EFFECTS OF KNEE EXTENSION CONSTRAINT ON KNEE FLEXION ANGLE AND GROUND REACTION FORCES AFTER ACL RECONSTRUCTION

by
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ABSTRACT

CHRIS STANLEY: Effects of Knee Extension Constraint on Knee Flexion Angle and Ground Reaction Forces After ACL Reconstruction
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Small knee flexion angles at initial contact and large ground reaction forces during landing have been identified as risk factors for non-contact anterior cruciate ligament injuries that are common in sports. This study investigated the effects of knee extension constraint on knee flexion angle and ground reaction forces for patients who were 4-6 months post ACL reconstruction surgery, and also compared these patients to healthy subjects' motion patterns. Three-dimensional videographic and force plate data were collected for 12 ACL reconstruction patients and 12 age and gender matched healthy subjects performing level walking, jogging, and stair descending under 3 conditions: knee brace with a constraint to extension, traditional knee brace, and no knee brace. The constrained knee brace significantly increased knee flexion angle at landing, but did not significantly affect the peak ground reaction forces at landing. The constrained knee brace may be useful in the prevention and rehabilitation of non-contact anterior cruciate ligament injuries in sports.

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CHAPTER I

INTRODUCTION

Anterior cruciate ligament (ACL) rupture is a common knee injury in sports. ACL injury results in short-term disability and puts the individual at an increased risk for developing secondary knee disorders, such as osteoarthritis (Yu et al, 2004). Studies have found that many individuals with complete ACL rupture had chronic knee instability and secondary damage to menisci and chondral surfaces (Irvine and Glasgow, 1992; Finsterbush et al, 1990). These injuries can affect participation in sports, but may also impair the ability to perform functional activities during daily living. A study by Noyes et al (1989) found that ACL rupture may lead to moderate to severe disability in 31% of patients for walking activities, 44% for routine activities of daily living, and 77% during sporting activities with frequent cutting and pivoting. The majority of ACL injuries occur in individuals between the ages of 15-45 with the incidence of injury being 1 in every 1,750 people in that age range (Griffin, 2000). The annual incidence of ACL ruptures in the general population is estimated at 1 in 3,000 people (Frank and Jackson, 1997; Miyasaka et al, 1991). Approximately 175,000 primary ACL reconstruction surgeries are performed annually in the United States, and the estimated annual cost for these surgeries is over $2 billion (Gottlob et al, 1999). Not all ACL ruptures are treated surgically, and the treatment option generally depends on the patient and their lifestyle.

The ACL helps to join the femur and the tibia at the knee joint to allow for normal motion. The primary functions of the ACL are to control anterior tibial translation relative to the femur and to provide rotary stability (Cabaud, 1983). Anterior tibial translation occurs from an anterior shear force on the tibia, which is a major contributor to ACL strain.

About 70-80% of ACL injuries are categorized as non-contact in nature (Griffin, 2000; Kirkendall and Garrett, 2000; Boden et al, 2000). Non-contact ACL injuries frequently occur during the deceleration phase of landing after a jump or in preparation for a cutting maneuver (Kirkendall and Garrett, 2000) with the knee most often in a position near full extension (Boden et al, 2000; Kirkendall and Garrett, 2000). Females sustain ACL injuries at a higher rate than males during athletic competition (Arendt, Agel, and Dick, 1999; Malinzak et al, 2001). A characteristic of female recreational athletes is their small knee flexion angle during landing tasks that are preceded by horizontal movements, which may predispose females to ACL injury (Boden et al, 2000; Malinzak et al, 2001; Decker et al, 2003). Female athletes also tend to have greater knee valgus angles, increased quadriceps muscle activation, and decreased hamstring muscle activation during the stance phase of running and cutting tasks (Malinzak et al, 2001). Another proposed reason for the higher incidence of ACL injury among women is that female athletes have greater ground reaction forces than males (James et al, 2004; Hewett et al, 1996).

Risk factors of non-contact ACL injuries can be separated into intrinsic and extrinsic factors. Intrinsic factors deal with the anatomical structures, physiological properties, and motor control related biomechanical factors. Extrinsic factors involve things that are external to the person that may affect their motion patterns. Risk factors are identified in the hopes of developing effective injury prevention programs (Griffin et al, 2000).

A small knee flexion angle during landing or cutting maneuvers is a major risk factor in ACL injury. The posterior fibers of the ACL provide the principle restraint to hyperextension and are most taut when the knee is near full extension, which means that they are more susceptible to injury at full extension (Markolf et al, 1995; Cabaud, 1983). The length of the ACL decreases as the knee flexion angle increases (Li et al, 2005). Decreasing the knee flexion angle during landing, therefore, increases the length of the ACL, which increases loading on the ACL and the chance for injury.

Increasing the knee flexion angle at landing should help to reduce anterior shear force imposed on the knee. Research shows that ACL loading increases as the knee flexion angle decreases (Fleming et al, 1999; Heijne et al, 2004). Anterior shear force on the tibia imposed by the patellar tendon decreases as the patellar tendon-tibia shaft angle decreases. The patellar tendon-tibia shaft angle decreases as the knee flexion angle increases (Yu et al, 2004). Chappell et al (2002) found that women had a significantly greater proximal tibia anterior shear force than men during the landing of a stop-jump task. Proximal tibia anterior shear force is an important contributor to anterior tibial translation, which causes strain on the ACL (Chappell et al, 2002). The increased proximal tibia anterior shear force during landings of female recreational athletes is most likely due to decreased knee flexion angle, increased quadriceps muscle activation, and/or decreased hamstring muscle activation (Malinzak et al, 2001). Posterior ground reaction force is a major contributor to the anterior shear force at the proximal tibia (Yu et al, 2004). Results from Yu et al (2004) show that females had increased posterior ground reaction force at landing in a stop-jump task. Females also exhibit greater peak vertical and posterior ground reaction forces than males in a drop landing (Kernozek et al, 2005). High ground reaction forces at landing have been cited as a risk factor for ACL injury (Malinzak et al, 2001; Chappell et al, 2002; Decker et al, 2003). Decreasing the impact forces at landing should decrease the loading on the knee.

Yu, Lin, and Garrett (2005) concluded that peak posterior ground reaction force, peak knee extension moment, and peak proximal anterior shear force occur at approximately the same time during stance. Therefore, the ground reaction forces, knee extension moment, and proximal anterior shear force will be the largest at the same point in stance. A small knee flexion angle increases ACL loading, so the knee flexion angle at this peak proximal anterior shear force may also be an important variable.

Knee braces are commonly prescribed to ACL reconstruction patients to help regain knee function and protect against new injuries. Knee bracing may also be used as a gait-training tool to prevent ACL injuries. The knee brace primarily prevents excessive anteromedial rotation and subluxation of the tibiofemoral joint (Wu, Ng, and Mak, 2001). The newly designed knee brace used in this study has a constraint to knee extension designed to increase the knee flexion angle during landings of functional activities. In a previous study, the new knee brace with a constraint to extension significantly increased the knee flexion angle during the landing phase of a stop-jump task (Yu et al, 2004).

The purpose of this study is to determine the effects of the newly designed knee brace on the lower extremity kinematics and kinetics of patients after ACL reconstruction in functional activities such as level walking, jogging, and stair descending. More specifically, this study will compare ACL reconstruction patients' motion patterns over all brace conditions, and compare ACL reconstruction patients' motion patterns with healthy subjects' motion patterns. It is hypothesized that (1) the newly designed knee brace will increase the knee flexion angle at initial contact and decrease peak ground reaction forces during level walking, jogging, and stair descending for ACL reconstruction patients; and (2) ACL reconstruction patients wearing the brace with a constraint to knee extension will have motion patterns similar to healthy subjects without a knee brace. The results of this study will provide significant information for future clinical applications of the newly designed knee brace with a constraint to knee extension for ACL injury prevention and rehabilitation.

CHAPTER II

LITERATURE REVIEW

The incidence of ACL injuries, structure and function of the ACL, risk factors of non-contact ACL injuries, and effects of wearing a functional knee brace were reviewed in this chapter.

2.1 Incidence of ACL Injury

ACL rupture is a common injury in sports that may affect an athlete's performance and quality of life. ACL injury results in short-term disability and increases the risk of developing secondary knee disorders, such as osteoarthritis, chronic knee instability, and secondary damage to menisci and chondral surfaces (Yu et al, 2004; Irvine and Glasgow, 1992; Finsterbush et al, 1990). The annual incidence of ACL ruptures in the general population is estimated at 1 in 3,000 people (Frank and Jackson, 1997; Miyasaka et al, 1991), but ACL injury occurs in 1 in 1,750 individuals between the ages of 15-45 (Griffin, 2000). Approximately 175,000 primary ACL reconstruction surgeries are performed annually in the United States with an estimated annual cost over $2 billion (Gottlob et al, 1999). Treatment of ACL ruptures generally depends on the patient and their lifestyle. The number of ACL ruptures annually is larger than 175,000 because not all injuries are treated surgically. Given a hypothetical case of a 20 year-old athlete who ruptured his ACL and wanted to return to collegiate athletics, 18 of 58 orthopedic surgeons chose non-surgical treatment (Johnson, 1983). A different study of 167 patients with ACL rupture compared three treatment options: surgery with augmentation of the ACL with a strip of the IT band (50 patients), surgery without augmentation (25 patients), and non-surgical treatment (92 patients). The non-surgical group had more knee joint laxity and instability, and 17% of them ended up getting ACL reconstruction surgery (Andersson, Odensten, and Gillquist, 1991). Although ACL reconstruction surgery seems to be the best treatment, it may not be the best option for all patients and is not always the treatment that is chosen. Therefore, the number of ACL reconstruction surgeries that are performed each year underestimates the actual number of ACL ruptures.

About 70-80% of ACL injuries are categorized as non-contact in nature (Griffin, 2000; Kirkendall and Garrett, 2000; Boden et al, 2000). Non-contact ACL injuries often occur during the deceleration phase of landing after a jump or in preparation for a cutting maneuver (Kirkendall and Garrett, 2000) with the knee frequently in a position near full extension (Boden et al, 2000; Kirkendall and Garrett, 2000).

2.2 ACL Structure

The ACL helps to join the femur and the tibia at the knee joint to allow for normal motion. It attaches the anterior surface of the tibia to the posterior part of the medial surface of the lateral femoral condyle (Clemente, 1997). The ACL contains two bundles of fibers, and the tautness of these fibers varies with knee flexion angle. The anteromedial bundle is taut in 90° of knee flexion, and the posterolateral bundle is taut as the knee moves into extension (Cabaud, 1983).

The ACL functions to control anterior translation of the tibia relative to the femur and provide rotary stability (Cabaud, 1983). A secondary function of the ACL is to prevent hyperextension of the knee. The hamstring muscles act with the ACL to control anterior translation of the tibia relative to the femur so if either of these structures is injured it will place more strain on the other. Increased quadriceps activity will increase the anterior tibial translation, which increases ACL strain and risk of injury (Ramsey et al, 2003; DeMorat et al, 2004). DeMorat et al (2004) found that an aggressive quadriceps load produced enough anterior translation of the tibia to compromise the ACL structure. When the knee is near full extension, the hamstring muscle contraction is unable to provide a large enough posterior shear force on the tibia to resist anterior tibial translations relative to the femur (Pandy and Shelburne, 1997). The ACL is the other major structure that prevents anterior tibial translation so it is strained more since the hamstring muscle cannot provide enough resistive force on its own at small knee flexion angles. Anterior shear force on the tibia through the patellar tendon decreases as the patellar tendon-tibia shaft angle decreases; the patellar tendon-tibia shaft angle decreases as the knee flexion angle increases (Yu et al, 2004). Therefore, increasing the knee flexion angle should decrease the anterior shear force, which will decrease the strain on the ACL. Figure 1 illustrates the patellar tendon-tibia shaft angle.

2.3 Risk Factors

Risk factors of non-contact ACL injuries are generally separated into intrinsic and extrinsic factors. Intrinsic risk factors are those related to anatomic structure, physiological properties, and motor control related biomechanical factors such as Q-angle, the width of femoral condyle notch (Shambaugh et al, 1991), knee joint laxity (Wojtys et al, 1998), hormonal effects (Wojtys et al, 1998), imbalanced lower extremity strength (Boden et al, 2000; Kirkendall and Garrett, 2000), lower extremity malalignment (Kirkendall and Garrett, 2000), and the altered lower extremity motion patterns (Boden et al., 2000; Malinzak et al., 2001; Decker et al., 2003). Extrinsic factors for ACL injury are playing surface (Powell and Schootman, 1992), shoe to surface interface (Garrick and Requa, 1996), and athlete's playing style (Kirkendall and Garrett, 2000). The playing surface can affect the athlete's landing during competition or practice. There is a reduced risk of ankle and knee sprains on grass fields as compared to Astroturf fields (Orchard and Powell, 2003). In addition, there is less risk for ACL injuries and knee sprains in cold weather, which the authors attribute to reduced shoe-surface traction from cold weather (Orchard and Powell, 2003).

Running, jumping, and landing mechanics may put some athletes at higher risk for injury. The following characteristics increase strain on the ACL and risk of injury: small knee flexion angle during landing tasks (Boden et al, 2000; Malinzak et al, 2001; Decker et al, 2003), large knee valgus angles, increased quadriceps muscle activation and decreased hamstring muscle activation during the stance phase of running and cutting tasks (Malinzak et al, 2001), and high ground reaction forces during landing (James et al, 2004; Hewett et al, 1996).

Some other risk factors for ACL injury are abnormal gait and imbalance in muscle strength between the quadriceps and hamstring muscles. After ACL injury, many patients alter their gait patterns to compensate for their injury. Abnormal gait causes the body to be strained in areas that it is not used to, which may lead to ACL re-injury or contralateral injury. Studies have shown that individuals with ACL injuries tend to use greater extensor torques at the hip and ankle and reduced extensor torque at the knee during the stance phase of running (Berchuck et al, 1990; DeVita, Blankenship, and Skelly, 1992), and the same adaptations have been observed in walking (Andriacchi and Birac, 1993; Berchuck et al, 1990). A study over a 5-year period determined that about 12% of patients who have had ACL reconstruction sustain a repeat ACL rupture or contralateral ACL rupture (Salmon et al, 2005). The hamstring muscle helps to prevent the tibia from moving anterior relative to the femur, which is what the healthy ACL does. Increased quadriceps activity will increase the anterior tibial translation, which increases ACL strain and risk of injury (Ramsey et al, 2003; DeMorat et al, 2004). This study will focus primarily on the knee flexion angle during the stance phase of functional activities.

2.3.1 Effects of Ground Reaction Forces on ACL Loading

Ground reaction forces are measured in order to calculate the forces and moments in lower extremity joints. An estimate of the loading on the ACL can be made if the loading at the knee joint is determined. High ground reaction forces at landing have been cited as a risk factor for ACL injury (Malinzak et al, 2001; Chappell et al, 2002; Decker et al, 2003). Using inverse dynamics, a known ground reaction force can be used to estimate forces at joints so a large impact force would translate to a large knee joint resultant. Decreasing the impact forces at landing should decrease the loading on the ACL. Yu, Lin, and Garrett (2005) studied lower extremity kinematics and kinetics during the landing of a stop-jump task. Female subjects were found to have significantly smaller hip and knee flexion angles at landing than males. Yu, Lin, and Garrett state that large hip and knee flexion angles at landing will not necessarily reduce the impact force, but active hip and knee flexion motions will reduce impact. The hip and knee joint angular velocities will increase the time of impact

(Δt), which should decrease the impact force (F) according to the impulse-momentum equation ($F\Delta t=m\Delta v$). This study found that the hip and knee joint angles did not significantly affect the ground reaction forces and joint resultants during landing, but the authors believe that the angles may still affect the loading of specific joint structures, like the ACL.

Nunley et al (2003) proved that the patella tendon-tibia shaft angle increases as the knee flexion angle decreases, which increases loading on the ACL. Although it was not found significant in the study by Nunley et al, females had greater vertical ground reaction force, proximal anterior shear force, and smaller knee flexion angle at peak proximal tibia anterior shear force. The peak proximal tibia anterior shear force is the moment when the ACL is strained the greatest. Large vertical ground reaction force and small knee flexion angle are two characteristics that occur when the ACL is strained the most, so decreasing vertical ground reaction force and increasing knee flexion angle may help to reduce the risk of ACL injury. Yu, Lin, and Garrett (2005) found that peak posterior ground reaction force during landing had significant correlation with peak vertical ground reaction force, peak knee extension moment, and peak proximal anterior shear force during landing. Peak vertical ground reaction force was also significantly correlated to peak knee extension moment and peak proximal anterior shear force, and peak proximal anterior shear force was correlated to peak knee extension moment. Yu, Lin, and Garrett concluded that peak posterior ground reaction force, peak proximal anterior shear force, and peak knee extension moment during landing during a stop-jump task occurred at about the same time. Hence, the authors believe that peak ground reaction forces may be used to predict ACL loading.

2.3.2 Gender Differences

ACL injuries are more frequent in females than in males during athletic competition; the rate has been reported anywhere from 2-8 times higher for females (Arendt, Agel, and Dick, 1999; Malinzak et al, 2001). Understanding the differences in lower extremity kinematics between genders will help to determine characteristics that cause females to be at a higher risk of injury. Female recreational athletes tend to have a small knee flexion angle during landing tasks that are preceded by horizontal movements, which may predispose females to ACL injury (Boden et al, 2000; Malinzak et al, 2001; Decker et al, 2003). Female athletes also commonly have greater knee valgus angles, increased quadriceps muscle activation, and decreased hamstring muscle activation during the stance phase of running and cutting tasks (Malinzak et al, 2001). Another proposed reason for the higher incidence of injury among women is that female athletes have been found to have greater ground reaction forces than males (James et al, 2004; Hewett et al, 1996). Other intrinsic factors that increase the risk of ACL injury in women are smaller cross-sectional area of the ACL, narrower intercondylar notch (Shelbourne, Davis, and Klootwyk, 1986), greater knee joint laxity (Huston and Wojtys, 1996; Wojtys et al, 1998), and hormonal variations (Wojtys et al, 1998). Extrinsic factors between genders include level of conditioning, muscle strength, and different motor control strategies (Malinzak et al, 2001; Malone et al, 1993). These traits that make females more likely than males to suffer ACL injury can be called risk factors.

Malinzak et al (2001) compared knee joint motion patterns between genders for running, side-cutting, and cross-cutting tasks. Gender did not significantly affect the approach run speed or takeoff speed in any of the tasks. Female subjects had a smaller knee flexion angle and more valgus than male subjects throughout the tasks. Female subjects also tended to have more quadriceps muscle activation and less hamstring muscle activation than the males. The combination of the increased quadriceps and decreased hamstring activation increases the chances of greater anterior shear force at the knee. This study proves that male and female athletes have different knee motion patterns in selected athletic tasks. The smaller knee flexion angle, larger valgus angle, increased quadriceps activation, and decreased hamstring activation by the females increases their risk of ACL injury.

Lephart et al (2002) found similar results in that females have smaller knee flexion angles during landing of both a single leg landing and a forward hop task when compared to males. For both tasks, females had less time to maximum angular displacement for knee flexion than males, which means that there is a more abrupt absorption of impact forces during landing. The rapid knee flexion during landing of females may be attributed to weak quadriceps and hamstring muscles because they are unable to control knee flexion during impact. These two factors put females at higher risk for sustaining ACL injury.

During a stop-jump, women had a significantly greater proximal tibia shear force than men (Chappell et al, 2002). This proximal anterior shear force is a major contributor to anterior tibial translation, which causes excessive strain on the ACL. Chappell et al state that the increased proximal tibia anterior shear force in women may be attributed to small knee flexion angle, increased quadriceps muscle force, decreased hamstring muscle force, or a combination of these factors. This study also found females to have valgus moments at the knee during landings of vertical and backwards stop-jumps while the males tended to have varus moments. Chappell et al did not find a difference in the magnitude of the knee varus-valgus moments between genders. Therefore, the authors do not believe that knee varus-valgus is responsible in the gender difference of ACL strain during stop-jump tasks. The results also showed that females had a knee extension moment during landing, and the males had a knee flexion moment during landing. The knee extension moment seen by the females means that they have greater quadriceps muscle activation than hamstring activation. These differences in neuromuscular motor control strategies may be a cause of increased ACL strain in females as compared to males. Chappell et al may have underestimated the ground reaction forces and peak joint resultant forces and moments because the video cameras had a relatively low sampling rate. Another limitation is that the females on average had a lower approach run speed than the males. The authors believe that the females would have had even greater peak proximal tibia anterior shear forces if the approach speeds were the same between genders.

James et al (2004) studied healthy high school and collegiate basketball players to compare cutting techniques by gender. Females had 5.8° less knee flexion at landing and 1 N greater ground reaction force at maximum knee flexion than males. Females also had a higher peak ground reaction force, although this was not a significant difference. James et al state that the greater ground reaction force at maximum knee flexion for females was not caused by body mass or running velocity so the difference must be attributed the knee kinematics during the cutting maneuver.

Decker et al. (2003) studied gender differences in lower extremity kinematics, kinetics, and energy absorption during landing of a drop-jump. The participants were healthy recreational athletes. This study did not find any difference in vertical ground reaction force between males and females, which contrasts with other studies (James et al, 2004; Hewett et al, 1996). There were no significant differences in peak moments at any joints, but there was a difference in the temporal occurrence of peak knee extensor moment. The time to the peak knee extensor moment from landing corresponded to the peak vertical ground reaction of the forefoot force for females, whereas the peak knee extensor moment corresponded to the peak vertical ground reaction of the rearfoot for males. The results showed that females were in a more erect position at landing than males and also displayed greater knee and ankle range of motion and angular velocities throughout the landing phase.

2.4 Rehabilitation Program

ACL rehabilitation is a long process that begins soon after surgery. The first two weeks generally focus on reducing pain and swelling and increasing range of motion (Maksic, 2003; Cross, 1998). The next four weeks continue to increase range of motion and weightbearing (Maksic, 2003; Cross, 1998). Then from 6-12 weeks patients begin to return to sport-specific activities in order to improve proprioception, strength, and muscular control (Shelburne et al, 1995; Cross, 1998). Functional activities like running, cutting, and jumping are introduced around 3-4 months after surgery (Maksic, 2003). If the rehabilitation program is successful, athletes may return to full sports activity at six months (Shelburne et al, 1995; Maksic, 2003; Cross, 1998).

Literature supports the use of a rehabilitative brace, functional brace, or a combination of the two braces from 2 weeks to 4 or 5 months following ACL reconstruction surgery (Blackburn, 1985; Paulos, Wnorowski, and Beck, 1991; Shelbourne and Wickens, 1990). Optimal duration or stop and start times for knee bracing have not been determined in the literature. Beynnon et al (1997) determined that the ACL graft resembles a healthy ACL after 8 months of healing, so they concluded that bracing may not be need after this time. Animal studies on monkeys found that the ACL graft is still weak and continues to remodel after 3 months (Butler et al, 1989; Clancy et al, 1981). This suggests that knee bracing is needed during this time period of 3-8 months.

The selection criteria of 4-6 months after ACL reconstruction surgery was chosen for this research study. This is the time when patients have begun functional activities, but have not been cleared to return to sports. Their ACL graftis still healing so it needs the protection of a functional knee brace. In addition, the patients are just returning to activities and are probably still relearning the motion patterns so the functional knee brace may be helpful in training the gait pattern.

2.5 Effects of a Knee Brace

A study by Yu et al (2004) compared the lower extremity kinematics and kinetics of a stop-jump using a newly designed knee brace with a constraint to knee extension to a stop-jump without a knee brace. The stop-jump is a common task in basketball and volleyball, which involves an approach run, and a two-footed landing followed by a two-footed jump. Yu et al concluded that the newly designed knee brace with a constraint to knee extension significantly increased the knee flexion angle from a non-brace condition for both males and females during the landing of a stop-jump. The increased knee flexion angle should decrease the strain on the ACL. In addition, females were found to have significantly smaller knee flexion angles than males at the landing in a stop-jump for both brace and non-brace conditions. This means that females are at higher risk for ACL injury because their mechanics cause them to strain the ACL more than males. Females had significantly greater vertical and posterior ground reaction forces than males in the stop-jump task for both conditions. However, there was not a significant effect on the maximum posterior or vertical ground reaction forces for the braces. There was not an effect on the knee joint resultants, which was expected because they are determined from the ground reaction forces.

A common concern is that knee braces may affect the athlete's performance or comfort while playing sports. The mean running approach speed and jump height during this task were essentially the same with and without the brace for both male and female subjects participating in this study. This means that the knee brace did not positively or negatively affect the athlete's running or jumping performance.

Healthy recreational athletes participated in this study, and the results show that this knee brace may be used as a training tool to help prevent ACL injuries. However, it is not known what effects the knee brace with a constraint to knee extension will have on ACL reconstruction patients. Another limitation of that study is that it did not compare the lower extremity kinematics and kinetics of the knee brace with a constraint to knee extension with a traditional knee brace, so the effects seen may be due to bracing in general.

A study by DeVita et al (1998) compared lower extremity kinematics and kinetics while walking with and without a functional knee brace in patients recently after ACL reconstruction surgery. Patients were tested pre-surgery, 3 weeks post-surgery, and 5 weeks post-surgery. The walking speeds were the same with and without the knee brace so there was no effect on performance. While walking with the functional knee brace, patients used smaller extensor moments at the knee and greater extensor moments at the hip and ankle compared to walking without a brace. The patients compensate for the weak ACL by using the hip and ankle more than the knee during walking, which decreases strain on the ACL. One issue with this study is that the patients may have been tested too soon after surgery because the typical rehabilitation program takes about 6 months. In this study, patients used an extensor moment at the knee during stance phase, which would actually increase the load on the ACL. Other studies have shown that ACL deficient patients use a flexor moment at the knee during most of stance (Berchuck et al, 1990; Andriacchi and Birac, 1993). The authors believe that functional knee braces may lead to gait adaptations in patients who have undergone ACL reconstruction surgery, even though their study only tested the immediate effects of functional knee bracing.

DeVita et al (1996) tested healthy subjects to determine whether a functional knee brace could cause them to walk and run with similar torque and power patterns in rehabilitated ACL-injured patients. The functional knee brace caused an increase in extensor angular impulse at both the hip and ankle during walking and an increase only at the hip during running. There was not a difference in torque variables at the knee during walking or running. Therefore, the healthy individuals did similar increases in extensor torques at the ankle and hip during walking as seen in ACL-injured individuals. DeVita et al believe that the functional knee brace may be a cause of the altered joint torque patterns seen in ACL-injured gait.

In 1989 various NCAA Division 1 female basketball players were trained to perform cutting maneuvers in which the knee was flexed and the feet were kept under the hips. This training program aimed to increase the knee flexion angle and decrease knee valgus during cutting. The results of this study showed a decrease in ACL injuries by 89% during a two-year period (Griffis et al, 1989). This study proves that training can cause gait adaptations. Training with a knee brace with a constraint to knee extension may help ACL reconstruction patients regain normal gait patterns, and may help healthy individuals acquire a gait pattern that will decrease the risk of sustaining an ACL injury.

CHAPTER III

METHODS

3.1 Subjects

Twelve subjects with rupture of the anterior cruciate ligament participated in this study. One subject was tested 3.5 months after ACL reconstruction surgery, one subject was tested 6.5 months post surgery, and ten subjects were tested between 4-6 months post surgery. The age range was 18-32 with an average age of 22.4 years old. Six males and six females participated. Two of the ACL patients had a previous lower extremity injury that was fully healed at the time of testing (ACL tear on opposite leg and ACL tear in same leg). Twelve healthy subjects without any know lower extremity disorders were matched by age within one year and gender to the ACL reconstruction patients.

3.2 Functional Activities for Testing

The functional activities for testing were level walking, jogging, and stair descending. In the level walking and jogging, subjects were asked to walk or run at a self-selected speed with the testing leg landing on the force plate. They performed 4-5 steps of walking or jogging before and after the force plate. For stair descending, subjects walked on a staircase consisting of four steps with the lowest two steps connected to force plates. Each step increased in height by 0.178 m. Stair descending was followed by 4-5 steps of level walking. The order of the functional activities was randomized. The subject performed five successful trials of each of the three functional activities at a self-selected speed for each of the three conditions: (1) without a knee brace, (2) with a knee brace without a constraint to knee extension, and (3) with the newly designed knee brace with a constraint to knee extension. The order of the functional activities and conditions were randomized.

The newly designed knee brace was designed from an existing functional knee brace (4titude; dj Orthopedics, LLC, Vista, CA). The brace frame was made of 6061-T6 aluminum with upright upper thigh and lower calf cuffs. Hook-and-loop straps attached the brace to the leg (Figure 2). The newly designed knee brace uses a spring mechanism that applies a gradual increasing resistance to knee extension from 40° of knee flexion until 10° of knee flexion, at which point there is a rigid stop to prevent further knee extension. The resistive torque is adjustable with a maximum of 3.5 N·m at 10° of knee flexion. Knee braces with and without constraint were made for the right and left legs in the following sizes: extra small, small, medium, and large.

3.3 Kinematic and Kinetic Data Collection

Subjects were tested in the Motion Analysis Laboratory of the Center for Human Movement Science at the University of North Carolina at Chapel Hill. Written consent was obtained after reviewing the experimental equipment and procedures. Women wore a sports bra. Both men and women wore spandex shorts and their own shoes and socks. Each subject's height, weight, ACL reconstruction date, and affected leg were recorded.

Passive reflective markers were placed on the subject bilaterally at the anterior superior iliac spine, acromion process of the shoulder, lateral thigh, proximal shank, distal shank, and lateral malleolus. A marker was also placed on the L4-L5 joint. The marker set can be seen in Figure 3. Each subject was allowed to practice each activity until they felt comfortable performing the task.

Three-dimensional videographic and force plate data were collected for the subject for each activity. Eight infrared video cameras were used to record the real-time 3-D trajectories of reflective markers on the subject at a frame rate of 120 Hz. The video cameras were calibrated for a 2.5 m long × 1.5 m wide × 2.5 m high calibration volume. Ground reaction force signals from two Type 4060A Bertec force plates (Bertec Corporation, Worthington, OH) were collected by a desktop computer at a sample rate of 1000 Hz. The data were recorded by the Peak Performance Motus videographic and analog data acquisition system (Peak Performance Technology Inc, Englewood, CO). The videographic and force plate data were time-synchronized.

After testing all functional activities for the three conditions, additional passive reflective markers were placed bilaterally on the lateral condyle of the tibia, medial condyle of the tibia, and medial malleolus. The subject was asked to stand in the middle of the calibration volume while three-dimensional videographic data of all reflective markers were collected. These additional markers were used to estimate the locations of critical body landmarks that were needed for calculating joint centers. They were placed on the subject during a static trial because these markers would not be visible during the performance of the functional activities.

3.4 Data Reduction

The real-time 3-D coordinates of the markers were filtered through a Butterworth low-pass digital filter at estimated optimum cutoff frequencies as determined by Yu et al.

(1999). The 3-D coordinates of the medial and lateral femoral condyles and medial malleoli were estimated from the 3-D coordinates of the markers added during the standing trial. The knee joint center was defined as the midpoint between the medial and lateral femoral condyles, and the ankle joint center was defined as the midpoint between the medial and lateral malleoli. The 3-D coordinates of the hip joint centers were estimated from the 3-D coordinates of the reflective markers on the right and left anterior superior iliac spines and L4-L5 joint and from anatomical data (Bell, Pedersen, and Brand, 1990). The 3-D coordinates of the knee joint centers, ankle joint centers, and medial and lateral malleoli were used to define the shank reference frame. The 3-D coordinates of the knee joint centers, hip joint centers, and medial and lateral femoral condyles were used to define the thigh reference frame. The knee joint angles were defined as Euler angles of the shank reference frame relative to the thigh reference frame rotated in order of: (1) flexion/extension (z-axis), (2) varus/valgus (y-axis), and (3) internal/external rotation (x-axis). All signal processing and data reduction were performed using MotionSoft 3-D motion data reduction program package version 6.5 (MotionSoft Inc, Chapel Hill, NC). Joint resultants were determined from segment kinematics and ground reaction force data using an inverse dynamic procedure (Greenwood, 1988). Landmark velocities, joint angles, and joint angle velocities were also calculated for each trial. The electric signals from the force plates were converted into forces. The data were normalized for the stance phase of the braced leg for each trial.

3.5 Data Analysis

Lower extremity kinematics and kinetics for the braced leg were used for data analysis. The braced leg was the leg with the ACL reconstruction surgery, and the same leg was used for the healthy subjects as their age and gender matched ACL patients. Knee extension moments were normalized by body weight*body height, and ground reaction forces were normalized by body weight. A two-way analysis of variance with repeated measures for brace condition and gender as an independent variable was performed to test the first hypothesis. Three separate ANOVAs were conducted to test each functional activity. The dependent variables were knee flexion angle at initial contact, knee flexion angle at peak posterior ground reaction force, peak knee extension moment in stance phase, vertical ground reaction force at peak knee extension moment, posterior ground reaction force at peak knee extension moment, horizontal hip velocity at initial contact (for all three functional activities), and vertical hip velocity at initial contact (for stair descending). The hip velocities were analyzed to compare approach speeds before contact with the force plate because significantly different approach speeds may affect the knee flexion angles or ground reaction forces. Yu, Lin, and Garrett (2005) concluded that peak posterior ground reaction force, peak proximal anterior shear force, and peak knee extension moment during landing during a stop-jump task occurred at approximately the same time. These data were used to support the analysis of ground reaction forces at peak knee extension moment, and knee flexion angle at peak posterior ground reaction force. If the brace condition effect was significant, then the brace conditions were compared using a paired t-test. Hip velocities were also compared between ACL reconstruction patients and healthy subjects for each functional activity using a one-way ANOVA because different approach speeds may affect the dependent variables.

The second hypothesis focused on the comparison of the patients wearing the knee extension constraint brace to the healthy subjects without a knee brace. ACL reconstruction patients wearing the constrained knee brace were compared to healthy subjects without a knee brace for each dependent variable for each functional activity. An independent t-test was conducted for each dependent variable for each functional activity.

A Type I error rate of 0.05 was used. All statistics were conducted using SPSS v.11.

CHAPTER IV

RESULTS

4.1 Gender Differences

Due to the small sample size, the gender differences found in this analysis may not be entirely accurate. Previous research has shown that gender may have an effect on knee flexion angle and ground reaction forces as described in the literature review, so it was included primarily to determine if there was an interaction between brace condition and gender. In this study, there were no interaction effects between brace condition and gender. The goal of this study was to determine the brace effect, and not to determine if there was a gender effect.

4.2 Approach Speed

The approach speeds were compared between brace conditions and gender for ACL reconstruction patients (Table 1). There was a significant knee brace condition effect for walking approach speeds ($F=3.454$, $p=.035$). The walking approach speed with the constrained knee brace was significantly slower than the walking speed with no knee brace ($t=-4.224$, $p=.000$), and the approach speed with the non-constrained knee brace was significantly slower than with no knee brace ($t=-2.656$, $p=.012$). There were no significant differences in jogging approach speed between brace conditions ($F=0.133$, $p=.876$), but males jogged at a significantly faster speed than females ($p=.002$). The vertical and horizontal hip velocities were significantly faster for males during stair descending (p=.000). There was also a significant knee brace effect on horizontal hip velocity during stair descending (F=3.255, p=.043), but there was not a significant difference in vertical hip velocities over brace conditions. ACL patients wearing a constrained knee brace had significantly faster horizontal hip velocities than when wearing the non-constrained knee brace (t=-3.374, p=.002). ACL patients had significantly slower horizontal hip velocities when wearing the non-constrained brace compared to wearing no knee brace (t=2.469, p=.019). The difference in horizontal hip velocities between the constrained and no knee brace conditions was not significant (t=-1.581, p=.123).

There was not a significant difference in walking (t=1.148, p=.255) or jogging (t=0.999, p=.321) approach speeds between ACL reconstruction patients and healthy subjects. ACL reconstruction patients had significantly faster horizontal hip velocities during stair descending (t=-2.679, p=.009), but there was no significant difference in vertical hip velocities between groups (t=-0.541, p=.590).

4.3 Hypothesis #1

The newly designed knee brace will increase the knee flexion angle and decrease the peak ground reaction force at initial contact of level walking, jogging, and stair descending for ACL reconstruction patients. The following results deal with testing the first hypothesis and are partitioned by functional activity. Tables 2-7 show the results of the statistical analyses for the first hypothesis. Tables 8-10 give the average values and standard deviations of the dependent variables of interest.

4.3.1 Walking

Statistical analysis found a significant knee brace effect (F=17.498, p=.000) and gender effect (F=13.956, p=.000) for knee flexion angle at initial contact during walking, and there was no interaction effect between brace condition and gender. Males had significantly larger knee flexion angles at initial contact than females. There was a significant knee brace effect for both females (F=23.413, p=.000) and males (F=20.832, p=.000). The constrained knee brace significantly increased the knee flexion angle at initial contact for females for both the non-constrained knee brace (t=6.226, p=.000) and the no knee brace conditions (t=5.858, p=.000). The knee flexion angle at initial contact for females was –1.32° with no knee brace, 0.02° with the non-constrained knee brace, and 2.98° with the constrained knee brace. The constrained knee brace significantly increased the knee flexion angle at initial contact for males from both the non-constrained knee brace (t=4.537, p=.000) and the no knee brace conditions (t=6.649, p=.000). The knee flexion angle at initial contact for males was 1.48° with no knee brace, 1.60° with the non-constrained knee brace, and 5.34° with the constrained knee brace. There was not a significant difference in knee flexion angle at initial contact between the non-constrained knee brace and no knee brace conditions for either females (t=2.032, p=.058) or males (t=0.196, p=.847).

There was not a significant knee brace effect on knee flexion angle at peak posterior ground reaction force (p=.173) during walking. Males had significantly larger knee flexion angles at peak posterior ground reaction force than females (p=.012). There was no significant knee brace effect (p=.817) or gender effect (p=.384) on peak knee extension moment during the stance phase of walking for ACL reconstruction patients. There was no significant knee brace effect (p=.324) or gender effect (p=.727) on vertical ground reaction force at the peak knee extension moment during walking for ACL reconstruction patients.

There was a significant knee brace effect (F=3.976, p=.028) on posterior ground reaction force during peak knee extension moment during walking for ACL reconstruction patients. The posterior ground reaction force significantly decreased (t=2.719, p=.01) from 0.192 N/(BW) with no knee brace to 0.161 N/(BW) with the constrained knee brace. The posterior ground reaction force was significantly different between the constrained and non-constrained knee brace conditions (t=2.246, p=.031). There was no significant gender effect (p=.386) on posterior ground reaction force.

4.3.2 Jogging

Statistical analysis found a significant knee brace effect (F=64.914, p=.000) for knee flexion angle at initial contact during jogging, but there was no gender effect (p=.189). ACL reconstruction patients wearing the constrained knee brace had significantly larger knee flexion angles at initial contact when compared to the non-constrained knee brace (t=8.412, p=.000) and also the no knee brace (t=10.570, p=.000) conditions. No significant difference was found between the non-constrained knee brace and no knee brace (t=1.194, p=.241) conditions. The knee flexion angle at initial contact for ACL reconstruction patients was 7.05° with no knee brace, 7.62° with the non-constrained knee brace, and 11.4° with the constrained knee brace.

There was a significant knee brace effect (p=.005) and gender effect (p=.000) for knee flexion angle at peak posterior ground reaction force during jogging. There was a significant knee brace effect for both females (F=4.452, p=.029) and males (F=35.411, p=.000). Female

ACL reconstruction patients wearing the constrained knee brace had significantly larger knee flexion angles at peak posterior ground reaction force than when they did not wear a knee brace (t=3.067, p=.007). There were no significant differences between the constrained and non-constrained (t=1.907, p=.074) knee braces or between the non-constrained knee brace and no knee brace conditions (t=1.581, p=.132) for females. Male ACL reconstruction patients wearing the constrained knee brace had significantly larger knee flexion angles at peak posterior ground reaction force than when they wore the non-constrained knee brace (t=2.989, p=.008), and also when compared to the no knee brace condition (t=8.208, p=.000). There was not a significant difference between non-constrained knee brace and no knee brace conditions (t=1.5, p=.152). Females actually had larger knee flexion angles than males at peak posterior ground reaction force.

There was no significant knee brace effect (p=.523) or gender effect (p=.127) on peak knee extension moment during the stance phase of jogging for ACL reconstruction patients. There was no significant knee brace effect for vertical ground reaction force (p=.657) at peak knee extension moment during jogging. Females had significantly larger vertical ground reaction forces (p=.000) at peak knee extension moment than males during jogging. There was no significant knee brace effect for posterior ground reaction force (p=.618) at peak knee extension moment. Females had significantly smaller posterior ground reaction forces (p=.001) at peak knee extension moment than males.

4.3.3 Stair Descending

Statistical analysis revealed a significant knee brace effect (p=.000) and gender effect (p=.000) for knee flexion angle at initial contact during stair descending for ACL reconstruction patients. Males had significantly larger knee flexion angles at initial contact than females. There was a significant knee brace effect for both females (F=52.553, p=.000) and males (F=37.069, p=.000). The constrained knee brace significantly increased the knee flexion angle at initial contact for females from both the non-constrained knee brace (t=8.243, p=.000) and the no knee brace conditions (t=9.716, p=.000). The non-constrained knee brace also significantly increased the knee flexion angle at initial contact for females when compared to the no knee brace condition (t=4.516, p=.000). The knee flexion angle at initial contact for females was 6.73° with no knee brace, 9.64° with the non-constrained knee brace, and 13.64° with the constrained knee brace. The constrained knee brace significantly increased the knee flexion angle at initial contact for males from both the non-constrained knee brace (t=6.998, p=.000) and the no knee brace conditions (t=7.162, p=.000). There was not a significant difference for males between the non-constrained knee brace and no knee brace conditions for knee flexion angle at initial contact (t=1.084, p=.294). The knee flexion angle at initial contact for males was 13.69° wtih no knee brace, 14.45 ° with the non-constrained knee brace, and 18.17° with the constrained knee brace. There was no significant knee brace effect (p=.222) or gender effect (p=.062) for knee flexion angle at peak posterior ground reaction force during stair descending.

There was no significant knee brace effect (p=.187) on peak knee extension moment during stance of stair descending for ACL reconstruction patients. Males had significantly larger peak knee extension moments than females (p=.003). There was no significant knee brace effect (p=.960) or gender effect (p=.609) for vertical ground reaction force during peak knee extension moment during stair descending. There was no significant knee brace effect (p=.475) or gender effect (p=.067) on posterior ground reaction force during peak knee extension moment during stair descending.

4.4 Hypothesis #2

ACL reconstruction patients wearing the brace with a constraint to knee extension will have motion patterns similar to healthy subjects without a knee brace. The following results deal with testing the second hypothesis and are divided by functional activity. Table 11 shows the results of the statistical analyses performed for the second hypothesis. Tables 8-10 provide the average values and standard deviations of the ACL reconstruction patients over all brace conditions as well as the healthy subjects without a knee brace.

4.4.1 Walking

ACL patients wearing the constrained knee brace had significantly larger knee flexion angles at initial contact than healthy subjects in the no knee brace condition during level walking (t=5.032, p=.000). The average knee flexion angle at initial contact was 4.156° for ACL patients and –0.0275° for healthy subjects. There was not a significant difference between ACL patients and healthy subjects for knee flexion angle at peak posterior ground reaction force during walking (t=-0.462, p=.646).

There was no significant difference in peak knee extension moment during stance between ACL patients and healthy subjects during walking (t=1.392, p=.168). The difference between groups for vertical ground reaction force at peak knee extension was not significant (t=-1.956, p=.054). ACL patients and healthy subjects showed no significant difference for posterior ground reaction force at peak knee extension moment (t=1.738, p=.087).

4.4.2 Jogging

There was not a significant difference in knee flexion angles at initial contact during jogging between ACL patients wearing the constrained knee brace and healthy subjects not wearing a knee brace (t=0.34, p=.735). Healthy subjects had significantly larger knee flexion angles at peak posterior ground reaction force during jogging than ACL patients wearing the constrained knee brace (t=-6.363, p=.000). The knee flexion angle for healthy subject was 28.293°, and ACL patients had an average knee flexion angle of 18.943°.

There was a significant difference in peak knee extension moment during jogging between ACL patients and healthy subjects (t=8.08, p=.000). Healthy subjects had average peak knee extension moments of 0.0877 N·m/(BW*BH), and ACL patients had average peak knee extension moments of 0.0384 N·m/(BW*BH). There was not a significant difference in vertical ground reaction force at peak knee extension moment between groups (t=-1.918, p=.059). Healthy subjects (0.214 N/BW) had significantly larger posterior ground reaction forces at peak knee extension moment than ACL patients (0.159 N/BW) during jogging (t=2.674, p=.009).

4.4.3 Stair Descending

The results showed that there was a significant difference in knee flexion angles at initial contact between groups during stair descending (t=3.423, p=.001). The ACL patients (15.906°) actually had larger knee flexion angles at initial contact than the healthy subjects

(10.942°). There was not a significant difference in knee flexion angle at peak posterior ground reaction force between ACL patients and healthy subjects (t=0.026, p=.979).

The difference between groups for peak knee extension moment during stance was not significant (t=-0.346, p=.730). No significant difference in vertical ground reaction force at peak knee extension moment was found between the groups (t=0.663, p=.509). There was no significant difference in posterior ground reaction force at peak knee extension moment between ACL patients and healthy subjects (t=0.167, p=.868).

4.5 Levene's Test Explanation

Levene's test determines if there is a difference in error variance of the dependent variable across groups. If Levene's test is significant than there is no difference in error variance of the dependent variable between groups. Some analyses yielded non-significant values in Levene's test. However, if the sample sizes are equal, a non-significant Levene's test is generally overlooked. The sample sizes were equal for all of the analyses in this study. Therefore, a non-significant Levene's test was not a problem in this study. Just to be noted, the following tests had a non-significant Levene's test: horizontal and vertical hip velocities compared between ACL patients and healthy subjects for stair descending, hip velocities compared by condition and gender between ACL patients for jogging and stair descending, knee flexion angle at initial contact compared by condition and gender between ACL patients during walking, knee flexion angle at peak posterior GRF compared by condition and gender between ACL patients during stair descending, peak knee extension moment compared by condition and gender between ACL patients during jogging, vertical GRF at peak knee extension moment compared by condition and gender between ACL patients during stair descending, and posterior GRF at peak knee extension moment compared by condition and gender between ACL patients during all 3 activities.

CHAPTER V

DISCUSSION

5.1 Performance

The knee extension constrained brace and the non-constrained brace both affected the approach speeds during functional activities. ACL reconstruction patients wearing either knee brace walked significantly slower than when not wearing any knee brace. However, the knee braces did not significantly affect the approach speed during jogging. During stair descending, ACL patients wearing the constrained knee brace had significantly faster horizontal hip velocities than when wearing the non-constrained knee brace, and patients wearing the non-constrained knee brace had significantly slower horizontal hip velocities than the no knee brace condition. There were no significant differences in vertical hip velocities for the brace conditions. Theoretically, a faster approach speed should produce a larger knee flexion angle and larger ground reaction forces, so the differences in approach speed may have affected the dependent variables. The results of this study show that the knee braces may have some minor affects on approach speed, but these differences were not consistent across all of the functional activities. It cannot be concluded that knee braces affect performance during athletic competition because the functional activities were performed under a moderate pace consistent with activities of daily living.

Many studies have focused on the affect that functional knee braces may have on an athlete's performance. Studies show that functional knee braces increase energy expenditure

(Highgenboten et al, 1991; Zetterlund, Serfass, and Hunter, 1986) and intramuscular pressures, which may lead to fatigue and decrease blood flow to muscles (Styf, 1999). Wu, Ng, and Mak (2001) determined that functional knee braces slowed down running and cutting speeds. Other studies on the effects of functional knee bracing on performance have been inconclusive (Kramer et al, 1997). However, many ACL reconstruction patients who use functional knee braces report subjective improvements in knee stability, pain attenuation, and confidence during athletic activities (Wojtys, Kothari, and Huston, 1996; Kramer et al, 1997; Colville, Lee, and Ciulio, 1986). Functional knee braces are effective in preventing anterior tibial translation at low loads, but not necessarily under high loads that may occur during athletic performance (Beck et al, 1986; Branch and Hunter, 1990; France and Paulos, 1994). Research tends to show that functional knee braces are effective in decreasing ACL loading, but they may negatively affect an athlete's performance. The newly designed knee brace with a constraint to knee extension can still be used as a training tool during practice and taken off during athletic competition if it is believed to negatively affect an athlete's performance.

Some other limitations of functional knee braces are brace migration, bulkiness, discomfort, price, and the single axis of rotation. Braces that do not fit properly may migrate so that the hinge axis of the brace no longer lines up with the axis of the knee joint. The brace will not function properly if it is not placed correctly on the leg or if the size does not match the leg. Many manufacturers offer braces in two or three different lengths to allow for different inseam measurements (Paluska and McKeag, 1999). Many people complain that functional knee braces are too bulky to wear in everyday living. This issue can only be solved by the companies that design the braces, but if braces are proven to help than people may be more inclined to wear them. The majority of ACL reconstruction patients in this study felt that the knee braces were uncomfortable, and Styf (1999) reported that intramuscular pressures might increase due to bracing, which can cause discomfort. Only 3 out of 12 patients actually wore functional knee braces outside of this study, so the other patients may have felt the braces were uncomfortable because they were not used to wearing one. Functional knee braces can be expensive, which limits the number of people who choose to purchase them. Paluska and McKeag (1999) list off-the-shelf braces that cost the brace provider anywhere from $105-$525, and custom braces tend to be more expensive. The functional knee braces used in this study have a single axis of rotation, which does not truly model the knee joint. The knee joint is not a true hinge joint because there is some translation of the tibia and femur as the knee flexes. Therefore, this knee brace may affect the normal bending of the knee.

5.2 Hypothesis #1

The goal of the newly designed knee brace with a constraint to extension was to increase the knee flexion angle during landings of functional activities, which should decrease the load on the ACL. The results of this study partially support the first hypothesis in that the knee extension constrained brace increased the knee flexion angle during landings of walking, jogging, and stair descending. ACL reconstruction patients wearing the constrained knee brace had significantly larger knee flexion angles at initial contact than both the non-constrained and no knee brace conditions. The constrained knee brace increased the knee flexion angle at initial contact from the no knee brace condition by 4.3° for females and 3.86° for males during walking, 4.7° for females and 4.0° for males during jogging, and

6.91° for females and 4.48° for males during stair descending. These increases in knee flexion angle at initial contact due to the constrained brace may be underestimated because the approach speeds for both functional knee braces were slower than the approach speed without a knee brace during walking. These results are consistent with a previous study by Yu et al (2004). Yu et al studied the same constrained knee brace used in this study, and found that it significantly increased knee flexion angle at landing of a stop-jump task by 5.3° for females and 5.1° for males. In this study, males had a significantly greater knee flexion angles at initial contact than females during walking and stair descending, but the difference for jogging was not significant. The females in this study were on average 11 cm shorter than the males, and shorter people tend to have larger knee flexion angles when walking on stairs. Therefore, the gender difference in knee flexion angle at initial contact may have been underestimated for stair descending. Previous studies have found that males tend to have larger knee flexion angles at initial contact than females do, which puts females at a greater risk for ACL injury (Yu et al, 2004; Boden et al, 2000; Malinzak et al, 2001; Decker et al, 2003; James et al, 2004; Lephart et al, 2002).

The part of the first hypothesis that was not supported by the results was that the constrained knee brace would reduce peak ground reaction forces during stance phase. There were no significant knee brace effects for peak vertical ground reaction force. ACL patients wearing the constrained knee brace had significantly smaller peak posterior ground reaction forces when compared to both the non-constrained and no knee brace conditions during level walking. There were no other significant differences in brace condition for posterior ground reaction force during jogging or stair descending. Yu et al (2004) did not find any differences in peak vertical or peak posterior ground reaction forces between the constrained knee brace and no knee brace conditions. Females had significantly larger peak vertical ground reaction forces than males during jogging for all knee brace conditions, but not for walking or stair descending. Yu et al (2004) found that females had larger peak vertical ground reaction forces than males in a stop-jump task for both the constrained and no knee brace conditions. Perhaps females only have larger vertical ground reaction forces than males for high impact activities like running and jumping.

The results of this study showed that females had significantly smaller peak posterior ground reaction forces than males during jogging, but there were no significant differences in walking or stair descending. The reason that females had smaller peak posterior ground reaction forces in this study may be attributed to their slower approach speeds during jogging. These results do not agree with previous literature. Yu et al (2004) determined that females had significantly larger peak posterior ground reaction forces than males during a stop-jump task. Chappell et al (2002) reported that females had larger maximum anterior shear force at the proximal tibia than males at the landing of three stop-jump tasks. Posterior ground reaction force is a major contributor to anterior tibial shear force (Yu et al, 2004). A difference between this study and the Yu et al and Chappell et al studies is that the later both tested healthy recreational athletes, which may be a reason for this discrepancy.

The results of this study suggested that the knee braces did not significantly affect the peak knee extension moment during the stance phase of any of the functional activities that were tested. The peak knee extension moment is determined primarily by the ground reaction forces because the relatively small masses and moments of inertia of the foot and shank. There should not be any significant differences in peak knee extension moments since the ground reaction forces did not show significant differences by brace condition.

The maximum torque exerted by the constrained knee brace is 3.5 N·m of knee flexion moment at 10° of knee flexion. The largest peak knee extension moment during stance phase without a knee brace was recorded by patient 15 during jogging with a value of 9.89 N·m. The smallest peak knee extension moment was 0 N·m, which was experienced by patient 14 during walking. Patient 12 actually had a net knee flexion moment of 2.43 N·m during walking. When comparing the values for the peak knee extension moment and the torque applied by the constrained knee brace, it seems that the knee brace may have an effect on the patients' knee extension moments during functional activities. However, according to statistical analysis there was not a significant difference in peak knee extension moments between any of the knee brace conditions. Therefore, the knee flexion torque of the constrained knee brace probably did not have a large affect on the knee extension moment during stance.

5.3 Hypothesis #2

The second hypothesis that ACL reconstruction patients wearing the constrained knee brace would have similar motion patterns to the healthy subjects without a knee brace was not fully supported. The ACL reconstruction patients with the constrained brace had significantly larger knee flexion angles at initial contact than the healthy subjects without a knee brace for both walking and stair descending, but no significant difference was found in jogging. The opposite is true for knee flexion angle at peak posterior ground reaction force. Healthy subjects had significantly larger knee flexion angles at peak posterior ground reaction force than ACL reconstruction patients wearing the constrained knee brace for jogging, but no differences were seen in walking or stair descending.

Healthy subjects without a knee brace had significantly larger peak knee extension moments and posterior ground reaction forces at peak knee extension moments during jogging than ACL reconstruction patients. There were no significant differences in peak knee extension moment or posterior ground reaction force at peak knee extension moment for walking or stair descending. In addition, there were no significant differences between groups in vertical ground reaction force at peak knee extension moment for any of the functional activities.

These results suggest that the constrained knee brace may have improved ACL reconstruction patients' motion patterns such that they may be at a lower risk of ACL injury than the healthy subjects without a knee brace. The knee flexion angles at initial contact are smaller during walking and stair descending for ACL reconstruction patients with the constrained knee brace, which should decrease loading on the ACL even though there were no differences in ground reaction forces. During jogging, the knee flexion angle at initial contact was not different between groups, but ACL reconstruction patients with the constrained knee brace did have smaller peak knee extension moments and posterior ground reaction forces than the healthy subjects without a knee brace. Peak posterior ground reaction force and peak knee extension moment are both directly related to peak proximal tibial anterior shear force, which is directly related to ACL strain (Yu, Lin, Garrett, 2005). The results of this study show that ACL reconstruction patients have smaller posterior ground reaction forces and peak knee extension moments than healthy subjects without a knee brace during jogging, so they should have less ACL strain during jogging.

5.4 Future Studies and Limitations

Future studies are needed to fully understand the effectiveness of the knee brace with a constraint to extension. The results of this study showed the immediate effects of the constrained knee brace on knee kinematics and kinetics. Further research needs to be conducted on the long-term effects of training with the constrained knee brace to determine if people will adapt their gait to increase knee flexion angle even after they stop using the knee brace. The ACL reconstruction patients in this study were all in a post-surgery range of 3.5-6.5 months, but the patients were at different stages in their rehabilitation programs. The patients may have had different motion patterns due to these different stages of rehabilitation. The motion patterns of the uninjured knee of the ACL reconstruction patients may also change after surgery, so it may be beneficial to investigate the motion patterns of the uninjured knee during functional activities. The ACL reconstruction technique may also affect the post-surgery motion patterns. Future studies may look into the type of ACL graft and its effects on lower extremity kinematics and kinetics.

A limitation of this study was that even though there were 4 sizes of knee braces, the brace did not always fit tightly and comfortably. Brace migration may have occurred with some subjects, which may have affected their gait pattern. The constrained knee brace had a stop at 10° of knee flexion, but the knee flexion angles at initial contact during walking were less than 10°. This may be attributed to brace migration and marker movement. The brace may not stop the soft tissues from moving, and the reflective markers will continue to move when the soft tissues move. Another limitation of this study is that only sagittal plane biomechanics were studied. Sagittal plane biomechanics have a significant impact on ACL loading, but valgus-varus and internal-external rotation angles and moments will also affect

ACL loading. The functional activities tested in this study were activities of daily living and did not accurately simulate activity levels during athletic competition. Future studies should test the effect of the constrained knee brace on lower extremity kinematics and kinetics in athletic tasks that are common to non-contact ACL injuries.

5.5 Conclusions

The results of this study provide significant information on the effects of a knee brace with a constraint to knee extension on the lower extremity kinematics and kinetics during functional activities that occur in daily living. The newly designed knee brace did significantly increase the knee flexion angle at initial contact from the no brace condition as well as the non-constrained knee brace, which shows that this effect is not due to bracing, but can be attributed to the constraint itself. Even though the constrained knee brace did not cause the ground reaction forces to decrease, the loading on the ACL should still decrease because of the increase in knee flexion angle. The increased knee flexion angle at landing with the constrained knee brace may help to reduce ACL loading during walking, jogging, and stair descending if subjects are trained with the constrained knee brace. Further studies are needed to determine the long-term training effects of using a knee brace with a constraint to knee extension for prevention and rehabilitation programs for ACL injuries.

Table 1:

Comparison of Hip Velocities by Brace Condition and Gender for ACL Patients

		SS	df	MS	F	p
Walking	Brace Condition	2.11E-01	2,17	1.05E-01	3.454	0.035*
	Gender	9.76E-02	1,17	9.76E-02	3.199	0.077
	Interaction	3.46E-02	2,17	1.73E-02	0.567	0.569
Jogging	Brace Condition	2.42E-02	2,17	1.21E-02	0.133	0.876
	Gender	8.90E-01	1,17	8.90E-01	9.796	0.002*
	Interaction	5.58E-02	2,17	2.79E-02	0.307	0.736
Stair Descending (horizontal)	Brace Condition	7.05E-02	2,17	3.53E-02	3.255	0.043*
	Gender	1.74E-01	1,17	1.74E-01	16.036	0.000*
	Interaction	1.22E-02	2,17	6.10E-03	0.563	0.571
Stair Descending (vertical)	Brace Condition	3.70E-02	2,17	1.85E-02	1.344	0.265
	Gender	1.78E-01	1,17	1.78E-01	12.965	0.000*
	Interaction	1.46E-02	2,17	7.32E-03	0.532	0.589

*significant difference at 0.05 level

Table 2:

Knee Flexion Angle at Initial Contact for ACL Patients

		SS	df	MS	F	p
Walking	Brace Condition	340.389	2,17	170.195	17.498	0.000*
	Gender	135.75	1,17	135.75	13.956	0.000*
	Interaction	6.853	2,17	3.427	0.352	0.704
Jogging	Brace Condition	402.524	2,17	201.262	14.842	0.000*
	Gender	23.682	1,17	23.682	1.746	0.189
	Interaction	5.86	2,17	2.93	0.216	0.806
Stair Descending	Brace Condition	608.951	2,17	304.476	12.68	0.000*
	Gender	796.211	1,17	796.211	33.159	0.000*
	Interaction	15.933	2,17	7.967	0.664	0.517

*significant difference at 0.05 level

Table 3:

Knee Flexion Angle at Peak Posterior Ground Reaction Force for ACL Patients

		SS	df	MS	F	p
Walking	Brace Condition	89.174	2,17	44.587	1.787	0.173
	Gender	162.076	1,17	162.076	6.496	0.012*
	Interaction	0.0688	2,17	0.034	0.001	0.999
Jogging	Brace Condition	277.932	2,17	138.966	5.563	0.005*
	Gender	352.163	1,17	352.163	14.098	0.000*
	Interaction	1.404	2,17	0.702	0.028	0.972
Stair Descending	Brace Condition	122.04	2,17	61.02	1.53	0.222
	Gender	142.195	1,17	142.195	3.565	0.062
	Interaction	5.213	2,17	2.607	0.065	0.937

*significant difference at 0.05 level

Table 4:

Peak Knee Extension Moment during Stance for ACL Patients

		SS	df	MS	F	p
Walking	Brace Condition	7.15E-05	2,17	3.58E-05	0.203	0.817
	Gender	1.35E-04	1,17	1.35E-04	0.764	0.384
	Interaction	4.37E-05	2,17	2.19E-05	0.124	0.883
Jogging	Brace Condition	6.22E-04	2,17	3.11E-04	0.652	0.523
	Gender	1.13E-03	1,17	1.13E-03	2.371	0.127
	Interaction	7.92E-04	2,17	3.96E-04	0.831	0.439
Stair Descending	Brace Condition	1.10E-03	2,17	5.51E-04	1.706	0.187
	Gender	2.94E-03	1,17	2.94E-03	9.09	0.003*
	Interaction	5.83E-05	2,17	2.92E-05	0.09	0.914

*significant difference at 0.05 level

Table 5:

Vertical Ground Reaction Force during Peak Knee Extension Moment for ACL Patients

		SS	df	MS	F	p
Walking	Brace Condition	8.38E-02	2,17	0.042	1.139	0.324
	Gender	4.52E-03	1,17	0.005	0.123	0.727
	Interaction	7.07E-02	2,17	0.035	0.961	0.386
Jogging	Brace Condition	0.209	2,17	0.105	0.422	0.657
	Gender	5.817	1,17	5.817	23.55	0.000*
	Interaction	0.13	2,17	0.065	0.264	0.769
Stair Descending	Brace Condition	3.68E-03	2,17	1.84E-03	0.041	0.960
	Gender	1.18E-02	1,17	1.18E-02	0.264	0.609
	Interaction	2.76E-02	2,17	1.38E-02	0.307	0.736

*significant difference at 0.05 level

Table 6:

Posterior Ground Reaction Force during Peak Knee Extension Moment for ACL Patients

		SS	df	MS	F	p
Walking	Brace Condition	2.05E-02	2,17	1.02E-02	3.392	0.037*
	Gender	2.47E-03	1,17	2.47E-03	0.818	0.368
	Interaction	1.75E-03	2,17	8.77E-04	0.29	0.749
Jogging	Brace Condition	8.90E-03	2,17	4.45E-03	0.483	0.618
	Gender	0.114	1,17	0.114	12.336	0.001*
	Interaction	2.05E-02	2,17	1.03E-02	1.113	0.332
Stair Descending	Brace Condition	7.45E-03	2,17	3.72E-03	0.75	0.475
	Gender	1.70E-02	1,17	1.70E-02	3.422	0.067
	Interaction	1.26E-03	2,17	6.31E-04	0.127	0.881

*significant difference at 0.05 level

Table 7:

Repeated Measures ANOVA for Select Variables for ACL Patients

	Task	Gender	Pillai's Trace	F	df	p
Knee Flexion Angle at Initial Contact	Walking	Female	0.745	23.413	2	0.000
		Male	0.723	20.832	2	0.000
	Jogging	All*	0.792	64.914	2	0.000
	Stair Descending	Female	0.868	52.553	2	0.000
		Male	0.822	37.069	2	0.000
Knee Flexion Angle at Peak Posterior GRF	Jogging	Female	0.358	4.452	2	0.029
		Male	0.816	35.411	2	0.000
Posterior GRF at Peak Knee Extension Moment	Walking	All*	0.19	3.976	2	0.028

*no significant difference between genders

Table 8:

Walking Average Values

		ACL reconstruction Patient						Healthy	
		constrained		non-constrained		no brace		no brace	
		mean	std dev	mean	std dev	mean	std dev	mean	std dev
Knee Flexion Angle (degrees) at Initial Contact	Female*^	2.978	3.058	0.0216	3.804	-1.319	3.054	0.415	4.99
	Male*^	5.335	2.949	1.598	2.511	1.475	3.195	-0.47	2.222
	All**	4.156	3.193	0.8096	3.275	0.0783	3.391	-0.028	3.833
Knee Flexion Angle (degrees) at Peak Posterior GRF	Female	9.623	4.583	7.676	7.674	7.662	5.554	10.565	6.479
	Male	12.037	4.226	10.091	2.774	10.183	3.661	12.387	7.691
	All	10.83	4.513	8.883	5.818	8.922	4.809	11.476	7.069
Peak Knee Extension Moment (N·m/BW*BH)	Female	0.008	0.0105	0.008	0.0214	0.008	0.0119	0.018	0.0178
	Male	0.011	0.0125	0.008	0.005	0.011	0.0129	0.012	0.0212
	All	0.01	0.0115	0.008	0.0153	0.009	0.0123	0.015	0.0195
Posterior GRF (N/BW) at Peak Knee Extension Moment	Female	0.155	0.0408	0.179	0.0551	0.193	0.0533	0.199	0.0456
	Male	0.167	0.0341	0.197	0.0719	0.192	0.0651	0.161	0.056
	All*^	0.161	0.0376	0.188	0.0638	0.192	0.0586	0.18	0.0538
Vertical GRF (N/BW) at Peak Knee Extension Moment	Female	0.855	0.139	0.893	0.108	0.939	0.139	1.011	0.114
	Male	0.873	0.107	0.966	0.308	0.887	0.253	0.865	0.225
	All	0.864	0.123	0.929	0.23	0.913	0.203	0.938	0.191

*significant difference between constrained and no brace conditions
^significant difference between constrained and non-constrained conditions
"significant difference between non-constrained and no brace conditions
**significant difference between ACL patient constrained and healthy no brace conditions

Table 9:

Jogging Average Values

		ACL reconstruction Patient						Healthy	
		constrained		non-constrained		no brace		no brace	
		mean	std dev	mean	std dev	mean	std dev	mean	std dev
Knee Flexion Angle (degrees) at Initial Contact	Female*^	11.234	4.485	6.888	5.173	6.534	4.023	13.048	4.89
	Male*^	11.561	2.197	8.346	2.559	7.5596	2.643	9.101	3.121
	All	11.397	3.484	7.617	4.089	7.047	3.391	11.075	4.511
Knee Flexion Angle (degrees) at Peak Posterior GRF	Female*	20.588	6.234	18.418	6.772	16.944	5.075	29.825	6.629
	Male*^	17.298	3.427	14.663	3.819	13.154	3.615	26.761	7.709
	All	18.943	5.231	16.541	5.743	15.049	4.749	28.293	7.097
Peak Knee Extension Moment (N·m/BW*BH)	Female	0.042	0.0202	0.039	0.0235	0.04	0.0186	0.09	0.0221
	Male	0.035	0.0195	0.04	0.029	0.028	0.0182	0.085	0.0381
	All	0.038	0.0199	0.039	0.026	0.034	0.0193	0.088	0.0308
Posterior GRF (N/BW) at Peak Knee Extension Moment	Female	0.138	0.0752	0.119	0.0642	0.158	0.0902	0.212	0.0496
	Male	0.181	0.114	0.223	0.124	0.205	0.0949	0.217	0.0948
	All	0.159	0.0977	0.171	0.111	0.182	0.0944	0.214	0.0746
Vertical GRF (N/BW) at Peak Knee Extension Moment	Female	1.458	0.408	1.522	0.179	1.546	0.173	1.482	0.128
	Male	0.993	0.68	1.144	0.667	0.998	0.588	1.402	0.423
	All	1.226	0.601	1.333	0.518	1.272	0.5099	1.442	0.311

*significant difference between constrained and no brace conditions
^significant difference between constrained and non-constrained conditions
"significant difference between non-constrained and no brace conditions
**significant difference between ACL patient constrained and healthy no brace conditions

Table 10:

Stair Descending Average Values

		ACL reconstruction Patient						Healthy	
		constrained		non-constrained		no brace		no brace	
		mean	std dev	mean	std dev	mean	std dev	mean	std dev
Knee Flexion Angle (degrees) at Initial Contact	Female*^"	13.642	2.292	9.643	1.851	6.728	2.487	9.603	8.566
	Male*^	18.169	6.724	14.448	6.88	13.687	6.054	12.282	4.157
	All**	15.906	5.458	12.045	5.531	10.208	5.767	10.942	6.773
Knee Flexion Angle (degrees) at Peak Posterior GRF	Female	24.584	8.046	21.99	7.379	22.102	5.187	23.291	6.198
	Male	26.481	6.224	24.897	5.943	24.182	4.382	27.693	5.726
	All	25.533	7.154	23.444	6.765	23.142	4.849	25.492	6.29
Peak Knee Extension Moment (N·m/BW*BH)	Female	0.034	0.0256	0.025	0.0228	0.026	0.0159	0.026	0.0151
	Male	0.042	0.0139	0.037	0.0145	0.037	0.0102	0.047	0.0164
	All	0.038	0.0208	0.031	0.0198	0.031	0.0142	0.036	0.0189
Posterior GRF (N/BW) at Peak Knee Extension Moment	Female	0.002	0.0964	0.027	0.0727	0.022	0.0862	0.02	0.0867
	Male	0.034	0.0481	0.043	0.0634	0.05	0.038	0.022	0.0777
	All	0.018	0.0768	0.035	0.0677	0.036	0.0671	0.021	0.0811
Vertical GRF (N/BW) at Peak Knee Extension Moment	Female	1.396	0.223	1.411	0.182	1.43	0.233	1.358	0.287
	Male	1.461	0.177	1.419	0.262	1.421	0.179	1.427	0.222
	All	1.428	0.201	1.415	0.222	1.426	0.205	1.393	0.255

*significant difference between constrained and no brace conditions
^significant difference between constrained and non-constrained conditions
"significant difference between non-constrained and no brace conditions
**significant difference between ACL patient constrained and healthy no brace conditions

Table 11:

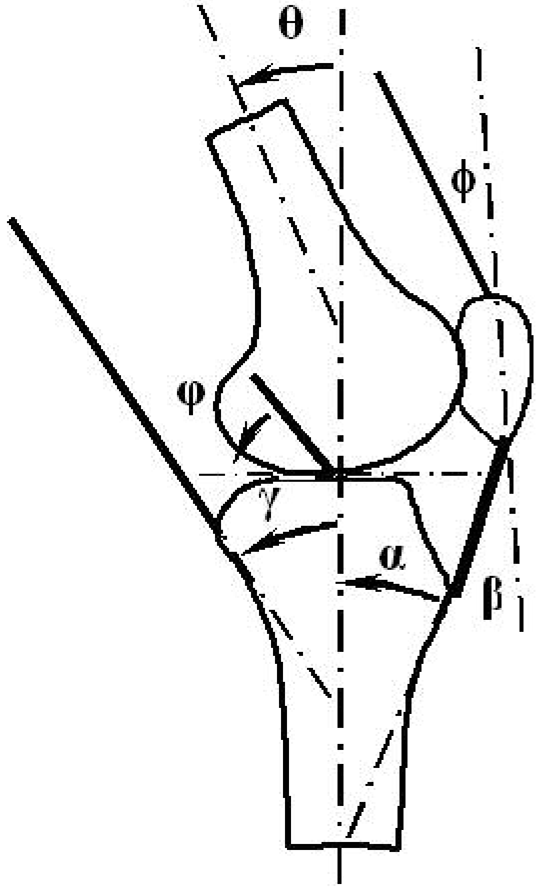
ACL Patients with a Knee Brace with a Constraint to Knee Extension Compared to Healthy Subjects Without a Knee Brace

		t	df	p
Knee Flexion Angle (degrees) at Initial Contact	Walking	5.032	1,70	0.000*
	Jogging	0.34	1,70	0.735
	Stair Descending	3.423	1,70	0.001*
Knee Flexion Angle (degrees) at Peak Posterior GRF	Walking	-0.462	1,70	0.646
	Jogging	-6.363	1,70	0.000*
	Stair Descending	0.026	1,70	0.979
Peak Knee Extension Moment (N·m/BW*BH)	Walking	1.392	1,70	0.168
	Jogging	8.08	1,70	0.000*
	Stair Descending	-0.346	1,70	0.730
Posterior GRF (N/BW) at Peak Knee Extension Moment	Walking	1.738	1,70	0.087
	Jogging	2.674	1,70	0.009*
	Stair Descending	0.167	1,70	0.868
Vertical GRF (N/BW) at Peak Knee Extension Moment	Walking	-1.956	1,70	0.054
	Jogging	-1.918	1,70	0.059
	Stair Descending	0.663	1,70	0.509
Horizontal Velocity (m/s) of Hips at Initial Contact	Walking	1.148	1,70	0.255
	Jogging	0.999	1,70	0.321
	Stair Descending	-2.679	1,70	0.009*
Vertical Velocity (m/s) of Hips at Initial Contact	Walking	N/A	-	-
	Jogging	N/A	-	-
	Stair Descending	-0.541	1,70	0.590

*significant difference at 0.05 level

Figure 1:

Patellar Tendon-Tibia Shaft Angle



α = patellar tendon-tibia shaft angle
θ = knee flexion angle

Figure 2:

DonJoy Knee Brace

Figure 3:

Marker Set

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