion of the study without penalty. I understand all results will be treated with strictest confidence and I will remain anonymous in any report of research findings; on request and within these restrictions results may be made available to me.

I understand that if I am injured as a result of my participation in this research project, Michigan State University will provide emergency medical care if necessary. I further understand that if the injury is not caused by the negligence of MSU, I am personally responsible for the expense of this emergency care and any other medical expenses incurred as a result of this injury.

Signed: ____________________________ Date: __________________

Address: ______________________________

Telephone Number: (   ) _____________

PRINCIPAL INVESTIGATOR: Claudia A. Angeli
(517) 353-0728 / (517) 351-7124

FACULTY ADVISOR: V. Dianne Ulibarri, PhD.

APPENDIX A2

Informed Consent Form

Subject's Name: ___________________________________

I understand that I am being considered as a subject in the anterior cruciate ligament reconstruction group for a biomechanical evaluation of gait characteristics of a group of individuals without injury to the lower extremity, and a group with anterior cruciate ligament reconstruction. I understand that I will need to be Cybex tested. The cybex test is for subject recruitment purposes only. If selected as a subject I understand that I will need to return on a designated day to participate in the study which has been explained to me. At the time, I will sign a second consent form, dealing specifically with the testing protocol.

I understand that participation in this study is voluntary, and that I am free to withdraw during any portion of the study without penalty. I understand all results will be treated with strictest confidence and I will remain anonymous in any report of research findings; on request and within these restrictions results may be made available to me.

I understand that if I am injured as a result of my participation in this research project, Michigan State University will provide emergency medical care if necessary. I further understand that if the injury is not caused by the negligence of MSU, I am personally responsible for the expense of this emergency care and any other medical expenses incurred as a result of this injury.

Signed: __________________________ Date: _________________

Address: ____________________________

Telephone Number: (   ) ____________

PRINCIPAL INVESTIGATOR: Claudia A. Angeli
(517) 353-0728 / (517) 351-7124

FACULTY ADVISOR: V. Dianne Ulibarri, PhD.

APPENDIX B

APPENDIX B

Anthropometric Data Sheet

Subject's Name: ______________________________ Code: ________

Group: _______________

Dominant side: Right Left (kicking leg) Measurements: cm.

in.

Standing Height: ________________ Weight: ____________ kg lb

Measurement	Right	Left
Leg length		
Upper thigh girth		
Calf girth		
Femur breadth		

Recorder: ____________________

Measured by: ___________________

APPENDIX C

APPENDIX C

Fatiguing Protocol

The submaximal bicycle ergometer test described in the American College of Sports Medicine guidelines (1991) for exercise testing and prescription were followed.

The test consisted of two minute stages, in which resistance was increased by 25 watts. The revolutions per minute (RPM) was set at 75. The resistance continued to be increased until the onset of fatigue.

STAGE I: minutes 1-2

Work load in watts: 75 | Work load in kg/m/min.: 450

STAGE II: minutes 3-4

Work load in watts: 100 | Work load in kg/m/min: 600

STAGE III: minutes 5-6

Work load in watts:125 | Work load in kg/m/min.: 750

STAGE IV: minutes 7-8

Work load in watts: 150 | Work load in kg/m/min.: 900

The subjects were asked to concentrate on only rating their feeling of muscular fatigue and were told that 10 in the rating of perceived exertion (RPE) scale meant extreme muscular fatigue. Testing was stopped after the onset of fatigue and a perceived exertion and level of fatigue of 8 in the Borg scale.

The RPE scale described in Borg (1982) was used in this study.

Category RPE Scale

0	Nothing at all
0.5	Very, very weak
1	Very weak
2	Weak
3	Moderate
4	Somewhat strong
5	Strong
6	
7	Very strong
8	
9	
10	Very, very strong
*	Maximal

APPENDIX D

APPENDIX D

User's Guide to the Ariel Performance Analysis System

The grabbing and digitizing sections of APAS user's guide (Angeli, 1995), will be presented here.

GETTING STARTED

APAS Main Menu:
* Grab
* Digitize
* Transform
* Smooth
* View
* Graph
* Print
* Kinetics
* Analog
* Tree

A. Grab

(directory -- OK)

FRAME_GRABBING MAIN MENU						
Filename	Capture	Restore	Delete	VCR	TimeCodes	Exit
(1)	(2)	(3)	(4)	(5)	(6)	(7)

(1) *New* Old Close Quit

Root filename: _________________

Enter a name which will help you identify the trial and the camera. Each videotape (camera view) for each trial will require a different filename.

The top part of the screen will display the filename, plus other information. From this information only one number will be important to you at this time. The ***image capacity (frames)*** is the available space in memory to store images. In the next step you will go over the videotape and pick the images that you want to grab, and later digitize. Check that this number does no exceed the memory limit, take into consideration the skip factor, (explained later). If the number of images you plan to grab, is over the memory capacity, ***stop and call for help!***

-- <u>Note</u>: the available memory to store images will decrease each time images in a new file are grabbed.

(5) Play Stop Rew. Fwd. Pause

For this step is more convenient to use the VCR directly. For the slow motion control, you need to press play and search at the same time. When the white dash mark in the knob is on still the image will freeze.

Find the first frame of movement or the point which is going to determine the synchronization of both cameras. This is going to be critical during the transformation process to assure the correct correlation of the images. Once you find that point ***reset the counter***, which appears on the top of the screen, by pressing **F3.**

Move forward, using the search knob, until the "end" of the activity, (you decide when you have seen enough). The field counter is going to indicate the number of frames for the activity. <u>Remember this number,</u> it is going to become very important for future use. I recommend you write this number down, along with the filename, (take it from experience). Appendix A contains an example of a data sheet that can be used.

<u>Rewind</u> the tape so the counter is at 0 again, (first frame), use the searching knob.

F10 -- will take you back to the main menu.

(2) Capture menu

Control	Data	VCR	Params	Toggle	Confirm	Quit
(a)	(b)	(c)	(d)	(e)	(f)	(g)

(d) Params menu

ENTER PARAMETERS FOR GRABBING IMAGES		
File description: ________________________		VCR: Auto
First image offset: 0	# saved frames: ____	Skip factor: ___

Under *file description* enter useful information such as, subject's name, camera ID, trial number, etc. This information will appear on top of the screen during the digitizing process, allowing you to check that you have called the correct file.

Leave the *VCR* option in Auto. During the grabbing process, the computer will control the VCR, in terms of the frames to be saved. If the option is changed to manual, you will have to select frame by frame the ones to be saved.

The next entry, *first image offset,* is one that you do not have to worry about. Of course the key to this option was rewinding the tape back to the 0 point after counting the frames.

# of saved frames refers to the number of frames that you want to save in memory. Here is when the number in the field counter, that hopefully you wrote down (and you can find that paper), becomes important. To calculate the number of saved frames, first you need to decide if you want to grab all the frames in the

sequence from the starting point to the end point. This is going to depend on the speed of the activity. The number you need to input under this option is going to be equal to the number of frames (the number you wrote down), divided by the skip factor + 1.

The skip factor is equal to the number you want to skip in between grabbed frames, (very self explanatory). As mention above, if you are analyzing a very slow activity, it may not be necessary to digitize every frame. A skip factor of 1 will grab every other frame, reducing your number of frames by half.

** Make sure you *press enter after each entry*, otherwise your entries will not be saved and the default value will be taken. **

F10 -- After all options have been entered, return to the previous menu.

(b) The grabbing process begins.

After all the specified frames have been grabbed, the computer returns to the previous menu, (capture menu). At this point rewind the tape to the calibration structure image (make sure it is from the same camera view as the grabbed images). When you have that image on the monitor you are ready to grab the control frame.

(a) When you highlight the Control option and press enter, it automatically grabs the image of the control frame; actually it grabs the image that is on the monitor at that time.

(g) Quit out of the capture menu.

At this point you are finished with the grabbing process for *one* camera. Now you have the option to start digitizing the images for this view, or you can go ahead and grab the images from the other view(s).

As you start the grabbing process for the rest of the views, use a ***different filename*** for each. To start digitizing, after grabbing one or all the camera views, exit out of the main menu (**F10**). The main APAS menu will appear on the screen, and you will be ready to start digitizing, (here is the fun part!!!).

B. Digitize

(Directory -- OK)

New	Old	Calibrate	Exit

New - set up a sequence. This sequence is going to be the *same* for *all* the camera views. Give the sequence a name, it needs to be different from the filename used in the grabbing module.

Enter Sequence information

File description:________________________________

Type: System Units: cm Filming date:

____

Height: ________ Weight: ________

points: _____ #Control points: _____

File description allows up to 30 characters. Remember this is for the sequence information, were all the views are going to be combined.

In the *type* option, use the *system* defined points.

Under *units* choose the ones that were used to measure the calibration structure.

I do not think that *filming date* needs any explanation.

Height and *weight* refer to the subject.

Under *number of points*, type how many points you want to digitize per frame. It should correspond to the number of joint markers placed on the subject.

The *control points* refer to the number of points to be digitized in the calibration structure.

Press **F10** when all the information has been entered.

SELECT POINT ID SOURCE:	
File	Keyboard

The file option will allow you to use points already defined in a previous file. This is used when you are going through the digitization process for the another trial or subject, you conveniently call on the points from the first view, instead of entering each point again. Find the sequence name from which you want to copy the points. For the first view of the first trial, press enter on **keyboard**. This will allow you to select the points you want to digitize.

The color monitor is going to display a list from which you can choose the joints for digitization, (point ID definitions). Fill in the point identifiers with the two-letter code, in the small screen. The sequence that is used is *very important*. The computer will later use the same sequence during digitization and connection of points.

After all the points have been identified press **F10.**

The computer is ready to call on the grabbed images. From the screen select the filename of the grabbed images, (make sure it is the correct camera view; specially if you are digitizing the second view).

The color monitor is going to display the first frame from the saved images. The small screen will ask for a sample figure. This frame is going to be digitized for the purpose of point connection, so it is important that all the points are visible, to provide greater accuracy in your sample. If the first image is not representative of all the points, you can advance the frames with the F1 key until you find the "best frame" for the sample. To select the image press enter. Then press **F10**.

The small screen is going to display a connection table, while the color monitor is going to display the sample frame. On the top of the connection table, the computer is going to indicate the joint to be digitized. With the mouse place the + on top of the joint marker, (middle of the marker), and press the left button to mark it. A number is going to appear beside the marked joint, indicating the identification of the joint.
Press **F1** to advance to the next joint. After all the joints have been marked press **F10**.

CONTROL POINT LOCATION SOURCE	
File	Keyboard

Select keyboard for the first view of your first trial. Again, as with the point identification, for subsequent trials and subjects you can select file.
The control point location table will appear on the screen. Enter the location of the points, from the calibration structure measurements, <u>in the order to be digitized</u>.
Make sure the measurements are all in the same units. The system defines + x to the right, + y up, and + z coming out of the screen.
Press **F10** after entering all points.

SELECT SEQUENCE OPTION					
New view	Old view	Points	Control pts	Seq-info	Quit

Select **new view** for each camera view. If you used only two cameras during filming, each sequence name should end up having two views.
A screen asking for information about each view will appear on the small monitor.
Use relevant words which will later help you identify the images. Press **F10** when done.

SELECT DIGITIZER TYPE				
Grabbed_video	Live_video	External	Joystick	Quit

Grabbed_video will call on the file that has the saved video images. Select the appropriate filename from the list provided, (keep in mind the camera view). The small monitor will show you the file's information, confirm your choice by pressing *any key*.

The first frame will appear on the color monitor, and a set of commands will appear on the small monitor.

A fixed point must be digitized first for each frame. The computer will indicate the joint to be digitized on the top of the screen. As you mark each body joint, you will see the connection line, which will create the stick figure. If you need to correct a marked joint press **F3** and the last digitized point will be erased. After marking all the points in the frame press **F1** to advance to the next frame.

... a few hours later, when all the frames have been digitized, press **F9** to digitize the control frame.

SELECT DIGITIZING OPTION							
Control	Synch	Point	Time	Update_clock	Bell	Frame #	Quit

Select **Control**.

SELECT CONTROL POINT OPTION			
Digitize	View	Read	Quit

Select **Digitize**.

Again the first point to be digitized is the fixed point. Then digitize the points in the calibration structure in the same order as the locations were entered.

Press **F1** after *all* the points have been digitized.

Highlight **Quit**.

If you are digitizing your second trial or subject, then you can use the previously digitized calibration structure. Go to **Read**, and find the sequence name that has already been digitized, the extension on the sequence name will indicate the view. Agian make sure you call the right view. After your selection, the calibration structure frame will come up on the screen, and it will include the digitized points; this is a good time to check if you called the correct view.

Press **F10** twice to exit out of the menus.

At this point you can go ahead and digitize the second view, (if you have already grabbed the images). To digitize the second view, select **Old** from the sequence menu. After you pick the sequence name from the list repeat the same process.

If you have to go back to grab the images for the second view, press **F10** one more time, and then enter **Yes**.

APPENDIX E

APPENDIX E

Subject's Data Sheet
ACL injured group

Subject's Name: ____________________________ Code: ______

Address: _______________________

____

Telephone: (   ) ________________

Date of birth (m/d/y): ______________

Dominant side: right left

Injured knee: right left

1. Approximate date of injury (m/y): ____________

2. Have you had ACL reconstruction: Yes No If yes, when? (m/y):

____

3. Approximate duration of rehabilitation after surgery (in months): __________

4. Did you play any sports prior to the injury? Yes No

   If yes, list: ____

   Level: recreational college professional

   Frequency (times per week): _________________________

5. Did you resume the same sport after rehabilitation of the injury? Yes No

   If your answer is no, please explain why?_____________________________________

   ____

6. Are you physically active? Yes No

6a. How would you rate your present exercise intensity compared with your level prior to the injury?

very low low same high very high

6b. How long are your workouts? (in min.) __________

6c. How many times a week do you exercise?

One Two Three More than three

7. Do you currently experience any pain in the injured knee while you exercise?

Yes No If yes, explain____________________________________

____

8. Do you feel restricted by your injury when participating in physical activities?

Yes No If yes, explain how?

____

____

9. Were your every day activities affected as a consequence of your injury? Yes No

If yes, explain ____

____

Subject's Data Sheet
Control group

Subject's Name: ______________________________ Code: ______

Address: ________________________

____

Telephone: ( ) ________________

Date of birth (m/d/y): ______________

Dominant side: right left

1. Have you ever had any knee injuries? Yes No

If yes, explain ____

2. Are you physically active? Yes No If yes, at what level:

recreational college sports professional sports

3. How many times a week do you exercise?

One Two Three More than three

4. How would you rate you workout?

Very light Light Moderate Intense Very intense

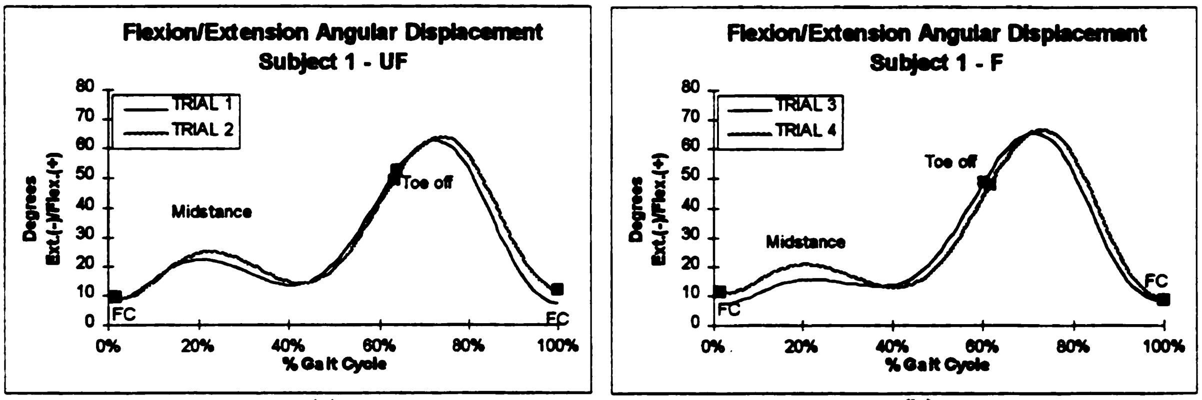
5. How long are your workouts? (in min.) _____________

APPENDIX F

APPENDIX F

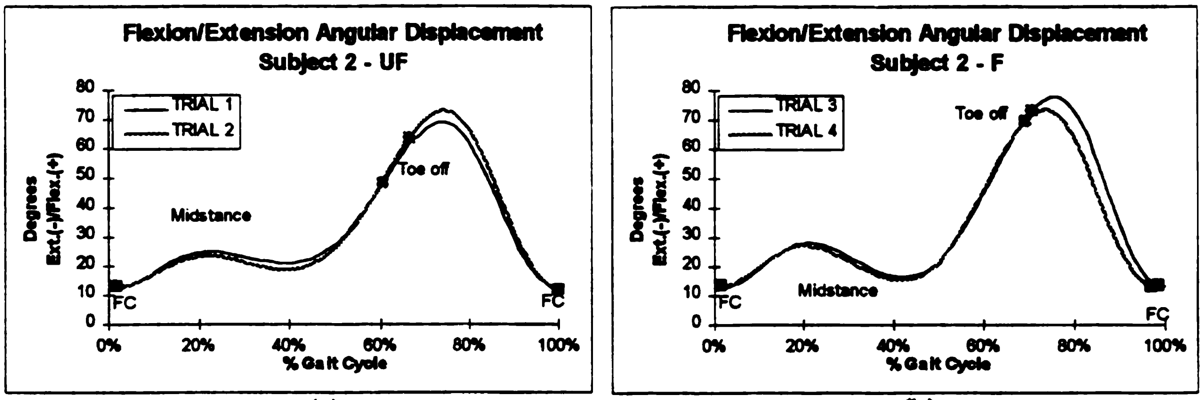
Angular Motion

Flexion/Extension and Abduction/Adduction angular displacements, velocities and accelerations. The Figures represent two analyzed trials under each tested condition. The first two trials under each condition were analyzed for all subjects. Subjects 1 - 3 formed part of the control group, while Subjects 4 and 5 were part of the ACL-reconstructed group.



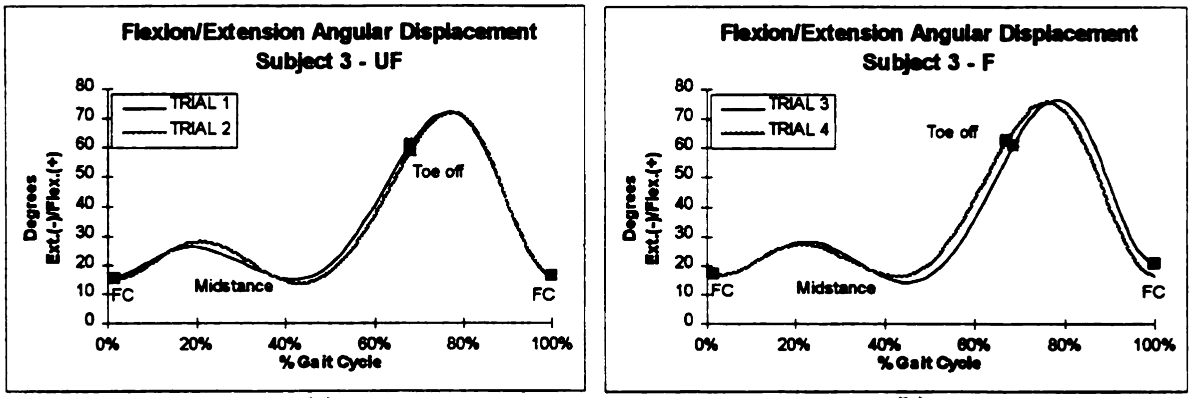
(a) (b)

Figure 1E: Knee Flexion/Extension Angular Displacement. Subject 1. (a) Unfatigued Trials (b) Fatigued Trials.



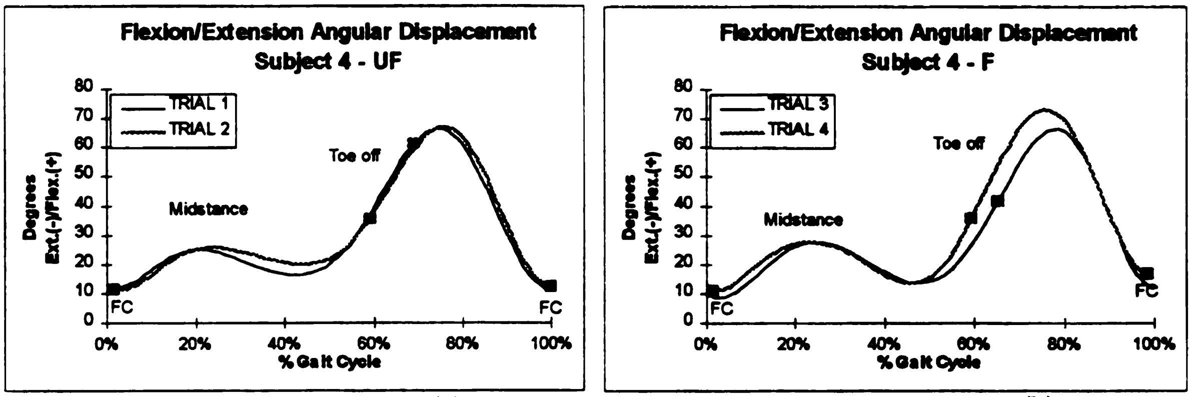
(a) (b)

Figure 2E: Knee Flexion/Extension Angular Displacement. Subject 2. (a) Unfatigued Trials (b) Fatigued Trials.



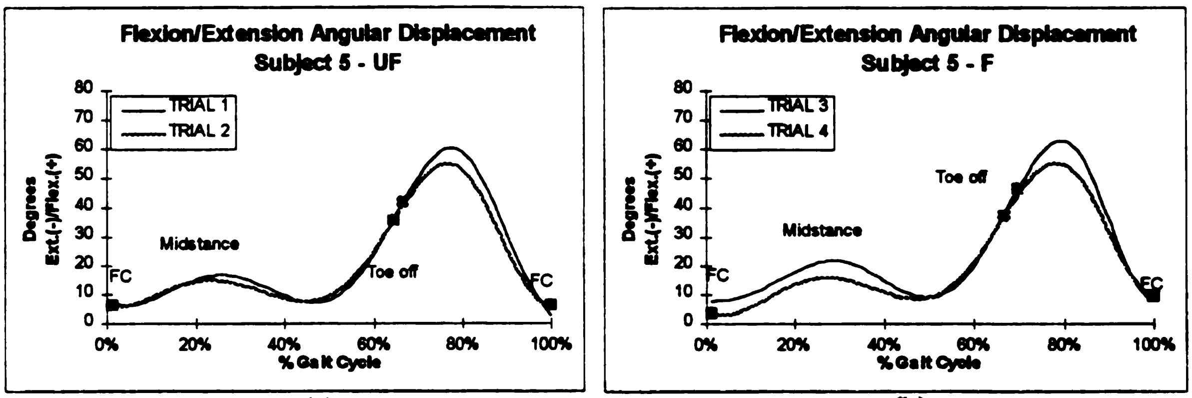
(a) (b)

Figure 3E: Knee Flexion/Extension Angular Displacement. Subject 3. (a) Unfatigued Trials (b) Fatigued Trials.



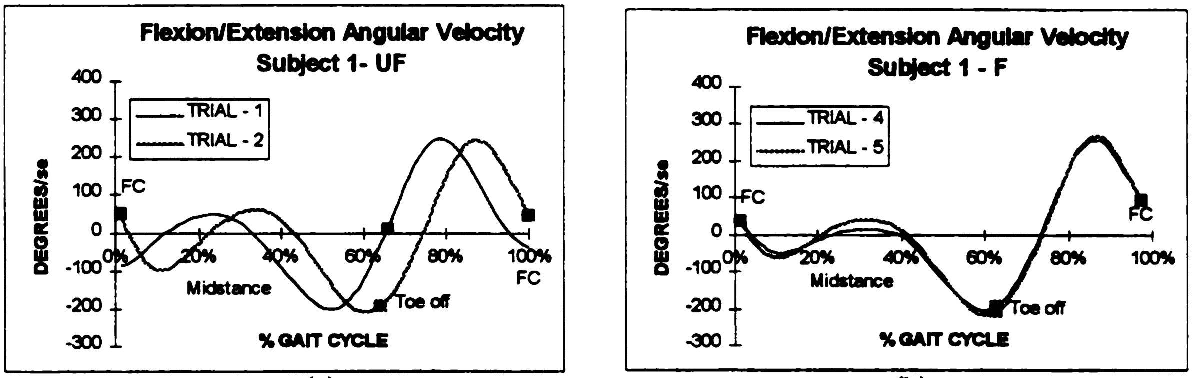
(a) (b)

Figure 4E: Knee Flexion/Extension Angular Displacement. Subject 4. (a) Unfatigued Trials (b) Fatigued Trials.



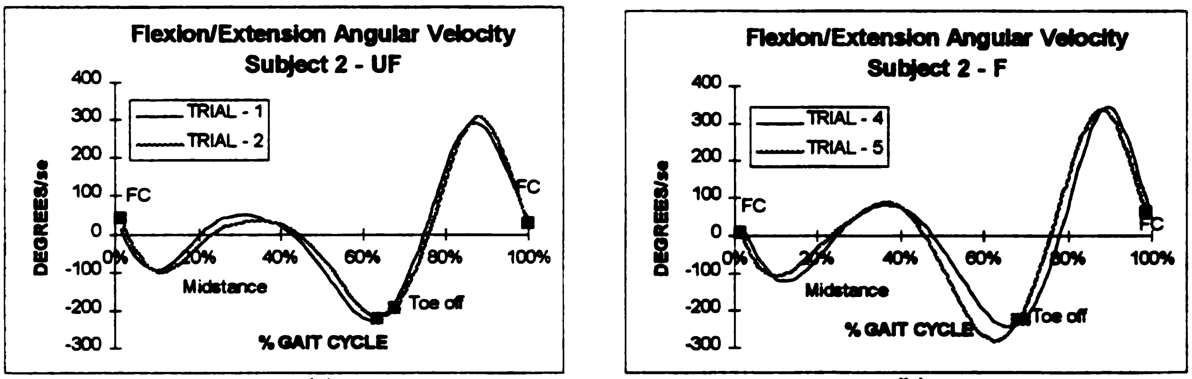
(a) (b)

Figure 5E: Knee Flexion/Extension Angular Displacement. Subject 5. (a) Unfatigued Trials (b) Fatigued Trials.



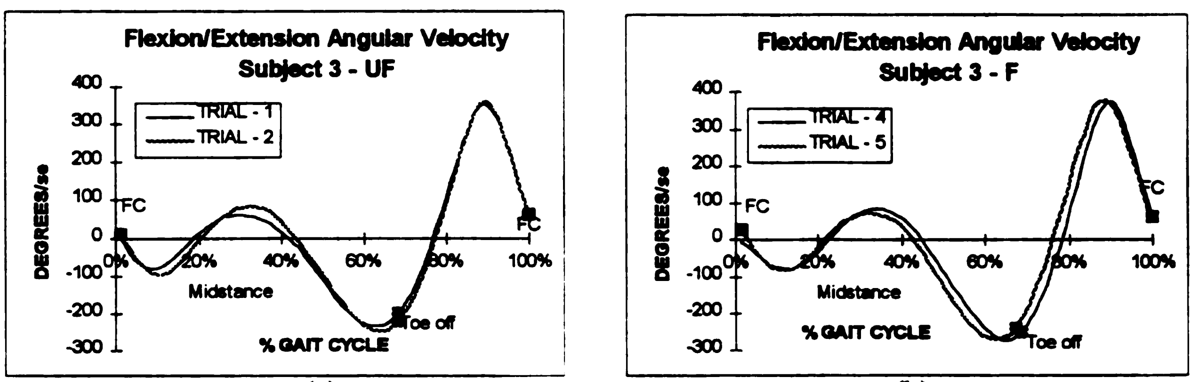
(a) (b)

Figure 6E: Knee Flexion/Extension Angular Velocity. Subject 1. (a) Unfatigued Trials (b) Fatigued Trials.



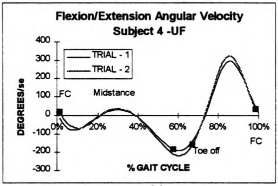
(a) (b)

Figure 7E: Knee Flexion/Extension Angular Velocity. Subject 2. (a) Unfatigued Trials (b) Fatigued Trials.

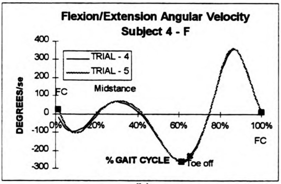


(a) (b)

Figure 8E: Knee Flexion/Extension Angular Velocity. Subject 3. (a) Unfatigued Trials (b) Fatigued Trials.

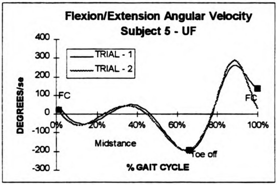


(a)

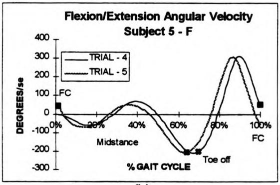


(b)

Figure 9E: Knee Flexion/Extension Angular Velocity. Subject 4. (a) Unfatigued Trials (b) Fatigued Trials.

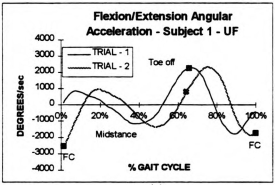


(a)

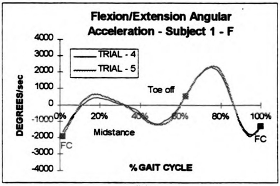


(b)

Figure 10E: Knee Flexion/Extension Angular Velocity. Subject 5. (a) Unfatigued Trials (b) Fatigued Trials.

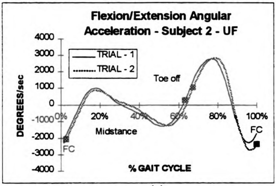


(a)

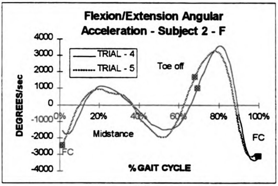


(b)

Figure 11E: Knee Flexion/Extension Angular Acceleration. Subject 1. (a) Unfatigued Trials (b) Fatigued Trials.

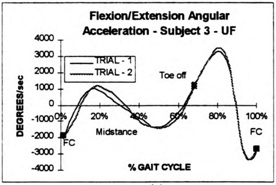


(a)

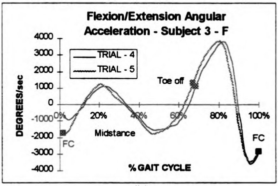


(b)

Figure 12E: Knee Flexion/Extension Angular Acceleration. Subject 2. (a) Unfatigued Trials (b) Fatigued Trials.

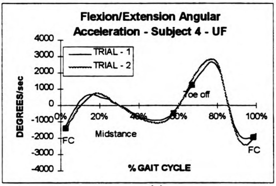


(a)

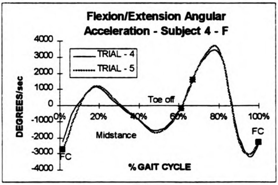


(b)

Figure 13E: Knee Flexion/Extension Angular Acceleration. Subject 3. (a) Unfatigued Trials (b) Fatigued Trials.

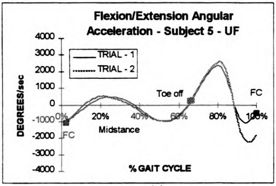


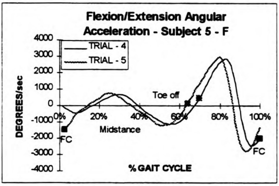
(a)



(b)

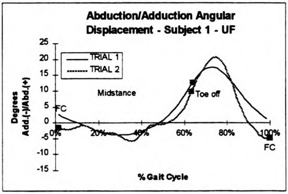
Figure 14E: Knee Flexion/Extension Angular Acceleration. Subject 4. (a) Unfatigued Trials (b) Fatigued Trials.

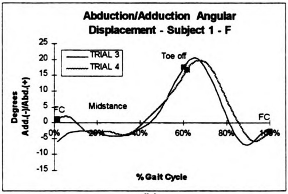




(a) (b)

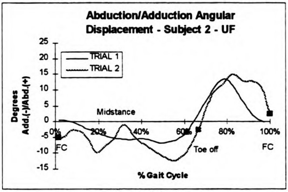
Figure 15E: Knee Flexion/Extension Angular Acceleration. Subject 5. (a) Unfatigued Trials (b) Fatigued Trials.

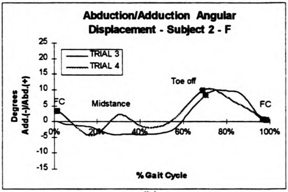




(a) (b)

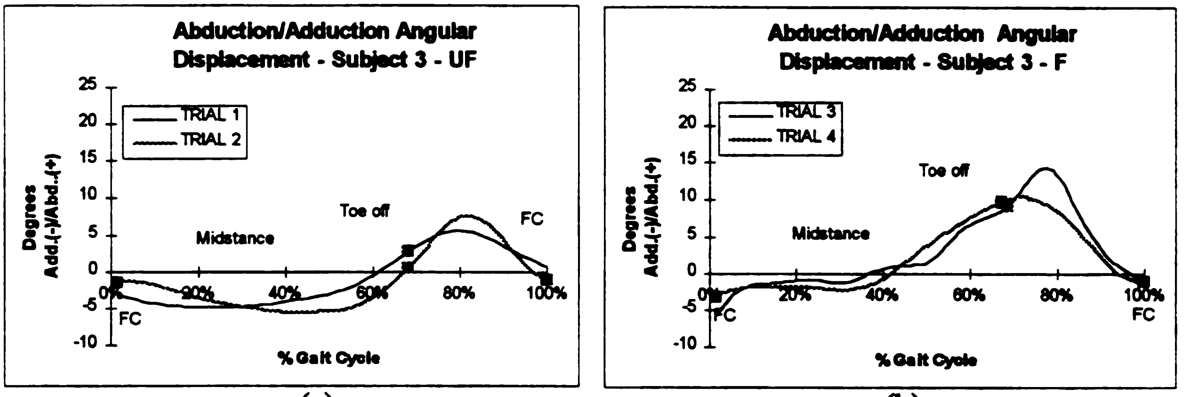
Figure 16E: Knee Abduction/Adduction Angular Displacement. Subject 1. (a) Unfatigued Trials (b) Fatigued Trials.





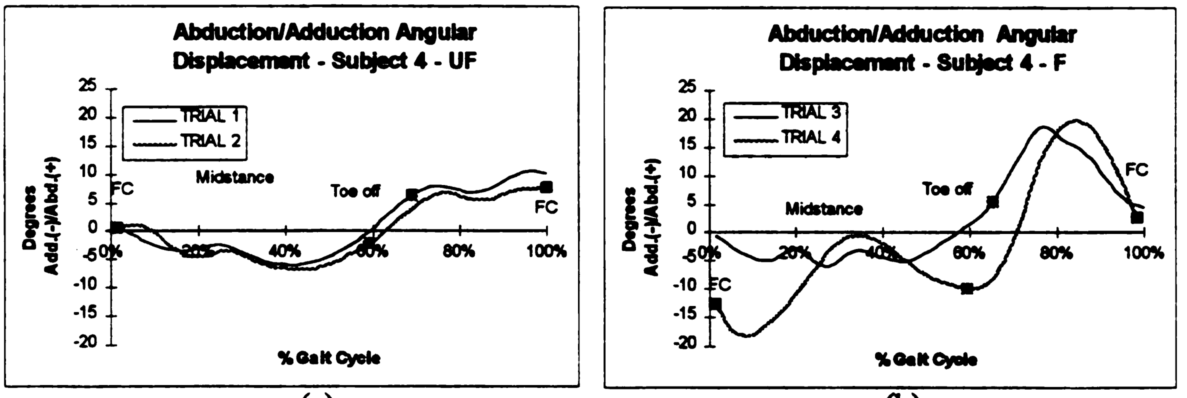
(a) (b)

Figure 17E: Knee Abduction/Adduction Angular Displacement. Subject 2. (a) Unfatigued Trials (b) Fatigued Trials.



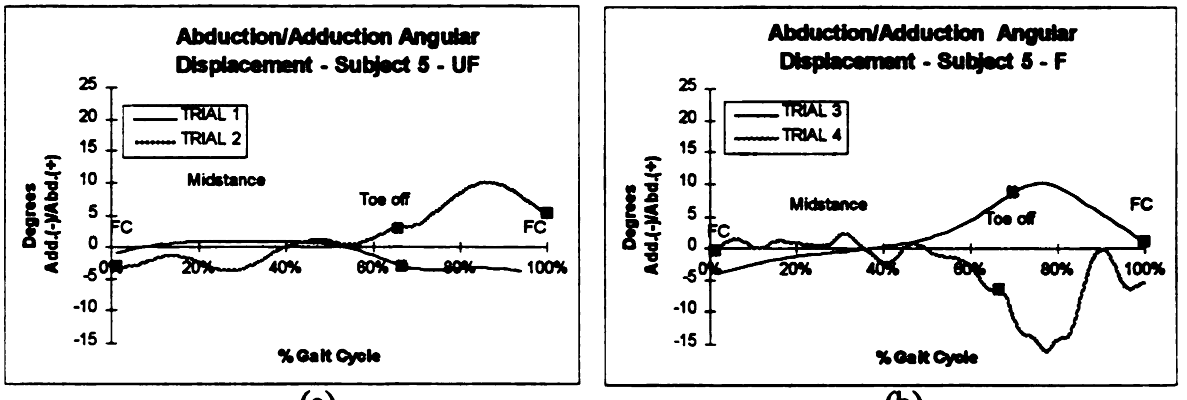
(a) (b)

Figure 18E: Knee Abduction/Adduction Angular Displacement. Subject 3. (a) Unfatigued Trials (b) Fatigued Trials.



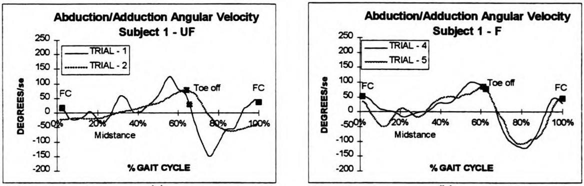
(a) (b)

Figure 19E: Knee Abduction/Adduction Angular Displacement. Subject 4. (a) Unfatigued Trials (b) Fatigued Trials.



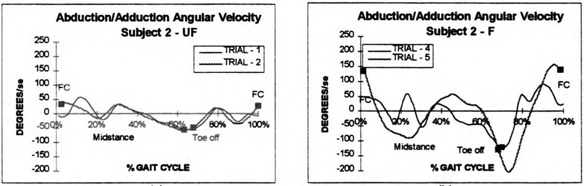
(a) (b)

Figure 20E: Knee Abduction/Adduction Angular Displacement. Subject 5. (a) Unfatigued Trials (b) Fatigued Trials.



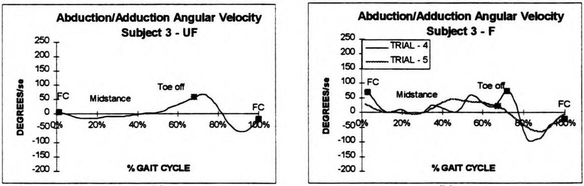
(a) (b)

Figure 21E: Knee Abduction/Adduction Angular Velocity. Subject 1. (a) Unfatigued Trials (b) Fatigued Trials.



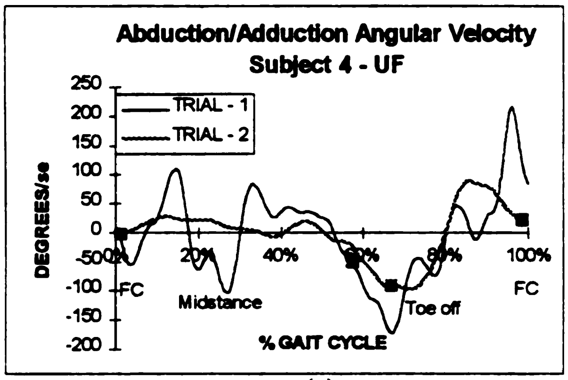
(a) (b)

Figure 22E: Knee Abduction/Adduction Angular Velocity. Subject 2. (a) Unfatigued Trials (b) Fatigued Trials.

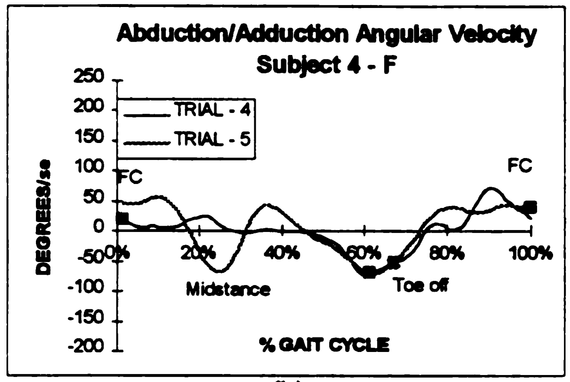


(a) (b)

Figure 23E: Knee Abduction/Adduction Angular Velocity. Subject 3. (a) Unfatigued Trials (b) Fatigued Trials.

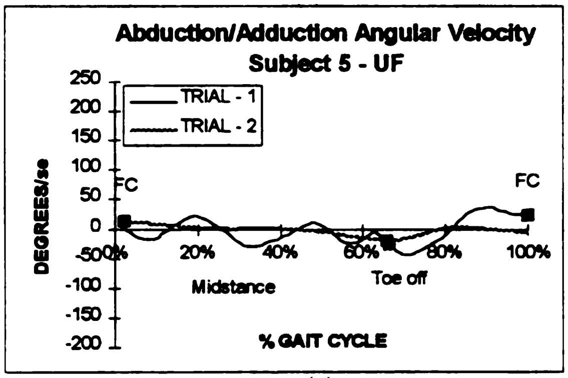


(a)

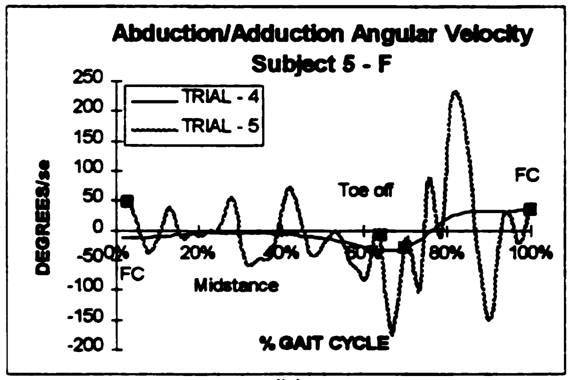


(b)

Figure 24E: Knee Abduction/Adduction Angular Velocity. Subject 4. (a) Unfatigued Trials (b) Fatigued Trials.

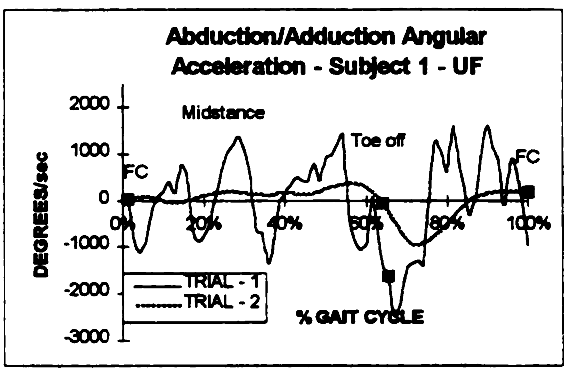


(a)

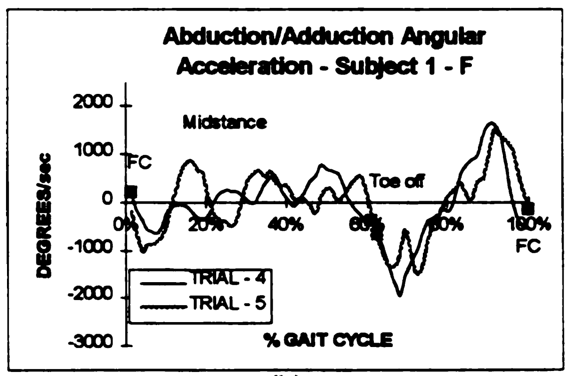


(b)

Figure 25E: Knee Abduction/Adduction Angular Velocity. Subject 5. (a) Unfatigued Trials (b) Fatigued Trials.

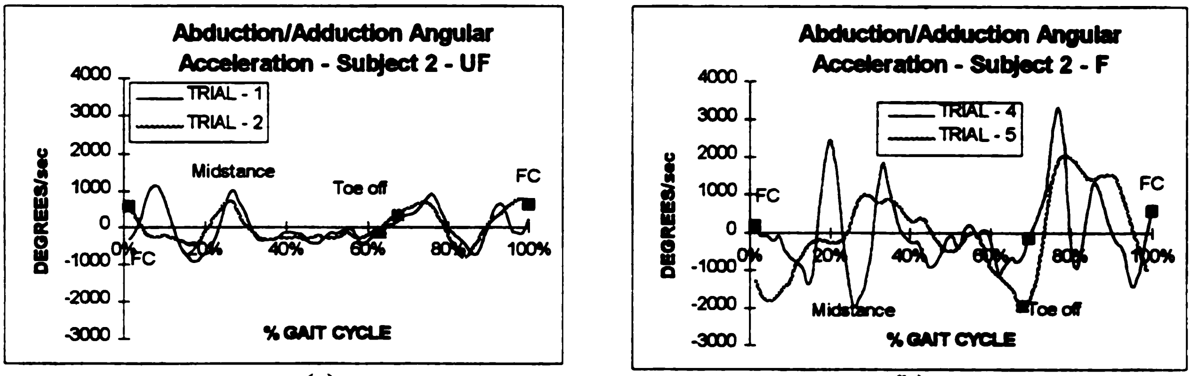


(a)



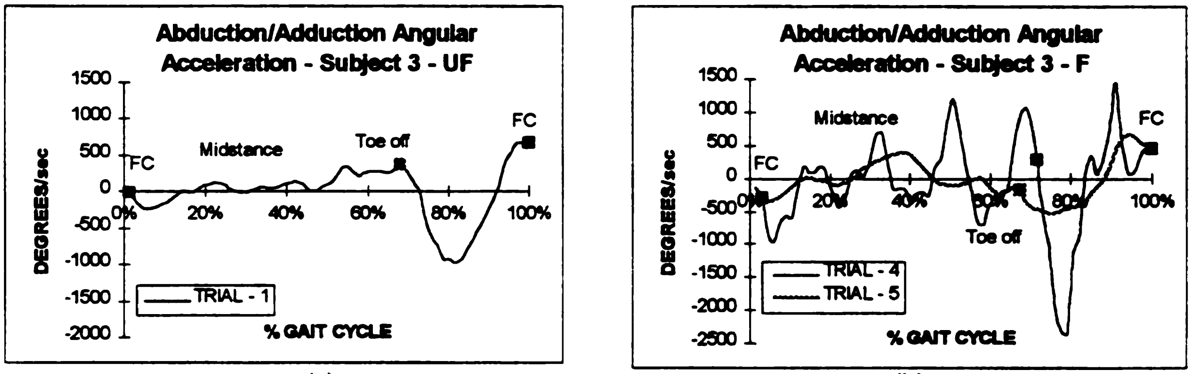
(b)

Figure 26E: Knee Abduction/Adduction Angular Acceleration. Subject 1. (a) Unfatigued Trials (b) Fatigued Trials.



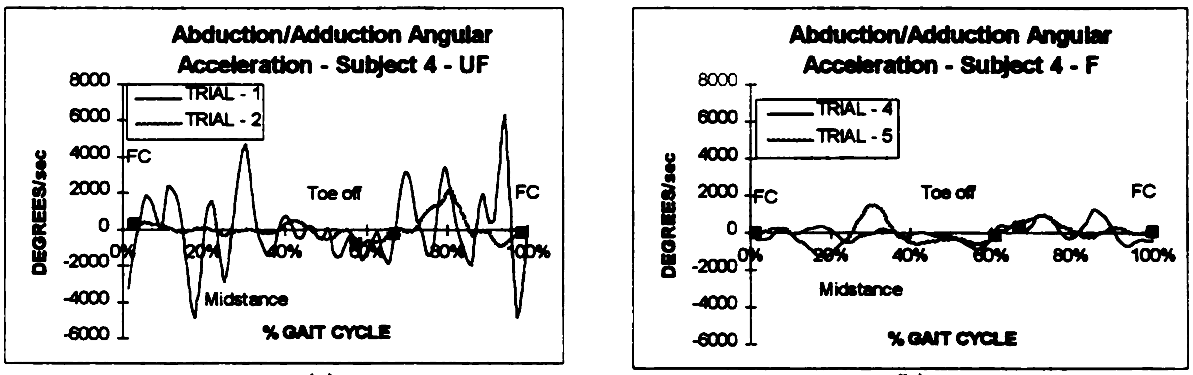
(a) (b)

Figure 27E: Knee Abduction/Adduction Angular Acceleration. Subject 2. (a) Unfatigued Trials (b) Fatigued Trials.



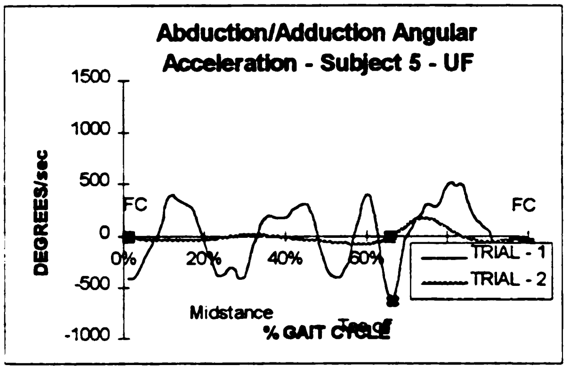
(a) (b)

Figure 28E: Knee Abduction/Adduction Angular Acceleration. Subject 3. (a) Unfatigued Trials (b) Fatigued Trials.

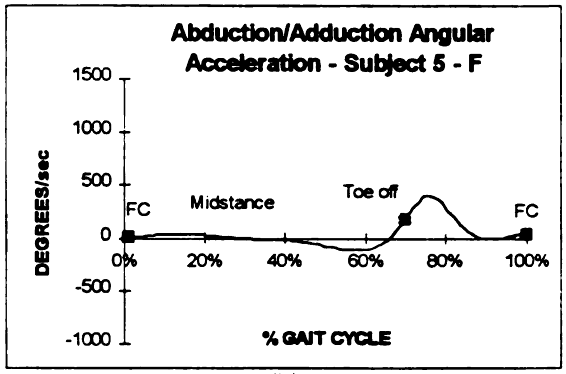


(a) (b)

Figure 29E: Knee Abduction/Adduction Angular Acceleration. Subject 4. (a) Unfatigued Trials (b) Fatigued Trials.



(a)



(b)

Figure 30E: Knee Abduction/Adduction Angular Acceleration. Subject 5. (a) Unfatigued Trials (b) Fatigued Trials.