THE ROLE OF FUNCTIONAL KNEE BRACING IN A DYNAMIC SETTING

by

NEETU RISHIRAJ

B.P.E., The University of British Columbia, 1990

A THESIS SUBMITTED IN PARTIAL FULFILLMENT OF
THE REQUIREMENTS FOR THE DEGREE OF
MASTER OF HUMAN KINETICS

in

THE FACULTY OF GRADUATE STUDIES

SCHOOL OF HUMAN KINETICS

We accept this thesis as conforming
to the required standard

THE UNIVERSITY OF BRITISH COLUMBIA

NOVEMBER, 1995

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Department of HUMAN KINETICS

The University of British Columbia
Vancouver, Canada

Date 1995-12-17

DE-6 (2/88)

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ABSTRACT

The controversy over functional knee bracing still lingers. Although there is ample literature available on both the possible positive and negative concerns of utilizing a functional knee brace, there is very little information available on functional knee bracing in a dynamic setting. The purposes of this study were to evaluate the effect of functional knee bracing on athletic performance during dynamic testing in a non-injured knee joint and to measure the effects of functional knee bracing (under dynamic testing) on an Anterior Cruciate Ligament (ACL)--deficient knee.

A total of 60, 30 non-injured and 30 injured, subjects were tested with and without a functional knee brace. Each subject performed five functional tests--10 meter dash, figure-of-eight run, slalom run, hop test, and running down the stairs test. Each subject performed each test 8 times--two submaximal effort trials, followed by 6 trials (3 trials with and 3 trials without a brace) at maximal effort.

A 2X3, repeated measures on both factors, ANOVA was conducted to determine the results during the accommodation phase. A single factor ANOVA analysis was performed on the best performance measures after accommodation had occurred to the functional knee brace. Furthermore, a correlation analysis was conducted between knee joint laxity of injured subjects and their performance levels.

During the accommodation phase, the non-injured, braced group had statistically significant inferior performances (when compared to the non-injured, non-braced group) in the 10 meter dash, figure-of-eight, and the slalom tests and statistically superior performance in the hop test. In the running down the stairs test no statistically significant difference was noted between the two groups. However, once the subjects had accommodated to

the brace (best performance) no statistically significant difference was noted between the non-injured, braced and the non-injured, non-braced groups for any test.

As expected, during the accommodation phase, the injured, braced group performed statistically significantly better than the injured, non-braced group. However, after accommodating to the functional knee brace, an analysis of best performance data found no statistically significant difference between the two groups.

A strong correlation was not evident between the injured athlete's knee joint laxity and performance levels.

This study provides evidence that performance levels of non-injured, braced individuals is either only marginally hindered or is enhanced during the accommodation period when compared with non-injured, non-braced individuals. Once non-injured, braced individuals have accommodated to a functional knee brace they either perform at the same level or they outperform non-injured, non-braced subjects. These findings are an important consideration when considering a functional knee brace for prophylactic purposes. For injured individuals, performance levels are enhanced when a functional knee brace is utilized.

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ACKNOWLEDGMENTS

I would like to take this opportunity to acknowledge individuals who have contributed greatly to the development and the final successful completion of this thesis.

To Dr. D. B. Clement, Dr. R. Lloyd-Smith, Dr. W. Regan, and Dr. R. Woollard who shared their excellent creative ideas and provided hours of support and valuable constructive criticism.

To my family, especially my mother and father, a big thank you for your continued support and guidance. A special acknowledgment to my mother, who even with her illness and injuries, always continued to provide love and encouragement. I wish them all good health in the future.

To MSM and TSJ, my deepest appreciation for their patients, understanding, and assistance. Many times you assisted me in completing tasks that I could not have possibly completed by myself. I thank you from the bottom of my heart.

To Mr. D. Taylor at Generation II Orthotics Inc. who not only supplied the "ideal" fitting braces but also provided excellent information on brace design.

Lastly, but most importantly, to Dr. J. E. Taunton, my thesis advisor, whose vast talents, knowledge, and expertise taught me the skills, knowledge, and patients that has allowed me to complete this degree. His enthusiasm, assistance, and high standards are qualities that I have always respected and honored and I hope to someday parallel in my future ventures.

DEDICATED TO MY FAMILY!

INTRODUCTION

One of the impacts of health-care is the increased attention being paid to preventative medicine. For example, in an attempt to reduce ankle injuries cloth tape is used by many players participating in sports. It is estimated that at the University of Utah 16,000 rolls, or 720,000 feet, of tape are used in a single football season--requiring approximately 1200 to 1500 trainer hours to apply the tape. Therefore, it has been estimated that a Division I college football program spends at least $16,000 per year just for 1.5 inch cloth tape, which is primarily used for ankle taping (Burks et al., 1991).

The risk of knee injuries varies between sports, but is recognized as being particularly high in collision sports--North American Football and Ice Hockey. The overall portion of football players who sustain an injury is estimated as being high as 81%. Of these injuries, 13% to 36% will be concentrated at the knee joint (Jackson et al., 1991; Thompson, 1987; and Olson, 1989). In Jackson's et al. (1991) study on football players, of the 32.6% of the injuries concentrated at the knee joint, 20.3% were classified as being minor (<7 days needed to return to activity), 4.3% were classified as being moderate (7-21 days needed to return to activity), and 8.0% were classified as being major (>21 days needed to return to activity). Of these knee injuries the anterior cruciate ligament is injured more often than the medial collateral ligament, lateral collateral ligament, and the posterior cruciate ligament. Montgomery and Koziris 1989, state that the knee accounts for more than 20% of all American football injuries as well as the greatest loss of participation. Furthermore, these authors state that surgery is required for about 19% to 22% of the players with knee injuries.

In Ice Hockey, injuries concentrating at the knee joint range between 14% and 42% of all the injuries sustained by participants. In 1987, Pforringer and Smasal reported that of the 22% of the injuries concentrated to the lower extremity, 42% were localized at the knee joint. Tegner and Lorentzon (1991), reported 13.2% of the total 285 injuries were localized at the knee joint--the second highest body part to be injured. Furthermore, in their report these authors state that 42% of the Swedish Ice Hockey (elite) League players had sustained a knee injury. Pelletier et al., (1993), provided data on a 6-year study that showed that 18.6% of the injuries were localized a the knee joint. Lastly, a four year study involving The University of British Columbia (UBC) ice hockey players found that of the injuries sustained 14% (third most prevalent body part) were concentrated at the knee joint (Rishiraj et al., 1995).

This 'epidemic' has prompted investigations aimed at identifying factors which will decrease knee injury rates (Thompson, 1987). Previous knee bracing studies have focused on subjective testing, subjective and objective testing, clinical testing, cadaveric experiments, and biomechanical testing. A deficiency with all these studies is the lack of knee brace testing in a dynamic setting. The focus of this study is the testing of a functional knee brace in a dynamic setting and to accomplish this goal the discussion will concentrate on the knee joint anatomy, contribution of ligaments and muscles to knee stability, mechanisms of acute knee injury, diagnosis and management of knee ligament injury and conclude with a discussion of the role of knee bracing in athletics.

HYPOTHESES

(1) Performance times for the non-injured athletes will initially deteriorate during accommodation, but once accommodation has occurred to the brace the performance measures will return to the non-braced levels.

(2) Anterior cruciate ligament (ACL)-deficient (injured) athletes will see an improvement in performance measures, for all 5 functional tests, as a result of utilizing a functional knee brace.

(3) The performance level improvement will be more evident in the straight run test (10 meter dash) as compared to the other four tests.

(4) A positive correlation will be noted for injured athlete's knee joint laxity (as measured by the KT-1000) and their performance measures--the greater the laxity exhibited, the lower performance levels.

GROSS ANATOMY

Introduction

The knee is the largest joint in the entire body. It consists of three articulations. The knee consists of two condyloid joints, one between each condyle of the femur and the corresponding condyle of the tibia, and a third between the patella and the femur. The stability of the knee is dependent on: (a) an intricate group of strong ligaments; and (b) the supporting muscles and tendons. The motions of the knee joint are an extensive flexion-extension rotatory gliding movement with limited rotation of the tibia upon the femur.

The bony components which make up the knee joint are connected by the following structures:

Articular capsule	Anterior cruciate ligament(ACL)
Ligamentum patella	Posterior cruciate ligament(PCL)
Oblique popliteal ligament of Winslow	Medial and lateral menisci[1]
Tibial (medial) collateral ligament (MCL)	The transverse ligament
Fibular (lateral) collateral ligament (LCL)	The coronary ligament.

The stability of the knee is helped by a group of muscles which include the biceps femoris, the semitendinosus, semimembranosus, popliteus, gastrocnemius, sartorius, gracilis, and the quadriceps. The latter ends as the patellar tendon inserting on the tibial tubercle.

The knee is the articulation between the femoral and the tibial condyles. The fibula does not enter the articulation, while the patella (contained within the quadriceps tendon) does. Both bones are joined together by a fibrous capsule and ligaments. The capsule is replaced anteriorly by the quadriceps muscle, its tendon, the patella, patellar tendon

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1 The menisci provide support for the knee joint (Burks, 1990 and Savastona, 1980).

and the retinaculum. It is reinforced medially by the MCL and the pes anserinus, and on the lateral side by the LCL and the iliotibial band (ITB). Posteriorly, the joint is strengthened by the oblique popliteal ligament of Winslow (which is an expansion of the semimembranosus) and by the origin of the double headed gastrocnemius.

The knee is not a true hinge joint because a certain amount of rotation occurs between the femur and the tibia during extension and flexion; due to the configuration of the medial condyle of the femur and the cruciate ligaments acting as guide ropes (Savastona, 1980).

The distal femur is expanded into two condyles whose long axes are not parallel. The medial condyle is obliquely placed and curved medially, its articular surface being narrow. The articular surface of the lateral condyle is wider in the sagittal plane. Both articular surfaces are complex curves, highly curved posteriorly but much less so anteriorly, especially on the medial condyle (Savastona, 1980).

The expanded upper end of the tibia is formed by the medial and lateral tibial condyles. The articular surfaces of the two condyles form a nearly horizontal plateau. It presents two oval, slightly concave facets. Between the two articular surfaces is an area surmounted by the intercondylar eminence composed of medial and lateral tubercles (Savastona, 1980).

The patella is the largest sesamoid bone of the body. It is embedded within the patellar tendon and is triangular. Its articular surface is unequally divided into larger (lateral) and smaller (medial) parts by a longitudinal ridge.

Medial Knee Anatomy

The supportive structures on the medial side of the knee consist of three layers (Warren & Marshall, 1979). Layer I, the most superficial, is the extension of the deep fascia covering the quadriceps and continues as the deep fascia of the leg. It invests the sartorius and serves as that muscle's insertion, unlike the discrete tendons of insertion of the underlying gracilis and semitendinosus. Layer II, is the superficial MCL. Layers I and II blend approximately 1-2 cm anterior to the leading edge of the superficial MCL; these fibers join with the fibers from the vastus medialis to form the medial patellar retinaculum. Layer I completely covers the medial aspect of the knee and is infrequently torn with injury (Burks, 1990). Therefore, this layer usually needs to be incised to find the underlying pathology. Incising layer I on the more posterior aspect of the knee allows it to be separated from the superficial MCL. The only area on the medial aspect of the knee where all three layers can be found together is directly over the superficial MCL (Burks, 1990, and Warren & Marshall, 1979). The gracilis and the semitendinosus run between layers I and II, and they insert distal to the tibial tuberosity. They overlie the tibial attachment of the superficial MCL. Although these tendons have discrete insertions on the tibia, they also have attachments to the deep fascia. These tendons need to be cut (proximally) in order to harvest the tendons for use in knee reconstructive procedures (Burks, 1990).

Layer II (superficial MCL), originates at the medial femoral epicondyle. It runs approximately 10-11 cm to its tibial insertion, where it is covered by the gracilis and semitendinosus (Burks, 1990; Warren & Marshall, 1974; and Brantigan & Voshell, 1943 and 1941). It has been called either the MCL, the tibial collateral ligament, or the superficial collateral ligament (Burks, 1990 and Kennedy & Fowler, 1971). Posterior to the long vertical fiber of the superficial MCL, layers II and III merge (see below for information on layer III). Along with the semimembranosus tendon and sheath, they form the posteromedial corner of the knee.

As the knee flexes, the femur moves posteriorly on the tibia, and the superficial MCL slides posteriorly over the proximal tibia, helping to maintain a more uniform tension in the fibers (Burks, 1990; Basmajian & Slonecker, 1989; and Barntigan & Voshell, 1943 and 1941). The posterior sliding is accentuated by external rotation of the tibia in relation to the femur. Because of this, there can be no meniscal attachment to the superficial MCL, since this would impede its change in position (Muller, 1983).

The posterior medial corner of the knee is an area of blending of layers II and III. This area is confluent with the posterior edge of the superficial MCL; it runs obliquely to the tibia and has been termed the "posterior oblique ligament" by Hughston and Eilers (1973). It has also been referred as the oblique portion of the tibial collateral ligament (MCL) or simply, the posteromedial corner (Burks, 1990). Hughston and Eilers (1973), reported that the origin of this ligament is from the adductor tubercle, slightly posterior and proximal to the femoral epicondyle, and proposed it as a distinctly separate ligament. However, Warren and Marshall (1979), were unable to identify a discrete separate ligament, and since the fibers are in the same layer as the superficial MCL, they preferred to call it the "oblique fibers of the superficial medial collateral ligaments."

The attachment sites of the fibers in this area move towards each other with increasing flexion. Hughston and Eilers (1973), measured this distance to be 8-18 mm with a progressively decreasing distance with knee flexion. The more posterior proximal fibers also move underneath, or they move deep to the more anterior fibers with knee flexion (Bartel et al., 1977). Burks

(1990), points out that the oblique fibers are reinforced by the semimembranosus and its tendon sheath.

Layer III is the capsule of the knee and attaches primarily to the articular margins (Warren & Marshall, 1979). It is thin anteriorly and provides little stability to the knee. The part of the capsule that holds the meniscal rim to the tibia is called the coronary ligament. It is short and holds the meniscus tighter in relationship to the tibia than to the femur (Burks, 1990). Beneath the superficial MCL, this layer is thickened by the deep MCL. It has also been named the deep medial ligament, deep collateral, or middle capsular ligament (Warren & Marshall, 1979, 1974 and Slocum et al., 1974). The deep MCL may be divided into the meniscofemoral and meniscotibial ligaments, which run from the medial meniscus to the femur and from the meniscus to the tibia, respectively (Burks, 1990). Layers II and III are readily separable at their midpoint. However, approximately 1-2 cm behind the anterior edge of the superficial MCL, layers II and III blend into the posteromedial corner of the knee. The peripheral fiber system of the medial meniscus is intimately blended with this area, but many of the capsular fibers run uninterrupted from the femur to the tibia (Muller, 1983; Brantigan & Voshell, 1943, 1941). The capsule posterior to the oblique fibers of the superficial MCL is redundant with knee flexion. An arthrotomy to gain access to the posterior aspect of the knee should be made in this redundant capsule, avoiding the oblique fibers of the superficial MCL (Burks, 1990).

The semimembranosus and its tendon sheath are important contributors to the posteromedial corner anatomy (Burks, 1990). The tendon is described as having five arms of insertions. The first is a direct attachment to the posteromedial tibia just below the joint line. The second direct attachment proceeds anteriorly just beneath the superficial MCL. A third arm, more from the tendon sheath, runs to blend with the posteromedial capsule. Fourth contributes substantially to the oblique popliteal ligament which runs over the posterior surface of the joint capsule. The fifth arm blends in with the superficial MCL distally (Burks, 1990).

Lateral Knee Anatomy

Seebacher et al. (1982), divided the lateral side of the knee into three layers, as Warren and Marshall (1979) did on the medial side. Hughston et al. (1976), prefer to divide it into three areas from anterior to posterior (Burks, 1990). Both systems may be useful in organizing the anatomical areas of importance, but the emphasis of this paper will be on the layer approach.

The superficial layer is the deep fascia of the thigh and the calf, with the laterally condensed fibers that make up the iliotibial tract. This layer is continuous with the prepatellar bursa to the fascia over the popliteal fossa. The iliotibial tract is connected with the intermuscular septum down to the supracondylar tubercle of the femur. It then continues free of connection until it inserts on Gerdy's tubercle (Burks, 1990 and Kaplan, 1958). There are capsuloosseous attachments of the iliotibial tract that Terry et al. (1986), reports as being important for lateral knee stability, and have termed them "anterolateral ligament of the knee." Some fibers sweep anteriorly to the lateral border of the patella, and they join with fibers of the vastus lateralis to form the lateral retinaculum (Reider et al., 1981 and Kaplan, 1958). Confluent with this layer is the biceps femoris tendon, which lies posteriorly and is considered in the superficial layer much as the sartorius in on the medical side (Burks, 1990). The biceps femoris tendon has a complex lateral insertion, and Marshall et al. (1972), found three layers. The biceps femoris tendon fibers insert to the:

(a) Gerdy's tubercle,

(b) Blend in with the crural fascia of the leg

(c) Loop around the LCL, and insert on the styloid process and the head of the fibula.

The second layer, laying deep to the superficial fascia is the retinaculum of the quadriceps and the patellofemoral ligaments (Seebacher et al., 1982). The patellofemoral ligaments run from the patella to:

(a) The terminal fibers of the intermuscular septum

(b) The lateral epicondyle

(c) The posterolateral capsule (Burks, 1990).

Reider et al. (1981), found that the more a patella tended towards a Wiberg III, the larger the lateral patellofemoral ligament would be. It was felt that this helped explain the greater tendency towards subluxation with the Wiberg III patella. The patellomeniscal ligament runs roughly parallel to the patellar tendon and attaches to the margin of the lateral meniscus and terminates at Gerdy's tubercle (Seebacher et al., 1982).

The third and deepest layer comprises of the lateral capsule as well as the LCL. This arrangement, therefore, differs from that on the medial side. The capsule attaches primarily to the articular margin region. The lateral meniscus is held to the tibia by the coronary ligament which is longer than the medial meniscus. This allows for greater movement of the lateral meniscus. The coronary ligament at the posterolateral corner may attach to the head of the fibula (Burks, 1990). The lateral meniscus has a bare area, or hiatus, of no capsular attachment; this area is approximately at the midpoint of the lateral meniscus and averages 1.3 cm in length (Cohn & Mains, 1979).

Just posterior to the midpoint of the joint, the capsule divides into two laminae (Seebacher et al., 1982). The superficial lamina encompasses the LCL, and it ends posteriorly in a variably sized fabellofibular ligament. It is because of this ligament that Seebacher et al. (1982), consider the LCL as part of the deepest or capsular layer. The LCL runs from the lateral epicondyle of the femur to the proximal lateral aspect of the fibular head. Because of its location behind the axis of rotation, the LCL is tightest in extension but relaxes in flexion, especially at angles greater than 30° (Burks, 1990). In the past the LCL has been given a minor role in varus stability; however, now it is believed to be the primary restraint to varus stress (Burks, 1990).

The popliteus originates just distal and slightly posterior to the femoral attachment of the LCL. It has a firm connection to the posterior horn of the lateral meniscus and to the arcuate ligament as well (Burks, 1990). The muscle runs extra-synovially within the joint and obliquely over the posterior proximal tibia and is covered by its own fascial investment on the lateral femoral condyle. It then inserts at the lateral femoral condyle, just distal and slightly posterior to the femoral attachment (Cohn & Mains, 1979).

Cruciate Ligament Anatomy

Histologic Anatomy

The anterior and posterior cruciate ligaments are intracapsular but extrasynovial ligaments. These ligaments appear crossed on viewing the knee anteriorly or laterally (Burks, 1990).

The ACL ligament is composed of collagen fibrils, which appear parallel at high magnification. These fibrils form fibers that run parallel to the long axis of the ligament. Many collagen fibers merge together to make the subfascicular unit. In humans, the amount of endotenon is great, thus giving the ligament a bundle-like and a less uniform appearance. Synovium covers the ligament, thus making it extrasynovial (Danyluck, Finlay, and Kreck, 1978).

An important aspect of the cruciate ligament anatomy is the change from the flexible ligamentous tissue to rigid bone, mediated by a transitional zone of fibrocartilage and mineralized cartilage (Arnoczyk, 1983). This helps prevent stress contraction at the attachment site by allowing a gradual change in stiffness (Arnoczyk, 1983).

Vascular Anatomy

The predominant source of blood supply is the middle geniculate artery, which leaves the popliteal artery and directly pierces the posterior capsule (Arnoczky, 1985, 1983 and Alm & Stromberg, 1974). The cruciates have a tree-like appearance of capillary vessels which penetrate the ligament transversely. There is significant blood supply from the fat pad via the inferior medial and lateral geniculate arteries, which may play a more important role when the ligament is injured (Burks, 1990 & Arnoczky et al., 1979).

Neurologic Anatomy

Nerve fibers, of the size most consistent with transmitting pain, are readily visualized in the intrafascicular spaces occupied by the vessels (Schutte et al., 1987 and Kennedy et al., 1974). These are presumably terminal branches from the tibial nerve in the popliteal fossa (Kennedy et al., 1974). Schultz et al. (1984), investigated mechanoreceptors in cruciate ligaments and found a few thin axons in the substance of the ligament, as well as bundles of axons running on the surface of the ligament. Mechanoreceptors were identified in the ligament and were felt to be similar to golgi tendon organs. They were mostly on the surface of the ligament, well beneath the synovial lining and primarily at the insertion sites. They were postulated to respond as proprioceptors and to signal potentially injurious deformation of the ligaments and joint. Schutte et al. (1987), found three morphological types of mechanoreceptors as well as free nerve endings. The three mechanoreceptors were as follows:

(1) Ruffini endings, are slow-adapting and respond to slight changes in ligament tension

(2) Ruffini mechnoreceptors, are also slow-adapting, resembling a golgi tendon organ

(3) Pacinian corpuscle, is a rapidly adapting mechanoreceptor.

Free nerve endings for transmitting pain were also identified but were far more scarce than the mechanoreceptors (Burks, 1990).

Insertion Site Anatomy

The bony attachments of the cruciates have been investigated by several authors. Odensten & Gillquist (1985), report that the ACL femoral attachment is oval and therefore wider.

All authors agree that the femoral attachment is oriented primarily in the longitudinal axis of the femur, and the tibial attachment is oriented in the anteroposterior axis of the tibia. This arrangement leads to the well-known twist of the ACL fibers when the knee moves from extension to flexion (Burks, 1990). There is also a twist of the ACL fibers in the coronal plane with external rotation of the fibers by approximately 90° as they approach the tibial surface (Burks, 1990 and Odensten & Gillquist, 1985). In 1986, van Rens et al., reported that in dogs, cutting all of the ligaments except the ACL and letting the tibia hang free resulted in a 180° derotation in a normal ACL. The same relationship is found in a human knee, with 90° twist of the fibers described by Burks (1990) and Odensten and Gillquist (1985).

The ACL tibial attachment fans out and forms a "foot" region. This allows the ACL to tuck under the roof of the intercondylar notch. This unique attachment of the ACL causes concern for certain ACL reconstruction techniques as the graft might be predisposed to impingement on the roof of the intercondylar notch. (Burks, 1990). The fibers of a normal ACL are able to slip under this point as a result of their sweeping nature (Burks, 1990).

The PCL femoral attachment as determined by Girgiset al. (1975), is half-moon shaped. Burks (1990), found that the femoral attachment of the PCL to be 21 mm by 10 mm with the longitudinal axis in the anteroposterior plane of the femur. However, the attachment is positioned more at the apex of the intercondylar notch, and less so on the inner wall. If the meniscofemoral ligaments are included, the femoral attachment appears larger, closer to the articular cartilage, and more on the inner wall. The tibial attachment of the PCL is below the level of the joint (in the middle or posterior tibia) and is rectangular. It is important to note that although the tibial attachment of the PCL is significantly below the joint surface, it is intraarticular to the posterior capsular attachment (Burks, 1990 & Hughston et al., 1980).

Fiber Orientation

Many authors have described separate bands in the ACL and the PCL (Burks, 1990; Furman et al., 1976; and Girgis et al., 1975). For the ACL, the bands are called anteromedial and posterolateral, with some including an intermediate band (Norwood & Cross, 1979). Although there is some disagreement on the actual anatomic division of the ligament, there is agreement that ACL's have "functional bands" so that tension is varied among the fibers in the ligament with range of motion (Arnoczky, 1983).

For the ACL, the anteromedial part is tighter in flexion and the posterolateral part is tighter in extension (Furman et al., 1976 & Girgis et al., 1975). Hughston et al. (1980), refer to the PCL as having an anterolateral band and a posteromedial band. In extension, the posterior fibers are taut, whereas the bulk of the ligament is relaxed. Conversely, the more posterior fibers become lax in flexion, whereas the remainder (anterolateral) become taut.

FUNCTIONAL ANATOMY

Ligaments

Ligaments function to limit joint motion and as static joint stabilizers. Disruption or sectioning of ligaments alone, or in combination, alters the limits of knee motion in a predictable way.

To describe motion limits properly, both the force (or moment) plus the displacement that results must be measured. Stiffness, by definition, is the slope of the force-displacement curve (change in force per change in displacement) at a given point. The non-linear relationship between applied force and displacement requires that the measured displacement be described with the applied force. Displacement and rotation values after sectioning a specific ligament often vary considerably. Factors contributing to these discrepancies include level of applied force, testing device, specimen condition, and specimen number (Burks, 1990).

Butler et al. (1980), introduced the concept of primary and secondary restraints to motion in a specific direction. A primary restraint is that structure which accounts for the majority of ligamentous force resisting an externally applied force. A secondary restraint provides a lesser contribution. Sectioning a primary restraint typically results in an increase in joint motion. Isolated disruption of a secondary restraint (in the face of an intact primary restraint) will not result in altering the limits of joint motion; whereas, sectioning both primary and secondary restraints will alter joint motion.

Specific ligaments may be considered as primary and secondary restraints. A ligament may function as a primary restraint to motion in one direction and a secondary restraint in another direction (Butler et al., 1980).

Anterior Cruciate Ligament

Primary Function

The ACL functions as the primary restraint to limit anterior tibial displacement (Fukubayashi et al., 1982). ACL sectioning results in greater anterior displacement in 30° of flexion than in 90° of flexion. The ACL offers no restraint to posterior tibial displacement (Nielsen & Helmig, 1985 and Shoemaker & Markolf, 1985). Daniel et al. (1988), tested 65 specimens with the KT-1000. Sectioning the ACL increased anterior displacement from 2.0 mm to 13.0 mm with a mean of 6.7 mm.

Secondary Function

The ACL functions as the secondary restraint to tibial rotation . Isolated ACL sectioning increased tibial rotation 38% (3° to 4°) at full extension (Markolf et al., 1976). Combined MCL-ACL sectioning yielded increases in tibial rotations that were larger than those changes resulting after sectioning these structures individually. Based on the available data (Markolf et al., 1981; Seering et al., 1980; Shoemaker and Markolf, 1986; and Markolf et. al., 1976), the ACL probably functions as:

(1) A major secondary restraint to internal rotation

(2) A minor secondary restraint to external rotation.

The relative contribution of the ACL in restraining rotation is greater in full extension than it is in early (20° to 30°) flexion (Markolf et al., 1976).

The ACL functions as a minor secondary restraint to varus-valgus angulation at full extension. No significant changes were noted from 30° to 90° of flexion with the ACL sectioning (Markolf et al., 1976). Changes in stiffness about the neutral position and varus-valgus angulation following combined MCL-ACL section were similar to those changes noted after isolated MCL sectioning. Changes after combined LCL-ACL sectioning were moderately larger than those noted following isolated LCL sectioning. Although, no breakdown between varus and valgus angulation was reported, these findings suggest that the ACL offers little additional restraint to valgus angulation (and little restraint to varus angulation) beyond that afforded by the primary stabilizers--MCL and LCL (Grood et al., 1981; Seering et al., 1980; and Markolf et al., 1976).

Posterior Cruciate Ligament

Primary Function

The PCL is the primary restraint to posterior tibial displacement. The increase after PCL sectioning is greater at 90° of flexion than at 30° of flexion (Grood et al., 1988 & Gollehon et al., 1987).

Secondary Function

The PCL acts as a minor as well as a major secondary restraint to external rotation but, does not appear to limit internal rotation (Markolf et al., 1976). However, PCL sectioning performed after sectioning the LCL and deep posterior capsule increased the limits of external rotation when the knee was flexed more than 30°. Grood et al., (1988), confirmed these findings and added that increases in external rotation limits due to sectioning of both PCL and deep posterior lateral structures were largest at 90° of flexion and minimal at 0° and 15° of flexion. The two studies differ slightly in the structures sectioned, but both indicated that PCL functions as a minor secondary restraint to external rotation at full extension and as a major secondary restraint to external rotation at 90° of flexion.

The PCL offers little (Markolf et al., 1976) to no (Gollehon et al., 1988 and Grood et al., 1987) resistance to varus-valgus angulation with the collateral ligaments intact. If the lateral structures are sectioned, sectioning of the PCL does increase the varus angulation (Gollehon et al., 1987 and Grood et al., 1988).

Medial Structures

For this paper, the medial structures of the knee have been divided into superficial MCL and the deep medial capsule. The deep medial capsule has been further divided into anterior, middle, and posterior thirds. The deep MCL is considered mid-medial capsule, and the posterior medial corner is considered posterior-medial capsule.

Primary Function

Medial structures, particularly superficial MCL, act as the primary restraint to limit valgus angulation (Grood et al., 1981; Seering et al., 1980; and Markolf et al., 1976). Since the collateral ligaments are better positioned to control rotation than the cruciate ligaments, the medial structures act as the primary restraints to internal tibial rotation. Sectioning the superficial MCL and the mid-medial capsule (deep MCL) results in increased tibial rotation with the knee in extension and flexion (Shoemaker & Markolf, 1986; Seering et al., 1980; and Markolf et al., 1976). The posterior medial capsule plays a greater role as the knee approaches extension. The mid-medial capsule acts to limit internal tibial rotation (Markolf et al., 1976).

Secondary Function

Medial structures act as major secondary restraints to anterior tibial displacement. In two separate studies (Shoemaker & Markolf, 1985 and Markolf et al., 1976), combined sectioning of superficial MCL and mid-medial capsule lead to small or insignificant changes in anterior limits. When MCL sectioning was performed after the ACL (primary restraint) had been divided, the resulting increases in anterior displacement far exceeded those changes seen after individual ligament sectioning.

Lateral Structures

The lateral structures restrain varus angulation and external tibial rotation. A major limitation of in vitro experiments examining the role of lateral structures is the inability to assess the ITB and the popliteus properly. Both structures function as dynamic as well as static stabilizer. To date, the relative contributions of each structure have yet to be adequately determined. In addition, the lateral structures function as a complex, except the LCL, no one structure is responsible for preventing varus angulation (Shoemaker & Markolf, 1985).

Primary Function

Several studies (Grood et al., 1988; Gollehon et al., 1987; Grood et al., 1981; and Markolf et al., 1976) indicate the LCL acts as the primary restraint to limit lateral joint space opening. However, according to the above authors discrepancies in varus limits and also the angle which produces the greatest varus limits, due to LCL sectioning, exists.

Two major conclusions can be drawn from these studies:

(1) Although the LCL acts as a primary restraint to limit varus angulation, deep posterior lateral structures (DPLS) provide considerable restraint as secondary stabilizers.

(2) Increases in varus limit following LCL disruption may be small and difficult to detect clinically, whereas combined injury to LCL and DPLS will result in large changes (Shoemaker & Markolf, 1985).

Lateral structures also act as primary restraints to limit external rotation though relative contributions of individual structures are not as apparent as those for varus limits (Grood et al., 1988; Gollehon et al., 1987; Markolf et al., 1976). Gollehon et al. (1987), noted increases in external rotation at all flexion angles after combined sectioning of the LCL and DPLS. When the LCL was left intact and the DPLS were cut, the only increase in external rotation limit observed was at 90° of flexion. The authors attributed the latter finding to contributions of popliteus tendon but did not section this structure separately to confirm this. Common to all these studies was the finding that changes in external rotation limits were small after individual structures were cut, yet the limit increases following combined sectioning of LCL and deep lateral structures exceeded the sum of component changes. One interpretation of these findings in that no individual structure acts as the primary restraint to external rotation but that, instead, the posterior lateral corner (LCL, arcuate ligament, posterior lateral capsule, and popliteus tendon) function in concert as a complex to limit external rotation.

Secondary Function

Lateral structures act as secondary restraint to limit anterior and posterior motion. Gollehon (1987), illustrated that isolated sectioning of either LCL or DPLS did not change limits to posterior displacement, yet in combination, small (3 mm) but significant increases resulted. In addition, those changes in posterior limits at 0° and 30° of flexion following combined LCL and DPLS sectionings were comparable to increases seen after the PCL was cut. Grood et al., (1988), reported similar findings.

In summary, the LCL and DPLS, when considered individually, act as minor secondary restraints to posterior displacement at full extension.

However, in combination, these structures serve as a major secondary restraint from full extension to 30° of flexion. Although data examining effects of lateral structure sectioning on anterior limits is lacking, lateral structures alone or in combination act as minor secondary restraints to anterior displacement (Grood et al., 1988; Gollehon et al., 1987; Grood et al., 1981; and Markolf et al., 1976).

Muscle Structure

There are seven muscles that flex the knee. The knee flexors are the semimembranosus, semitendinosus, biceps femoris, sartorius, popliteus, gracilis, and the gastrocnemius. All the knee flexors, except the short head of the biceps femoris and the popliteus muscle, are two-joint muscles. Four of the flexors, the popliteus, gracilis, semimembranosus, and the semitendinosus, are considered to medially rotate the tibia on the fixed femur; only the biceps femoris is considered to laterally rotate the tibia on the fixed femur.

Knee Flexors

Of the seven muscles that flex the knee joint three, semitendinosus, semimembranosus, and the biceps femoris, are collectively known as the hamstrings. These muscles all, except the short head of the biceps femoris, originate from the ischial tuberosity. The short head of the biceps femoris takes origin from the lateral lip of linear aspira on the posterior femur. The semimembranosus and the semitendinosus insert on the posteromedial and the anteromedial aspect of the tibia, respectively, and therefore belong to the medial compartment. Both heads of the biceps femoris belong to the lateral compartment since they insert on the lateral condyle of the tibia and the head of the fibula. The sartorius muscle arises anteriorly from the anterior superior spine of the ilium and crosses the femur to insert into the anteromedial surface of the tibial shaft posterior to the tibial tuberosity. At its insertion, the sartorius joins with the tendons of the semitendinosus and the gracilis to form the pes anserinus. The gracilis arises from the inferior half of the symphysis pubis and the pubic arch and inserts on the medial tibia by way of the pes anserinus. All of the above flexors, except the biceps femoris, are medial compartment muscles, and all of the above, except for the short head of the biceps femoris, cross the hip joint and the knee joint.

The only other one-joint muscle, besides the short head of the biceps femoris which flexes the knee, is the relatively small popliteus. This muscle originates on the posterior aspect of the lateral femoral condyle and attaches on the medial aspect of the tibia. The fibers of the muscle run medially across the posterior aspect of the knee, and some of these fibers may attach to the lateral meniscus. The popliteus muscle is a medial rotator of the tibia, when the tibia is fixed in a closed kinematic chain. The popliteus is considered to play two important roles:

(1) It initiates unlocking of the knee, because of its rotatory action on the bony levers.

(2) Because of its attachment to the lateral meniscus, it is thought to pull this cartilage posteriorly during flexion of the knee and lateral rotation of the tibia so it will not be crushed.

The gastrocnemius muscle arises from the posterior aspect of the medial and lateral condyles of the femur by two heads. It inserts into the calcaneus by way of the Achilles tendon. With the exception of the plantaris muscle, the gastrocnemius is the only muscle at the knee that crosses the ankle and the knee.

Knee Extensor

The four extensors of the knee are known collectively as the quadriceps femoris. The only muscle of the quadriceps that crosses two joints is the rectus femoris, which has two head of origin; the straight head from the inferior spine of the ilium and the reflected head from the groove

Table 1.0. Summary of Structures Providing Knee Stability

	Ligaments	**Musculotendinous Structures**
Medial Stability	Medial Collateral Meniscofemoral Coronary Posterior Cruciate	Pes Anserinus Semimembranosus
Lateral Stability	Lateral Collateral Meniscofemoral Coronary Anterior Cruciate Posterior Cruciate	Popliteus Biceps Femoris Iliotibial Band
Anterior Stability	Anterior Cruciate Medial Collateral Lateral Collateral	Extensor Retinaculum Patella
Posterior Stability	Posterior Cruciate Oblique Popliteal Arcuate	Biceps Femoris Gastrocnemius Semimembranosus Popliteus
Anteromedial & Anterolateral Rotatory Stability	Medial Collateral Oblique Popliteal Anterior Cruciate	

From Norkin & Levangie, 1983.

along the acetabulum. The vastus intermedius, vastus lateralis, and the vastus medialis originate on the femur and insert by way of a common tendon, the quadriceps tendon, into the base of the patella. The vastus medialis and the vastus lateralis also insert directly into the medial and lateral aspect of the patella and by way of the ligamentum patellae into the tibial tuberosity. The vastus medialis position of the quadriceps consists of two parts that are separated by a fascial plane; the vastus medialis longus and the vastus medialis oblique. The fibers of the longus are small and are directed almost horizontally.

MECHANISM OF ACUTE KNEE INJURIES

The major ligaments of the knee can be torn in isolation or in combination. Depending on the application of forces, injury can occur from a direct straight-line or single-plane force or from a rotary force.

Single-Plane Injuries

Usually, single-plane knee injuries occur when the athlete's foot is fixed. The traumatic force may be directed, such as being hit in the knee by another player, or indirect, through sudden valgus, varus, anterior, or posterior movement. In sustaining direct or indirect ligamentous injury, the knee may be in a position of extension or flexion (Arnheim, 1985).

MCL Injuries

The MCL and capsular ligaments can be torn by a direct blow to the lateral aspect of the athlete whose foot is firmly planted. The MCL can also be injured as result of indirect valgus force with the tibia in external rotation. This same mechanism can occur to skiers who catch the inside of their skis in the snow. The indirect valgus force with the tibia in external rotation could also tear the ACL or the PCL along with the medial meniscus (Arnheim, 1985).

LCL Injuries

Sprain of the LCL is much less prevalent than sprain of the MCL. The force required to tear this ligament is one of varus, often with the tibia internally rotated. Because of the usually inaccessible medial aspect, a direct blow is rare. In skiing, the LCL can be injured when the skier fails to hold a snowplow and the tips cross, throwing the body weight to the outside edge of the ski. If the force or blow is severe enough, both cruciate ligaments, the attachments of the IT Band, and the biceps muscle may be torn. This same mechanism could also disrupt the lateral and even the medial meniscus (Arnheim, 1985).

ACL Injuries

Although the ACL is most vulnerable to injury when the tibia is externally rotated and the knee is in a valgus position, single-plane forces can also produce injury. Hyperextension, a sudden deceleration with the foot fixed as seen in basket or a force directed anteriorly on the femur or a force directed posteriorly on the tibia with the foot planted can tear this ligament. Also, the same mechanism that sprains the MCL, if severe enough, can tear the ACL (Arnheim, 1985).

PCL Injuries

The PCL is vulnerable to injury after the ACL has been torn and the knee forced into hyperextension. The PCL is most vulnerable when the knee is flexed to 90°. Falling with full weight on the anterior aspect of the bent knee or receiving a hard blow to the front of the bent knee can tear the PCL (Arnheim, 1985).

Rotary Ligament Injuries

A major mechanism of ligamentous injury is that of rotation with the foot fixed. During internal rotation and external rotation of the tibia, the ACL becomes taut. An athlete who is running fast and suddenly decelerates and makes a sharp cutting motion could produce an isolated tear of the ACL. The same mechanism could be true of the skier when the ski catches in the snow and the body twists medially or laterally. The two most common rotary injuries leading to knee instability are the anteromedial and anterolateral types. Anteromedial rotary motion can tear the MCL or both the MCL and the ACL. In anterolateral instability, the ACL is also involved, along with a tear or laxity of the posterolateral capsule (Arnheim, 1985).

DIAGNOSIS OF A LIGAMENT INJURY

Medial Collateral Ligament

Valgus Stress Test

The valgus stress test evaluates the MCL. The patient lies supine with the knees supported in 20° to 30° of flexion and neutral axial rotation. The examination is performed by stabilizing the femur and controlling the joint flexion angle. With one hand on the distal tibia, the physician exerts an axial load to place the joint surface in contact; this is the test starting position. The leg is then abducted while constraining axial rotation. The medial joint space opening is estimated and the stiffness of the motion limit is evaluated. The findings are compared to the patients contralateral normal knee. In a first-degree injury, there is pain and tenderness at the site of the ligament injury; the end point is firm and the joint space opening is within 2 mm of the normal knee. In a second-degree injury, the end point is relatively firm and the joint space opening is increased 3-5 mm compared with that of the normal knee. In a third-degree injury, the end point is soft and the joint space opens more than 5 mm greater than that of the normal knee (Ellison, 1977).

Lateral Collateral Ligament

Varus Stress Test

The varus stress test evaluates the LCL. The patient lies supine with the knees supported in 20° to 30° of flexion and neutral axial rotation. With one hand the examiner stabilizes the femur and palpates the lateral joint line. With the other hand on the total tibia the examiner first exerts an axial load to place the joint surfaces in contact; this is the test starting position. The leg is then adducted while constraining axial rotation. The lateral joint space opening, as well as stiffness of the motion limit, is estimated. The grading system of injury is the same as for MCL injuries (Daniel, 1988).

Posterior Cruciate Ligament

Posterior Drawer Test

Both the ACL and the PCL limit total anterior-posterior displacement of the tibia. The examination of the cruciate ligaments begins with the evaluation of the PCL. A PCL disruption results in greater posterior displacement of the tibia from the anatomic resting position; an ACL disruption allows greater anterior displacement of the tibia from the anatomic resting position. To determine which structure(s) are disrupted, the clinician must be able to determine the neutral position. The neutral position is the resting position of the tibia supported by the intact PCL. The neutral position can be determined when the patient is lying supine with the knee at 90° of flexion (Burks, 1990). The resting position of the injured knee should be compared with the contralateral normal knee. If the PCL is disrupted, the tibia will sag posteriorly. The sag may be seen by looking at the knee profile, palpated by feeling the femoral condyle-tibia step-off, and confirmed by the ***quadriceps active test*** wherein contraction of the quadriceps pulls the tibia anteriorly. To measure the posterior displacement the tibia is first placed in the reduced position. If the tibia does not sag posteriorly and can not be displaced posteriorly more than the contralateral knee with an intact PCL, the PCL is intact and the resting position is the anterior-posterior (A-P) neutral position (Burks, 1990).

Anterior Cruciate Ligament

Anterior Drawer Test

The anterior drawer test is the most commonly used test to evaluate anterior stability of the knee joint. This test is best performed with the patient in a comfortable, relaxed, supine position. The hip is flexed approximately 45°, and the knee about 80° to 90°, with the foot resting flat on the table. The foot should be in a neutral position, facing straight ahead. While stabilizing the foot the examiner encircles the proximal end of the tibia and fibula, immediately below the knee joint, with both hands. The fingers of the examiner are positioned in the popliteal space of the affected limb, with the thumbs on the medial and lateral joint lines. On application of anterior tension, the tibia sliding forward from under the femur is considered a positive anterior drawer sign. If a positive anterior drawer sign occurs, the test should be repeated with the athlete's leg rotated internally 20° and externally 15°. Sliding of the tibia anteriorly when the leg is externally rotated is an indication that the posteromedial aspect of the joint capsule, the ACL, or possibly the MCL could be torn. Anterior displacement of the tibia, when the leg is internally rotated, indicates that the ACL and the posterolateral capsule may be torn (Arnheim, 1985 and Booher and Thibodeau, 1985).

The anterior drawer test may also be performed with the patient sitting and the knee hanging over the edge of the table. The foot is then stabilized between the knees of the examiner and raised slightly to reduce the effect of gravity. If the foot is not supported, tension is applied to the remaining structures and the test is much more difficult to evaluate. The rest of the procedure are as stated above (Arnhein, 1985 and Booher and Thibodeau, 1985).

Lachman Test

The Lachman test (a more recent test) is an excellent test to evaluate the anterior tibial displacement limit (Jonsson et al., 1982 & Torg et al., 1976). Before performing this test, the integrity of the PCL should be established with the quadriceps active test at 90° of flexion (See page 31).

To perform the Lachman test, the knee is placed in a position of 20° to 30° of flexion. The patient lies supine with a support under the thigh. The clinician stabilizes the femur against the thigh support and applies an anterior displacement force to the calf without enhancing or restraining axial rotation. The examiner senses the tibial displacement limit (end point). If both are normal, the Lachman test is negative. If either are pathologic, the Lachman test is positive. The displacement limit or "end point" may be graded and the side-to-side displacement difference estimated. Endpoint is graded as firm (normal), marginal, or soft. Displacement is estimated in millimeters. An estimated right-left difference of 3 mm or greater is classified as pathologic. If the PCL is intact, abnormal A-P displacement on the Lachman test indicates an ACL disruption. Examiners are better able to detect end-point differences than displacement differences. An experienced examiner can correctly diagnose an ACL disruption, even when there is only a 4 mm right-left displacement difference, because of the alteration in end-point stiffness (Burks, 1990).

Pivot Shift

Pivot shift tests are complex tests of limits of the knee motion. The pivot shift tests have been described by numerous authors as pivot shift tests (Galway & MacIntosh, 1980 and Fetto & Marshall, 1979), Loose test (Loose et al., 1978), side lying test (Slocum et al., 1976) and flexion-rotation drawer test (Noyes et al., 1980). The test produces anterior subluxation and internal axial rotation in early flexion as a result of an ACL disruption. The posterior pull of the iliotibial tract reduces the tibia at 20° to 40° of flexion. The tests are performed by lifting the tibia and allowing the femur to fall posteriorly. As the knee is flexed and the iliotibial tract is tightened and moved from a position anterior to the axis of knee flexion to a position posterior to the axis of flexion, the anteriorly displaced and internally rotated tibia reduces. It is the relocation event that the clinician usually grades. The pivot shift is graded as 0 (normal), 1+ (slight slip), 2+ (moderate slip), or 3+ (momentary locking). The effect of tibial rotation on pivot shift (Clancy and Ray, 1987), and the effect of hip adduction (Bach et al., 1988), have been reported. Internal rotation of the pivot and adduction of the hip both tighten the iliotibial tract; therefore the tibia reduces sooner, and the pivot shift grade is reduced. Disruption of the medial collateral ligament (MCL) allows the limb to go into valgus alignment and relax the ITB. This reduction in the ITB tone will result in a decrease in the pivot shift reduction event. Likewise, surgery altering the ITB may alter the pivot shift reduction event, even when joint subluxation is not altered. This is important to consider when evaluating ACL reconstruction patients who have had ITB procedure. The pivot shift is consistently positive in the relaxed patient with a chronic ACL disruption and in the acutely injured anesthetized patient with an ACL disruption (Daniel et al., 1988).

ACL MANAGEMENT

The treatment of ACL tears is either nonsurgical or surgical. The decision of whether to treat with rehabilitation and functional bracing or to reconstruct the ligament, depends on the amount of anterior laxity displayed clinically, amount of instability during dynamic activity, and the physical demands placed across the knee joint. Treatment recommendations by the physician will usually be made based on the assessed degree of instability in relation to the athlete's desired level of function and willingness to change from it. If a "giving way" and functional instability are significantly altering the athlete's desired level of performance, there are three treatment options:

1) The athlete can reduce the demands placed on the knee by appropriate changes in the lifestyle.

2) The athlete can aggressively rehabilitate the knee (conservative treatment) and use a functional knee brace, with the hopes of returning to previous levels of activity.

3) The athlete can have ACL reconstruction in an attempt to restore normal joint mechanics and eliminate joint instability (Halling et al., 1993).

What ever option the athlete chooses, s/he must be willing to put time, effort, and energy into the laborious rehabilitation process.

Basic Physiology of Ligament Healing

There is an initial state of hematoma formation and infiltration of inflammatory cells which occupies approximately the first week of healing. During the second week, fibroblastic proliferation about the site of the ligamentous tear occurs, and from the third to sixth week after injury, the healing process is characterized by increasing collagenization and organization. Subsequently, the newly formed collagen matures according to the stresses placed upon it and organizes from a random pattern to a more organized ligamentous-like histology. The state of this organization takes at between 6 to 16 weeks. However, the outcome of ligament repair is directly effected by:

(1) The proximity and apposition of torn ligaments ends.

(2) The local available blood supply.

(3) Local environment factors (Andrish, 1984).

Conservative Rehabilitation

With an acute injury, the patient is initially placed in a knee immobilizer and given crutches for 1 to 2 weeks; this allows hemarthrosis to resolve (Nisonson and Goldberg, 1991).

After immobilization, a physical therapy regimen can begin with a range-of-motion program. The key to a successful nonsurgical program is an extensive lower-extremity strengthening program (examples being hip extensions and leg press) emphasizing the hamstrings--especially the biceps femoris. Although, hamstring strength is vital to the outcome of the ACL-deficient rehabilitation, strengthening of other muscles should not be ignored. Quadriceps strength in a good functional relation to the hamstrings is critical, along with agility and proprioceptive training (Halling et al., 1993 and Nisonson and Goldberg, 1991). Closed kinetic chain exercises such as the eccentric drop squat protocol and endurance training (stationary bicycle riding) is also beneficial during this period. Also during this period of time it is best to avoid competitive sports requiring pivoting (Nisonson and Goldberg, 1991).

A functional brace is an important adjunct for athletics and should be used especially for sports involving sudden pivoting or changing directions. If however, instability persists inspite of a brace further examination is required and surgical reconstruction must be considered (Nisonson and Goldberg, 1991).

ACL-Reconstructed Rehabilitation

The return of an athlete to pre-injury levels of activity following a reconstruction of the ACL depends on the surgical re-establishment of proper joint mechanics and the successful implementation of an aggressive rehabilitation program. The primary objective of rehabilitation is to safely promote healing of the graft while restoring normal function to the knee and affected limb.

Whatever surgical procedure used, rehabilitation must take into consideration the histology, biomechanical adaptations, and vascular adaptations of the graft tissue. Other principles that must be incorporated in the rehabilitation program are:

1) Mobilization versus immobilization

2) Passive knee extension exercises versus resistive knee extension exercises

3) Extensive use of resistive knee flexion exercises (hamstring exercises) early in the program.

4) Use of closed kinetic chain exercises (squats, horizontal leg press, cycling, step-ups, stair climbing, and resisted gait training) for early quadriceps strengthening.

5) Early use of continuos passive motion (CPM). The effects of CPM are to improve joint nutrition, promote healing, decrease pain, decrease joint effusion and maintain or increase range of motion (ROM).

6) Proprioceptive training of the lower extremity following surgery is necessary to minimize or prevent neuromuscular inhibition, and to attain normal motor control, strength, and coordination. Proprioceptive neuromuscular facilitation (PNF) and balance therapy should be instituted early.

7) Use of a functional knee brace--during rehabilitation and many times for the rest of the athlete's competitive athletic career (Halling et al., 1993).

By following the above principles of rehabilitation and strength training, the ligament usually requires 12 to 18 months before complete healing has occurred. However, with extensive rehabilitation, strength training, as well as utilizing a functional knee brace, the injured athlete can expect to return to competitive sports in 6 to 8 months post injury.

KNEE BRACING LITERATURE REVIEW

Until 1979, the principal use of knee braces was for rehabilitation. In 1973, Nicholas, provided information on the first brace to be utilized in a sporting domain. In 1985, the American Academy of Orthopedic Surgeons (AAOS) sponsored a seminar to discuss the effectiveness of knee braces. From the results of this seminar, the AAOS, (in 1987) classified all braces into 3 distinct types of braces--rehabilitation, prophylactic and functional knee braces. Rehabilitation braces are designed to allow protected motion of injured knees treated operatively or nonoperatively. Prophylactic braces are "off-the-shelf" knee braces designed to prevent or reduce the severity of knee injuries. Functional knee braces are custom made knee braces that are designed to provide stability for unstable knees (Wirth and DeLee, 1990). The concentration of this paper will be on the latter two braces, prophylactic and functional.

Research on prophylactic braces has focused on two domains. These two domains are epidemiological and cadaver and surrogate studies. Studies on functional knee bracing have focused on five domains. These include subjective and clinical functional assessments, energy expenditure, performance testing, kinematics, and force plate analysis.

Prophylactic Knee Braces

Prophylactic knee braces are of two design types. The first consists of a lateral bar design with a single axis, dual axis, or polycentric hinges fitted with a hyperextension stop. Examples include the McDavid Knee Guard (McDavid Knee Guard), Anderson Knee Stabler (Omni Scientific), and the Protective Knee Guard (Don Joy Orthopaedic) (Wirth and DeLee, 1990).

The second type consists of plastic cuffs with polycentric hinges; this is often custom fitted. Examples include the Loose Knee Defender (Don Joy

Orthopaedic), Am-Pro Knee Guard (American Prosthetics), and the Iowa Knee Orthosis (Am-Pro Knee Guard) (Wirth and DeLee, 1990).

Epidemiological Studies

Ten studies are reviewed. Four reports (Jackson et al., 1991; Hansen et al., 1985; Schriner, 1985; and Anderson, et al., 1979) support prophylactic knee bracing. Four reports (Rovere et al., 1987; Hewson et al., 1986; Taft et al., 1985, and Albright et al., 1994) show no significant change in knee injuries with bracing. Two studies (Grace et al., 1988 and Teitz et al., 1987) showed an increased incidence of knee injuries with knee bracing. The most recent study supports prophylactic knee bracing but only for players playing specific positions (Albright et al., 1994). See Table 2.0. A critical analysis of each report offers much more insight.

During the 1970's, knee braces were used primarily for rehabilitation purposes. Devices like the Lenox Hill brace were viewed as too bulky and restrictive to be used during competition. Anderson et al. (1979), described their experience with the 'Anderson Knee Stabler', a "double-hinged", single sided brace to support the collateral structures of the knee (Montgomery and Koziris, 1989). Their subjects included 9 professional football players, each having a previous MCL injury. Severity of injury was described in terms of time missed from practices and games. The injuries were such that 4 players had not missed a single game, and one player had missed 7 games. After wearing the Anderson brace, the 9 players played a total of 29 games over 2 seasons without re-injury to the knee. With this limited data, the authors concluded: "Its use in all sports, especially contact sports, is highly recommended to protect injured knees. Its use as a preventative device by athletes in vulnerable positions is highly applicable." (Montgomery & Koziris, 1989).

Table 2.0. Summary of Ten Epidemiological Studies Using Prophylactic Knee Braces

Reference	Subjects	Knee Brace Type	Result(s)
Anderson (1979)	9 professional players with knee injuries	Anderson Knee Stabler	Subjects played 29 games over 2 seasons without reinjury to the knee
Hansen (1985)	University of South California. 329 non-braced players and 148 braced players	Anderson Knee Stabler	Injury rate for nonbraced players was 11% versus 5% for braced players.
Schriner (1985)	1246 highschool players from 25 school in Michigan	Don Joy, Cutter Anderson, Omni, McDavid Knee Stablizer	45 injuries from lateral blows in non-braced group versus 0 injuries from lateral blows in braced group.
Taft (1985)	University of North Carolina--(non-braced 1980-1982; braced 1983-1985)	Anderson Knee Stabler & McDavid Knee Stablizer	No statistical difference in MCL, ACL or meniscal injuries between the non-braced and the braced periods.
Hewson (1986)	University of Arizona. Exposures to injury non-braced (1977-1981 [28,191]); braced (1981-1985 [29,293])	Anderson Knee Stabler	No statistical difference in MCL injuries between the non-braced and braced periods.
Rovere (1987)	Wake forest University--1981-1984	Anderson Knee Stabler	6.1% knee injuries during non-braced period and 7.5% knee injuries during braced period.

Table 2.0 continued.

Teitz (1987)	NCAA Div. 1. 6307 players in 1984 (71 schools); 5445 players in 1985 (61 schools).	Many	1984, 6.0% injury rate for non-braced versus 11% for braced players. 1985, 6.4% injury rate for non-braced versus 9.4% for braced players. Significantly more MCL injuries among braced versus non-braced players.
Grace (1988)	Albuquerque & Santa Fe high school players: 250 non-braced players versus 83 double hinged players.	Primarily McDavid Knee Stablizer and Stromgern	Knee injury rates: non-braced = 4% single-hinge = 15% double-hinged = 6%
Jackson (1991)	1 professional team in the Canadian Football League (1977-1988)	McDavid Knee Stabilizer, Depuy, and Anderson Knee Stabler	The number and the severity of knee injuries decreased--MCL injuries decreased by 33%.
Albright (1994)	NCAA Division I college football players.	Unspecified	Found noticeable differences in the rates of injury (decrease) for braced and unbraced knees in all position except backs and kickers. Other factors play a role as well.

Modified from Requa & Garrick, 1990 and Montgomery & Koziris, 1989.

Since 1979, over 30 manufacturers have introduced braces that have claimed to prevent knee injuries. Before 1985, accounts of the successes or failures of these devices were largely anecdotal (Garrick & Requa, 1987).

In 1985, 5 studies on prophylactic knee braces were presented at national refereed sports medicine meetings or published in peer-reviewed journals. These epidemiological and statistical studies created controversy (Montgomery & Koziris, 1989). Hansen et al. (1985), reviewed the medical records of football players at the University of Southern California from 1980 to 1984. During that period, 329 players were non-braced and 148 were braced. No definition of a "knee injury" was given. Only injuries requiring surgical intervention were included in the comparison between the 2 groups. According to these authors, bracing was beneficial in preventing collateral ligament and meniscus injuries to the knee. There were 35 knee injuries to non-braced players (11% injury rate) and 7 injuries to braced players (5% injury rate). The authors recommended that all linebackers and interior linemen wear the Anderson Knee Stabler brace during practices and games. This study did not define the criteria for brace usage, nor the rate of exposure to injury for each group (Garrick & Requa, 1987). Exposure to injury was simply "being on the team" at some period (spring or fall) during the 5 years of the study (Montgomery & Koziris, 1989).

Schriner (1985), surveyed the injury rate of 1246 football players from high schools in Michigan during the 1984 season. Four types of prophylactic knee braces were worn by 197 players (16% of the sample) from 12 schools. The criteria for bracing and the rate of exposure to injury for each group were not reported. The diagnosis of injury was made by physicians, but the survey was completed by the coaches. Injuries were classified and analyzed in relation to cause:

(a) From lateral blows

(b) From medial or posterior blows

(c) From hyperextension.

The braced and unbraced groups had a similar rate of injury from medial or posterior blows and from hyperextension. In the braced group, there were no injuries attributed to lateral blows. It was concluded that preventative knee braces reduced knee injuries to the MCL from lateral blows. Schriner (1985), stated that knee braces reduced knee injuries to the MCL from lateral blows. Furthermore, Schriner (1985), suggested that knee braces become mandatory for all football players in all positions (Montgomery & Koziris, 1989).

Taft et al. (1985), reported the experience of the University of North Carolina Football team from 1980 to 1982 (unbraced) and from 1983 to 1985 when all players wore knee braces. The definition of an injury was a modification or absence from practice for at least 1 week. Injuries were evaluated by an orthopaedic surgeon and classified as operative or non-operative. There was no significant difference between the braced and unbraced period in the number of MCL, ACL or meniscal injuries. The authors state, "The indications for surgery had not changed from that used before the advent of bracing." MCL injuries requiring surgery dropped from 5.7 per year in the unbraced condition to 1.4 per year in the braced condition (Montgomery & Koziris, 1989).

Hewson et al. (1986), compared the experience of the University of Arizona football team from 1977 to 1981 (unbraced) with 1981 to 1985 (braced). The Anderson Knee Stabler brace was made compulsory for players at greatest risk of knee injury--linemen, linebackers, and tightends.

Orthopaedic surgeons classified all MCL injuries. There were 28,191 exposures during the 5 years of brace use and 29,293 exposures during the control period. There were no significant differences in the numbers, types or severity of knee injuries between 'at-risk' players with or without knee braces. There were a total of 54 knee injuries in the unbraced period and 48 in the braced period. There was no statistical difference (33 versus 41) in MCL injuries between the braced and unbraced periods. Although there was a reduction in the number of missed practice days during the bracing period, this decrease was attributed to changes in treatment techniques. During this period of brace usage, 7 rule changes, directed at reducing knee injuries, were introduced. The authors concluded that neither the rule changes, nor the braces, nor their combination had any effect on knee injuries (Montgomery & Koziris, 1989).

Rovere et al. (1987), reported the experience of Wake Forest University Football team during the 1983 and 1984 seasons. All players were required to wear the Anderson Knee Stabler brace during this time period. Statistics on injuries were compared to the 1981 and the 1982 seasons in which braces were not worn by any players. Injuries were evaluated and graded (I, II, or III) by an orthopaedic surgeon. MCL and ACL trauma were assessed using valgus and varus stress testing at 30°. Anterior instability was evaluated using both the Lachman and the anterior drawer tests. The rate of injury was expressed as the incidence of injury per 100 players averaged over the 2 seasons. During the non-braced period, there were 24 knee injuries (6.1 per 100 players) compared to 29 knee injuries (7.5 per 100 players) during the braced period. The incidence rate for Grade I strains of the MCL were 4.0 per 100 players during the non-brace period and 4.8 per 100 during the brace period. In addition, there were 5 knee operations during the non-braced period and 9 operations during the braced period. Brace usage was also associated with increased episodes of cramping in the triceps surae muscle group. This study found the Anderson Knee Stabler to be ineffective as a prophylactic device.

Most studies, concerned with the effectiveness of knee bracing, have been conducted with a limited sample size at a single institution. The retrospective investigation by Teitz et al. (1987), used many institutions and a large sample size. The study questioned the benefits of prophylactic of knee bracing. It concluded that so-called preventative braces are not preventative and may in fact be harmful.

This study included data on 6037 football players from 71 schools in 1984 and 5445 players from 61 schools in 1985. In 1984, 36% of the players wore knee braces and in 1985, 44% of the players wore braces. In both years, the players who wore the prophylactic braces had a significantly increased rate of injury to the knee compared with the rate of injury in players who did not wear braces (for 1984, 11.0% compared with 6.0%; for 1985, 9.4% compared with 6.4%). In addition, the injuries that did occur were as severe in players who wore braces as those who did not. Severity of the injury was assessed using 3 criteria: loss of playing time, grade of the MCL injuries and injuries that required surgery. Despite the purported value of the brace in prevention of injure to the MCL, there were significantly more injuries to the MCL among players who wore braces (for 1984, 7.6% compared with 3.5%; for 1985, 5.4% compared with 3.6%).

In their study, Teitz et al. (1987), attempted to control for many of the known biases. The data for 1985, did not include players who had previous knee injuries thus removing the possible bias that previously injured players might be more likely to wear braces or be re-injured, or both. In addition the level of player, first, second, third and fourth string, were examined separately. At each level, braced players had a higher incidence of injury than unbraced players. Also, the data from the schools in which the players did not wear braces one year, but then wore braces during the following season were analyzed separately.

The positive results experienced by some schools after switching to knee bracing may simply have been due to the effect of "regression to the mean" (Montgomery & Koziris, 1989). In Teitz's study, 6 of the 9 schools started to utilize braces after seasons in which the rates of injury for their teams were much higher than the average rate of knee injury. Improvement in the rate of injury in the following season had possibly nothing to do with the use of braces. Rather, it could have been due to chance fluctuation or, merely, reversion towards the mean rate of injury. In spite of this apparent possibility of bias for bracing, the average rates of injury during the years when players wore braces were slightly higher than in those years when players did not wear braces (Teitz et al., 1987). The authors speculated that the increased rate of injury in players who wore braces may have been the result of decreased agility caused by the braces, carelessness of players who believed that they were protected, or preloading of the MCL in players who had genu varum.

Critics of the retrospective study by Teitz et al. (1987), point out that the exposure to injury for each group was not calculated. Also, variables such as the way the braces were attached, the definition of injury, and the treatment of knee injuries by different physicians are difficult to control and affect the determination of the severity of injury (Montgomery & Koziris, 1989; Garrick & Requa, 1987).

Grace et al. (1988), compared 247 players who wore single-hinged braces, 83 who wore double-hinged braces, and 250 players who did not wear braces. The groups were matched according to size and playing position. Injuries were graded from mild to severe. There were 37 injuries (15% injury rate) in the group wearing single-hinged braces and 11 injuries (4% injury rate) in the control group. The group wearing the single-hinged braces had an injury rate that was 3.7 times higher than the unbraced group. There was no significant increase in the incidence of knee injuries in the group wearing the double-hinged braces. There was an increased number of foot and ankle injuries in players wearing both types of braces. These results were confirmed during the second year of the study; 23% of the braced group suffered foot or ankle injuries compared to only 8% for the non-braced group.

Jackson et al. (1991), conducted a study lasting 12 years. All definitions (mechanism of injury, grading the severity, type of exposure--game/practice, the length of exposure, playing surface, player experience, and player position) were clearly accounted for. Furthermore, the validity of the study was increased as only two individuals (Team Physician and Team Trainer) diagnosed, treated, and recorded all injuries.

During the 5 year period from 1984-1988, 25 players wore prophylactic knee braces for all games. These 25 players averaged 13 games per season (70% game participation) and accumulated 524 game exposures. The unbraced control group, matched for position, experience, and the number of years played, also averaged 13 games per season and accumulated 544 exposures (Jackson et al., 1991).

The braced group's risk of injury was reduced by 33% (16 knee injuries compared to 25 knee injuries). Also, the severity of injury was reduced in the braced group, with no major injuries and 87.5% being minor injuries (Table 3.0).

The risk of MCL injuries (grade I minus, I, II, and III) is higher (Table 4.0) in the nonbraced control group. Furthermore, the braced group had far less severe MCL injuries than the unbraced group, with no grade II or III lesions.

Albright et al. (1994), conducted a two-part, 3 year study on the

Table 3.0. The Effects of Wearing Prophylactic Knee Braces on the Severity of Knee Injuries (5-Year Period)

	Non-Braced	Braced
Total Number of Injuries	25	16
Minor (<7 days missed)	13 (52%)	14 (87.5%)
Moderate (7-21 days missed)	6 (24%)	2 (12.5%)
Major (>21 days missed)	6 (24%)	0 (0.0%)

From Jackson et al., 1991.

effectiveness of prophylactic knee braces. In part one of their study, these researchers determined that testing and analysis would provide accurate results only if influential factors were identified. These influential factors consisted of player's session (game or practice), position, brace wear patterns, and string, where they share similar job descriptions. Albright et al. (1994), found that during games, injuries occurred at a rate of 6 to 12 times greater than the rate associated with practices, and that the patterns of injury are different than in practices. Furthermore, these researchers felt that individual positions should be divided into 3 general position groups, according to similarity of brace wear preference patterns in practices and games. Lastly, players, (starters and substitutes) should be analyzed separately from nonplayers. Nonplayers should be analyzed only during contact practice sessions, while players can be studied during contact practices and games.

Table 4.0. The Effects of Wearing a Prophylactic Knee Brace on the MCL Injury Rates and Severity (5-Year Period)

	Non-Braced	**Braced**
Total Number of Injuries	25	16
MCL Injuries (% of Total)	16 (65%)	9 (56%)
Grade I "minus"[2]	4 (44%)	8 (89%)
Grade I	5 (31%)	1 (11%)
Grade II/III	4 (25%)	0 (0%)

From Jackson et al., 1991.

In part two of their study, the above researchers found noticeable differences in the rates of injury for the braced and unbraced knees, depending on player or nonplayer status, in almost every position during practices. Furthermore, when these investigators included the above influential factors, they found a consistent but not a statistically significant tendency for the players wearing prophylactic knee braces to experience a lower rate of injury than their unbraced counterparts. For starters and substitutes in the line positions, as well as the line backers and tight ends, there was a consistent trend towards a lower injury rate in both games and

[2] Grade 1 "minus" injuries are minor sprains resulting in one or two games or practices being missed.

practices. However, in the skill positions (backs/kickers) braced players, at least during games, demonstrated a higher injury rate (Albright et al., 1994).

Critique

Except for Jackson's study, the other nine studies have to be questioned regarding their validity and variance. Many studies lacked significant numbers to obtain statistically significant results. Others had high variance as large numbers were used but no controls were set-up in terms of diagnosis, treatment or in recording the injuries. Furthermore, many studies did not establish a denominator to calculate exposure and injury rates. Moreover, most studies did not examine for knee joint laxity and/or for previous knee injury prior to knee bracing. Both of these factors can result in a higher risk of injury or reinjury. Lastly, most studies did not provide any statistical analysis for their results.

Cadaver and Surrogate Studies

Several studies have examined the knee brace issue from a biomechanical perspective using cadavers and surrogate models. Paulos et al. (1986, 1976), applied strain gauges to the bone of cadaver knees at the ligament attachment sites to determine the forces and joint openings necessary to disrupt the valgum--restraining ligaments in braced and unbraced knees. Forces were applied with a hydraulic test apparatus. Joint openings were measured with a single-axis electropotentiometer. Tests performed on the 4 braces showed that the mechanical stiffness varied 3-fold among the braces. The unbraced knee was more rigid than the braces alone. The average stiffness of the unbraced knees at the point of ligament failure was equivalent to 105.8 kN/m compared to 25.1 kN/m for the braces alone. The resting tension in the MCL was increased in 60% of the braced knees. This tension was attributed to preloading of the ligament. Knees with a slight to moderate varus demonstrated as much as a 160% increase in ligament tension when braces were in place.

The report by Paulos et al. (1986), was criticized by McDavid (1986) and Pipes (1986). Pipes claimed that the raw data did not support the concept that lateral bracing preloads the MCL whereas McDavid stated that the data did not support a correlation between knee alignment and preloading. McDavid claimed that preloading was not a result of brace application but, rather, a consequence of using joint openings as the control measurement. Paulos et al. (1986), reply mentioned that the axis of rotation for the knee was relatively fixed, so joint opening measurement was not significantly affected by brace application.

Paulos et al. (1987), evaluated brace-knee composite loading responses and knee ligament injury mechanisms due to valgus loading using 18 human cadavers. Two biomechanical tests, static nondestructive and low-rate destructive testing, were conducted using laterally applied loads to produce medial joint openings. For destructive testing, specimens from young adults were used to ensure failure of the ligament in substance rather than at its bony attachment. Before, applying the braces, individual ligament contributions to valgus-restraining function were established. The effects of lateral bracing (McDavid Knee Guard and Omni Anderson Knee Stabler) were then analyzed using the criteria of valgus force, joint line opening, and ligament tensions.

Mechanical terms were described as follows:

(1) Impact loading occurs when the duration of the external load applied is a fraction of the natural period of frequency for the material being loaded

(2) Impact failure occurs when the rate of loading is greater than the rate of energy absorption and thus one gets deformation of the material.

The 3 principal factors that determine the impact response characteristics of a brace/knee composite are force distribution, energy absorption, and energy transmission (Montgomery & Koziris, 1989).

Surprisingly, disruption of the MCL from valgus forces occurred at higher ligament tensions than disruption of the ACL and PCL in both the braced and unbraced conditions. The average peak failure tensions under valgus load were 1122N, 1406N, and 2346N, respectively, for ACL, PCL, and MCL. The average lateral loads at peak ligament failure were 837N, 977N and 1058N for ACL, PCL, and MCL, respectively. The fractional contributions of each ligament to medial restraint were 11%, 9%, and 80%, respectively, for ACL, PCL, and MCL (Paulos et al., 1987). Although large joint displacements were needed for complete ligament failure, bundle disruption in the 3 ligaments was evident with much smaller joint opening. Paulos et al. (1987), concluded that lateral bracing with the McDavid Knee Guard and Omni Anderson Knee Stabler did not offer any significant protection. The braces permitted from 3cm to 9cm of deformity before permanent deformation, and this occurred at low forces. The lack of rigidity coupled with a small joint line clearance made the braces ineffective in resisting valgus forces. The results differed from the data provided by the 2 brace manufacturers (Montgomery & Koziris, 1989; McDavid, 1986). Paulos et al. (1987), claimed that the tests conducted by the brace manufactures were poorly designed.

Four potentially adverse effects of lateral bracing were noted:

(a) MCL preloading

(b) Center axis shift

(c) Premature joint line contact

(d) Brace slippage[3] .

These effects may be magnified in new models unless braces are designed to accommodate variations in knee alignment and leg size (Paulos et al., 1987).

In their follow-up study, the objectives were to:

(a) Determine the clinical significance of brace-induced MCL preload

(b) Define the functional character of an ideal brace

(c) Design and validate a surrogate knee model for testing brace effectiveness

(d) Determine brace performance under impact loading using

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3 (1) MCL preloading is characterized as an increased static MCL tension associated with brace application. The significance of this finding is unclear because constrained cadavers were tested with-out axial loading to simulate weight bearing. Axial loads increase knee stiffness which could negate the preload effect (France et al., 1987).

(2) Center axis shift is a shift of the axis of valgus rotation from the Center of the knee laterally towards the brace. Premature contact of the Center of the brace with the lateral bony structures of the knee decreases efficiency by reducing the effective lever arm (France et al., 1987).

(3) Joint line contact may concentrate force, normally distributed along the lateral surface of an unbraced leg, directly in the knee joint. This may cause more damage to the knee ligaments than that incurred in the unbraced condition (France et al., 1987). Premature joint line contact results from improper brace fit, material properties, structural properties, and fixation techniques (France et al., 1987 & Paulos et al., 1987).

(4) Brace slippage relates to the fit and fixation of the brace on the knee. It is influenced by knee varus/valgus angulation, paddle contour, hinge design and brace fixation technique (France et al., 1987).

the surrogate knee and 6 commercially available brace types.

The effects of MCL preload from knee braces were studied in 13 football players. A surrogate knee model was developed and validated using information from previous cadaver studies and analyses on the effects of high strain rates on MCL failure. Over 500 impact tests were performed on the surrogate knee in unbraced versus braced conditions. Tests were conducted for 3 impact masses (23, 75 and 127kg), 2 flexion angles (0° and 30°), and free or constrained limb positions. Impact safety factors were calculated for each test condition and brace type. An impact safety factor of 1.50 established. This value corresponded to a load reduction of 30% in the MCL and an overall ligament protection of 50%.

The results were:

(a) Braced induced MCL preload in vivo was negated by joint compressive forces

(b) The 'ideal' brace should increase the lateral force at MCL injury by 80%

(c) At a 1000% strain/second strain rate, MCL failure force was increased by 28%

(d) Only one brace (DonJoy) exceeded the minimum impact safety factor.

For the 6 types of braces, the average impact safety factor ranged from 1.18 to 1.51. The 6 braces performed differently depending upon the mass and speed of impact and the degree of constraint of the limb.

In the discussion of their paper, France et al. (1987), gave their opinion on 3 questions frequently posed by practitioners. First, "Can lateral bracing help prevent injury?" Their response was "maybe." "Under specific test conditions, a properly designed brace can provide a significant protective effect for the MCL. However, a newer brace design must be implemented before a complete answer can be given."

Second, "Can lateral or prophylactic braces be harmful?" The response to this question was "probably not, if the braces are properly constructed, adaptable to changing limb contours, and resistant to slippage without premature joint line contact. A poorly designed brace could fail prematurely and concentrate forces at the joint line with the potential for increased ligament or bone damage."

Third, "Should I recommend lateral braces to my patients or athletes?" The response to this question was "no." Although the authors demonstrated biomechanically that it was possible to protect the knee against valgus injuries in limited situations, they could not recommend prophylactic knee bracing until adequate clinical trials had been conducted. France et al. (1987), stated, "Based on presently available data, no physicians, coach, or athletic organization should recommend these braces as mandatory equipment or, on the other hand, prevent their use."

Two other cadaver studies (Baker et al., 1987 and Hoffman et al., 1984), examined the static stability effect of knee braces on the MCL. Five braces classified as functional (DonJoy, Generation II-Poli-Axial, Lenox-Hill, Pro-am, and CTi) and 2 prophylactic braces (Anderson Knee Stabler and McDavid) were evaluated for their effect on abduction forces applied to 4 cadaver knees with no instability and with experimentally created medial instability (Baker et al., 1987). Force transducers were applied to the MCL and ACL. Abduction forces (0N to 30N) were applied perpendicular to the longitudinal axis of the tibia. Data were collected at 0°, 15°, and 30° of flexion, with and without knee braces. The results showed a reduction in abduction angle using functional braces, whereas prophylactic braces demonstrated little or no protective effect.

Knee instability, incurred by cutting the MCL in the cadaver knee, closely approximated the laxity seen following traumatic ligament tears or during surgical ligament repair. Significant differences existed among the 6 orthotic knee braces in their ability to stabilize ligamentous injuries of the knee (Hoffman et al., 1984). The differences in performance were attributed to brace design. Lateral and medial supports provided better stability than single posterior or posterior-anterior supports. Increased sidebar rigidity increased stability, but these observations were not quantified.

However, Brown et al. (1990), quantified the protection offered by prophylactic braces against impacts from lateral blows to the knee under ideal laboratory conditions. Brown et al. reported, "depending upon the brand, the braces absorbed from 15% to 30% of the force of the lateral blow."

Critique

Although cadaver and surrogate studies provide us with excellent information, the results can not be equated to on field performance. There are many other factors (examples being, proprioception, reflexes, muscular stability [even if limited], and the athlete's anticipation to contact) that can assist an athlete escape serious injury. Besides this comparison dilemma, one study raises the question regarding it's validity. Paulos' 1987 study, suggested that the ACL and the PCL disrupt prior to the MCL being torn during valgus loading. However, if this is true valgus loading then the MCL should be the first to disrupt followed by the ACL and then if the forces are severe enough, the PCL. Also, in this situation, since the medial meniscus is attached to the MCL, it also must be injured. The testing conducted by Paulos must have had some limb rotation.

Functional Knee Braces

In 1984, at the AAOS seminar on knee braces, Paulos introduced a classification system for functional knee braces that included two basic types. The first consisted of the hinge, post, and shell. Examples include the Generation II Poli-Axial (Generation II Orthotics), CTi (Innovation Sports), and Don Joy RKS (Don Joy Brace). The second type consists of the hinge, post, and strap. Examples include Lenox Hill (Lenox Hill Brace Shop), Feanny (Medical Design), and the Don Joy 4-Point (Don Joy Brace) (Wirth and DeLee, 1990).

Subjective and Clinical Functional Assessment

Subjective reports, by braces users, has been one method employed by researchers to evaluate the efficacy of functional braces. These reports, (usually a collection of responses to set questions) represented important insights into the capability and the acceptability of the brace(s) being tested. Whether these subjective reports were reflective of decreased knee laxity or performance became the focus of intense research in the mid 1980's (Branch & Hunter, 1990).

The derotation or ACL brace (Lenox Hill brace) was first mentioned by Nicholas in 1973. Nicholas collected data from 52 subjects. Of these patients 65% used an ACL brace preoperatively but found that the brace failed to control buckling and did not permit them to return to their sport. However, he noted that he had seen "many patients who did not wish to have surgical treatment after their instability was controlled by a brace" (Branch & Hunter, 1990).

In 1983, Basset and Fleming further tested the efficacy of the Lenox Hill brace. These authors reported that 70% of their subjects still complained of episodic giving way in sports requiring jumping, twisting, or cutting.

Colville et al. (1986), tested 45 subjects and observed that of the 45 patients, 28 (62%) continued to complain of knee instability while using the brace. Nearly 70% (31.5) of their patients felt it improved their athletic performance, and 91% (41) of their patients felt that the brace was beneficial to them. Of the 45 subjects, 60% (27) were part-time brace users and 40% (18) were full-time brace users. Part-time brace users wore the brace only for strenuous, twisting sports. However, "the full-time brace users never used the brace during jogging."

In another study, Mishra et al. (1989), examined 42 patients with unilateral knee injuries (23 right and 19 left). All patient had demonstrated ACL insufficiency in the injured knee by clinical examination and by instrumented testing with the MedicMedric (San Diego, California) KT-1000 Arthrometer. Thirty-one patients had arthroscopically proven ACL injuries; eight subjects had been treated with ACL reconstruction but still had functional disability; the other three subjects had pathologic anterior laxity and functional instability but had not been examined surgically.

Four types of braces were utilized. The Don-Joy Four-Point brace (Carlsbad, California), the RKS brace (Don-Joy, Carlsbad, California), the Lenox-Hill brace and the CTi brace (CTi, Irvine, California). Each patient had worn the brace for at least one month (mean duration, 9 months; range 1 to 37 months) (Mishra, et al., 1987).

The study questionnaires contained questions concerning comfort and suspension, specific task performance, pain and effusion during functional activities, athletic participation performance levels, and giving-way episode. Possible responses were none, tight, loose, slips, bulky, hot, abrasive, restricts range of motion (ROM), bruises, locks, or other.

For specific task performance, subjects were given questionnaires and asked to rate their performance both in and out of the brace in 14 specific tasks. These tasks were prolonged standing, walking, walking on uneven ground, climbing an incline, going upstairs, going down stairs, kneeling or squatting, jogging, running fast, stopping fast, jumping, twisting or pivoting, cutting, and getting out of a chair. Possible responses to the above tasks were:

(1) No problem

(2) Some difficulty

(3) Extreme difficulty

(4) Not able to do

(5) Not applicable

(6) Unknown.

An overall rating of good, fair, or poor were given based on a set criterion (Mishra, et al., 1989).

For pain and effusion during functional activity, patients were asked to evaluate symptoms of pain and swelling both with and without a brace (Mishra, et al., 1989).

For athletic participation performance levels, subjects were asked to evaluate themselves in 5 categories:

(1) Type of sport

(2) Pre-injury hours per year spent playing the sport

(3) Postinjury hours per year spent playing the sport

(4) An assessment of how well they are currently able to play the sport when wearing the a brace (pre-injury level was established as 100%)

(5) As assessment of how well they are currently able to play the sport without the brace.

The type of sport was categorized into three classes:

(1) Strenuous (football, basketball, and gymnastics)

(2) Moderate (Baseball/softball, jogging, and running)

(3) Mild (Walking, swimming, and cycling) (Mishra, et al., 1989).

For subjective functional testing, patients were required to do standing one-legged hop for a distance and a 36.57 meter (40 yard) shuttle run (Mishra, et al., 1989).

After the subjective functional testing, subjects were asked to cite the problems encountered with their functional brace. The primary brace problems cited by the subjects were:

(1) None (17/42)

(2) Slippage (13/42)

(3) Bulky (6/42)

(4) Hot (3/42)

(5) Bruising (1/42)

(6) Locking (1/42)

(7) Abrasions (1/42)

Concerning specific task performance, subjects reported the greatest number of problems with descending stairs, kneeling or squatting, stopping fast, jumping, twisting, or pivoting, and cutting. The number of subjects with poor or fair ratings was decreased (from 32/42 [76%] to 10/42 [24%]) with brace use, and as a result the number of subjects with good ratings increased (22/42 [52%]) (Mishra, et al., 1989).

Reports of pain and swelling decreased as a result of brace use; 14/42 (33%) of the subjects reported pain and swelling without use of the brace; only 6 (14%) subjects reported knee pain (mild and moderate only, no severe) with sports and 7 (17%) reported knee swelling (mild and moderate only, again no severe) with sports (Mishra, et al., 1989).

As for athletic performance, 24 (57%) subjects listed taking part in a strenuous sport, 18 (43%) listed taking part in a moderate sport. Prior to knee injury the mean hours per year of participation in their primary sport were 210; after knee injury with a brace the extrapolated mean hours per year equaled 122. Thirty-two patients listed a second sport, and there was a similar drop in the hours played post-injury. "Instead of switching from a strenuous sport preinjury to a moderate sport postinjury, patients preferred to play the same sport at the same intensity, but for fewer hours" (Mishra, et al., 1989).

According the Mishra et al., (1989), "of all the subjective criteria by patients to determine the efficacy of a brace, the phenomenon of giving way is the one that consistently received the greatest emphasis." In their study there was a reduction in the number of giving way episodes with the brace compared to without the brace--6/42 (14%) as compared to 24/42 (57%).

The next study was conducted in June of 1989 by Rink et al.. This study subjectively evaluated the ability of three functional knee braces (CTi [Innovation Sports], OTI [Omni Scientific] and the TS7 [Orthotech Incorporated]) to control anterorotary instability of the knee. They used 14 subjects, "weekend athletes", with conservatively managed arthroscopically proven ACL deficiency. Each subject utilized each brace for a period of one-month (worn at least six to eight hours per week) and then answered a questionnaire, and use of the next brace began. The questionnaire compared the braces for comfort, weight, running speed, slippage, overall condition, and whether the subject would wear the brace.

Rink et al., found that all braces reduced subjective symptoms of knee instability. However, each subject preferred a different brace.

Marans et al. (1991)[4], found statistically significant differences in the analysis of two of the three subjective criteria--stability and restriction of normal knee movement. Marans, defined subjective enhanced stability as "the perception that the brace makes the knee more stable and therefore gives

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[4] Marans' study design can also be found under the heading Performance Dynamic/Game-Like Testing below. It has been included in this section as part of the study was subjective evaluation of knee braces.

confidence that the knee is less likely to give way." Furthermore, Marans and colleagues felt that the "athlete's [perception] of knee stability may be a sensitivity assessment."

The subjects in Marans et al., (1991) study reported that the Lenox Hill, the Generation II, and the McDavid Knee Guard braces provided significantly greater subjective stability than the other three braces. On a scale of 5 (with 5 being very stable) the Lenox Hill was rated at 3.3, Generation II at 3.2, and the McDavid Knee Guard also at 3.2. With respect to restrictiveness of normal knee movement (for functional braces only), again on a scale of 5 (with 5 providing no restriction) the Generation II and the Lenox Hill braces rated the highest at 2.9, followed by the Don Joy RKS brace at 2.7 and then by the Don Joy 4-Point brace being rated at 2.6.

From these subjective studies it becomes evident that an ACL brace may be beneficial in reducing the frequency of giving way episodes (in Mishra's study 14% braced as compared to 57% non-braced) but not in eliminating them. Branch and Hunter (1990), conclude that the efficacy of an ACL brace is inversely related to the degree of athletic demands placed on it. They also state that sporting events that require stop-start, cutting, pivoting, or jumping actions will have an increased likelihood of causing instability with an ACL-deficient knee, even with an ACL brace. However, activities with a "straight-line" orientation not only expose an unstable knee to fewer chances at subluxation but also lower loads and thus "the result is an environment where bracing could be more beneficial" (Branch & Hunter, 1990).

Critique

Except for Nichloas' study in 1973, all studies conducted had good controls, validity, and variance. Nichloas conducted no tests but rather just reported on subjective verbal reports from his patients.

In Mishra's study, a high variance was noted in the brace accommodation period; it ranged from 1 month to 37 months. Again, most studies did not provide statistically analyzed data.

Energy Requirements

As various researchers have demonstrated, wearing a ACL brace requires additional energy. According to Branch and Hunter (1990), weight added to the feet tends to have larger impact on energy consumption than added elsewhere on the body. Therefore, since ACL braces can weigh as much as 2 pounds (Branch and Hunter, 1990) a greater demand can be placed on cardiovascular system.

Soule and Goldman (1969), demonstrated that wearing a 6 kg[5] weight on each foot while walking at 93 meters per minute (3.43 mph[6]) caused a 420% increase in energy expenditure.

Houston and Goemans (1982), reported that wearing an ACL brace might have a more immediate effect on the performance of the limb. They used 7 male athletes whom all used a prescribed hinged knee support brace. Three subjects had medial collateral ligament (MCL) instability, one subject had ACL instability, and the remaining three subjects had both MCL and

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5 Kilograms.

6 Miles per hour.

ACL instability. Four subjects wore the Lenox Hill brace, one the Toronto 2191 brace, and two the general fitting Kelly brace.

Testing was performed over a 4-week period, with at least one day rest between the testing days. Before testing all subjects were allowed to practice all procedures with and without their braces. Four tests were performed; isometric and isokinetic strength during knee extension, maximal unloaded angular velocity, vertical velocity and power via a short stair run, and an endurance test where a subject rode the bicycle ergometer at a fixed workload. All four tests were performed with and without a knee brace (Houston and Goemans, 1982).

In their first test they noted no differences in isometric strength associated with wearing a brace. However, differences were noted using the dynamic strength test. At slow speeds (30° per second) the braced limb produced 12% less torque (Nm) than the unbraced limb. This difference was increased to 30% less torque (Nm) at high speeds (300° per second). As for maximal angular velocity (maximal knee extension velocity) the braced slowed the knee speed by 15%. Furthermore, the braced subjects were slower (1.47 m/s as compared to 1.56 m/s[7]) and generated less power (129 kgm/s as compared to 138 kgm/s[8]) during the stair run. Lastly, during the 15-minute bicycle ergometer endurance test, the braced limb produced 40% greater end-trial blood lactate. (Houston & Goemans, 1982).

Zetterlund et al. (1986), took the above study one step further. Zetterlund et al., hypothesized that 2 pounds on one foot would produce a

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7 Meters per second

8 Kilograms meter per seconds.

30% increase in energy consumption. Moving the weight to the knee and increase the gait speed to 161 meters per minute (6 mph) results in a predicted increase of 5% in energy consumption.

Zetterlund et al. (1986), used 10 male subjects with arthroscopically proven ACL rupture for his study. All subjects had forgone surgical repair. Furthermore, only subjects with normal contralateral knee, ankle and hip function were used. The mean time of wearing the brace was 23.9 $\pm$ 28.0 months. Prior to the test each subject was familiarized with horizontal treadmill running at 161 meters per minute with and without the Lenox Hill Brace (LBH). The subjects were then tested randomly for four trials, two trials with the LBH and two trials without the LBH. Each run was at least 24 hours apart and performed at the same time of day. The subjects' running shoes and wearing apparel were kept constant for each run.

Expired air samples were collected between minutes 2 and 3 and also between minutes 5 and 6. Expired air was analyzed for oxygen and carbon dioxide concentrations. During the last 15 seconds and last 30 second of each air collection period the heart rate and the stride length were recorded, respectively.

Results recorded illustrated that overall oxygen consumption (L/min.) was increased by 4.58% and the heart rate was elevated by 5.10%. The ventilation (L/min. BTPS[9]) increased by 7.15%, the respiratory exchange ratio increased by 2.17%, and the stride length decreased by 0.72%. These findings raised the concern that increased energy consumption could have an impact on the endurance of the limb, and by reducing the limbs energy

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9 Body temperature and pressure, saturated.

reserves place the limb at risk of reinjury during a sporting event (Branch & Hunter, 1990).

In 1991, Highgenboten et al., expanded on the above study and tested asymptomatic subjects to determine the metabolic and the perceptual effects of wearing a functional brace while treadmill running at speeds of 6, 7, and 8 mph. The four functional braces used were the Generation II Poli-Axial Knee Cage (Generation II Orthotics Inc., Orange CA), the Orthotech Performer (Orthopedic Technology, Inc., San Leandro, CA), the CTi Brace (Innovation Sports, Irvine, CA), and the Lenox Hill Derotation Brace (Lenox Hill Brace Inc., Long Island City, NY).

Subjects performed six submaximal treadmill runs, on a motor driven treadmill, in a period of 2 weeks. Each subject ran for 5 minutes at 6 mph, 5 minutes at 7 mph, and 5 minutes at 8 mph. The first run was a familiarization run followed by 5 runs, performed in a random order, with or without the brace. The metabolic variables measured included oxygen consumption in liters per minute (LO_2) and in milliliters per kilogram per minute (MO_2), ventilation in liters per minute (V_e), and heart rate (HR). The perceptual variable used the Borg's scale to measure ratings of perceived exertion (RPE). RPE were taken at the end of the 5 minutes at each running speed and ratings were elicited for overall, central, and peripheral exertion (Highgenboten et al., 1991).

These authors found that wearing functional knee braces produced an elevated metabolic cost (oxygen consumption, heart rate, and ventilation) by 3% to 8% range compared to running without a brace in asymptomatic subjects. Furthermore, ratings of perceived exertion were elevated by 9% to 13%. However, no significant differences were found between the braces tested. These researcher's concluded that the weight of the brace accounted for the increase in metabolic cost.

Critique

All studies reported similar findings. Furthermore, all studies were conducted with low variability and thus high validity. Perhaps this high validity could be attributed to the vast amount of testing information that is available on energy expenditure studies.

Performance Dynamic/Game-Like Testing

The third method of evaluating the brace is its impact on athletic performance--dynamic or functional testing. Usually, prior to an injury athlete's find the a brace cumbersome or restrictive and thus an athlete ignores any suggestion(s) of utilizing a knee brace, even if strongly recommended by a physician. However, according to Branch and Hunter (1990), the impaired athletic performance during brace use may, infact, place the wearer at a higher risk of injury, by disrupting normal neuromuscular control patterns. Furthermore, this disruption would require a relearning process (if relearning is possible) to accommodate for the brace and that it is during this relearning process that the wearer "may" risk new injury (Branch and Hunter, 1990).

Functional, objective measurement of athletic performance has been limited to times or distances during isolated athletic tasks. These performance assessment tasks have been divided into two categories. For categories see Table 5.0.

In 1970 (Coran et al.), and then in 1979 (Reed), were the first researchers to address the subject of motor performance while wearing a supportive leg brace. Both of these studies reported an improvement in walking parameters. However, both studies were performed on hemiparetic and arthritic subjects--only remotely related to bracing and athletics.

Research, on this subject, was next performed in 1984. This unpublished study (presented at the Knee Brace Symposium) evaluated ten recreational collegiate athletes with and without the CTi brace. Subjects had no previous history of knee injury and all had a normal knee examination (Branch and Hunter, 1990).

The subjects utilized the CTi brace for 6 weeks prior to the study with a minimum usage of 8 to 10 hours per week. The subjects performed two category 1 (40-yard dash and the figure-of-eight run) and one category 2 (vertical jump) tests as well as quadriceps and hamstring isokinetic function evaluations. The researchers found no significant differences between the braced and the unbraced limbs. As a result, the researchers concluded that the brace did not limit normal physiologic function on the knee (Branch and Hunter, 1990).

The following year two studies were conducted. Tegner and Lysholm (1985), tested 42 subjects with documented ACL-deficient knees. Twenty-six subjects had chronic ACL-deficient knee, and 16 subjects had an ACL reconstruction with a subsequent normal knee examination. An ECKO

Table 5.0. Performance Assessment Categories

Category 1 -- Running	**Category 2 -- Jumping**
40 - Yard Dash	One-Leg Hop
Shuttle Run	Vertical Jump
Figure-of-Eight Run	
Stair Climb	
Slalom Circuit	
Cross-Cut Maneuver	

(Fonseca et al., 1992, Vailas et al., 1990, and Branch and Hunter, 1990).

(Orthomedics) brace was utilized for the study. The four tests performed were:

(1) Figure-of-eight totaling 40 meters (Category 1)

(2) Three one-legged hops (Category 2)

(3) Spiral staircase run of 25 steps (Category 1)

(4) Uphill-downhill run with a halfway turn (Category 1)

Results, found no statistical difference between the unstable and the reconstructed stable group in any of the events. An increased hop quotient suggested an increase in confidence by those patients with an unstable knee when using the brace (Branch and Hunter, 1990).

Also in 1985, Iglehart, used 10 male volunteers (20-30 years of age) to test the effect of the CTi Knee Brace on accommodation in athletic performance. Iglehart tested for muscular leg strength, agility, and speed and power. These parameters were tested by timed performance on the agility run, 40-yard dash, and the vertical jump.

Only subjects who participated in some athletic activity for at least ten hours per week were accepted. Furthermore, subjects had no history of previous knee injury. Each subject was provided with a custom fitting CTi supportive brace. Only the dominant leg was utilized. Initial testing took place on the day the braces were distributed, before the subjects had worn and experienced accommodation to the brace. Following the initial testing, the subjects were required to wear the brace for 2 to 3 hours per day and 5 days a week for a period of 6 weeks. The subjects were asked to perform movement tasks such as walking, running, cycling, or sporting activities, while wearing the brace. Post-testing was performed after 4 weeks of accommodation period and then again at the end of the sixth week (Iglehart, 1985).

Iglehart, concluded that "wearing a CTi Knee Brace has no significant ($p \leq 0.05$) effect on the strength and motor task performance of conditioned males." Igleharts reasoning on the discrepancies between her study and the Houston and Goemans study (1982) was four fold. One, Iglehart used a brace which weighed 14 oz.[10] less than the brace used by Houston and Goenman. Two, the CTi Brace has less skin-to-brace contact in the areas which may cause restriction in mobility and it has no restrictive straps crossing he joint line. Three, [probably most importantly] the fact that normal healthy subjects (with no previous knee injury) were used and as a result may not have been apprehensive of instability while testing. Lastly, accommodation, as testing was performed 3 times (instead of once, Houston and Goemans) and over a period of 6 weeks (instead of 4 weeks, Houston and Goemans), may have played a role as shown by Iglehart.

In 1989, Rink et al.,[11] tested "weekend athletes" in the timed figure-of-eight running event. Results form this test did not show any advantage of utilizing a functional knee brace. Five subluxation events occurred in four subjects while braced. As a result Rink et al. (1989), concluded that "functional braces appear to have a role in the ACL deficient knee, but only in conjunction with activity modification."

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10 Ounces

11 Methodology of the study Rink et al. (1989), conducted can be found above.

Another study conducted in 1989 was by Mishra et al..[12] These authors found that the hop test and the shuttle run showed some improvement in performance distance and times. The hop index[13] for the injured nonbraced leg was less than 80% in 26/42 of the subjects. When braced, 16/42 injured subjects had a hop index of less than 80%. However, no statistical significance was found. In the shuttle run, 79% of the males recorded a satisfactory time of 11.6 seconds. However, this number was increased to 86% with a brace. Fifty-six percent of the females recorded a time of 12.6 seconds or less (satisfactory rating) for both the braced and unbraced conditions. Overall, there was a mean time difference of 0.16 seconds between the no brace and braced conditions (Mishra, et al., 1989).

The next performance testing study was conducted in 1991 by Marans and colleagues. Their subject group consisted of 10 subjects. All subjects had symptomatic and arthroscopically proven ACL-deficient knees on one side only. All subjects were athletically active at the time of the study and had symptoms of functional instability during athletic activities but not during activities for daily living. Furthermore, all subject possessed a strongly positive pivot shift sign and had their untreated ACL lesion confirmed arthroscopically. All subjects had been advised to undergo ligament reconstruction; however, all had declined a surgical procedure. None had worn any type of brace before.

Six commonly used braces were chosen for the study (Table 6.0). Each brace was classified as either functional or custom made and

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12 The subjective functional assessment part of this study has been provided above.

13 The hop index is defined as the injured performance distance divided by the normal performance distance multiplied by 100.

prophylactic or off the shelf prophylactic. Furthermore, the braces were classified as laterally hinged or double hinged. Each subject was provided with six braces, three custom made to the patient's leg and three suitable fitted off the shelf. All functional braces were equipped with 15° extension stops.

Marans et al. (1991), modified (from Tegner and colleagues) and supplemented the activities which stressed the ACL (Table 7.0). These functional activities were quantified by measuring the time to perform a test or the distance achieved during the test.

Each subject performed all six of the functional tests, first unbraced and then wearing each of the six braces in random order. Random ordering was used so that the results were unbiased.

Results for objective performance measurement revealed that three braces significantly (in all but one performance test) improved performance when compared with the unbraced situations. In the straight-ahead 40 meter dash, the ACL-specific Don Joy braces were found to adversely affect performance (by 4-8/1000 of a second), compared with the unbraced knees. The Generation II and the McDavid Knee Guard had the least effect (1/1000 of a second) during the straight-ahead 40 meter dash. Performance in the figure-of-eight running and the acute angle cutting braced subjects completed the course 0.80 to 0.74 seconds faster as compared with the unbraced subject's unbraced. In the acute-angle cutting, again subjects utilizing the Generation II Poli-Axial Knee Cage managed the best results as compared to the unbraced subjects--0.84 seconds faster. Again, the Lenox Hill Derotation Brace scored second. All other tests were statistically non-significant (Marans et al., 1991).

Table 6.0. Knee-Brace Classification

Brace	Functional/ Prophylactic	Custom Made/ Off the Shelf	Type of Hinge
Don Joy Four-Point ACL Brace	Functional	Off the Shelf	Double
Don Joy Rotational Knee Stabilizer	Functional	Custom Made	Double
Lenox Hill Derotational Brace	Functional	Custom Made	Double
McDavid	Prophylactic	Off the Shelf	Lateral
Zimmer Double-Hinged Neoprene Sleeve	Prophylactic	Off the Shelf	Double
Generation II Poli-Axial Knee Cage	Functional	Custom Made	Lateral

(Modified from Marans et al., 1991.)

In a 1991, Veldhuizen et al., used a supportive knee brace (Push Brace 'Heavy') to evaluate the direct and long-run effects (accommodation) of this brace on leg performance. Eight healthy volunteers were used in the study. The testing protocol consisted of:

(1) Isokinetic muscle strength measurement

(2) 60 meter dash

(3) Vertical jump height test

(4) Treadmill running--Vmax and VO_2 at submaximal exercise.

Table 7.0. Dynamic Functional Tests Used In Marans et al., Study

Activity	Test
Straight-ahead Run	40 meter dash
Figure-of Eight Run	Around Pylon
Acute-Angle Cutting	Between Pylon
Agility Drill	Stimulated Running Through Tires
Forward Jump	One Leg Hop on Affected Leg
Lateral Jump	One Legged Hop on Affected Leg Towards Effected Side

(Modified from Marans et al., 1991)

Subject's performance was evaluated four times--three days prior to application of the brace, at day one and day 28 during the bracing period, and again one day after brace removal.

Veldhuizen et al. (1991), reported statistical significant differences on day one performance levels. Performance decreased by 4% in the 60 meter dash; Vmax was 6% lower; peak torque of knee flexion at 60°/sec and 120°/sec were 6% and 9%, respectively, lower; and peak extension torque at 60°/sec was 9% lower. All other tests showed only a slight decrease on day one.

However, after wearing the supportive brace for 4 weeks Veldhuizen et al. (1991), reported, "the test performances were practically identical to their base value." As a result, these authors concluded "performance in sports with test-like exercise patterns is not affected by the brace tested.

Bracing does not 'weaken the knee' at it is believed in sports practice and that brace wearer's go through a familiarization process."

Styf et al. (1992), conducted a study to determine the effect of functional bracing on intramuscular pressure during exercise in the anterior compartment of the leg. Intramuscular muscle pressure recordings were made at rest and muscle relaxation pressures[14] during exercise.

Eight subjects used three functional braces for the study:

1) The DonJoy Hinged Neoprene Knee Support (DonJoy Carlsbad, CA)

2) Omni II (Omni Scientific Inc., Lafayette, IL)

3) Bell-Horn Knee Sleeve (Bell-Horn, Philadelphia, PA).

Pressures were recorded with the microcapillary infusion technique while subjects were either supine, sitting, or standing.

Styf et al. (1992), found that pressure at rest increased significantly following application of each of the three knee braces regardless of posture. Similarly, muscle relaxation pressure during exercise also increased significantly on brace application. Furthermore, "the tested functional knee braces increased muscle pressures at rest and muscle relaxation pressure during exercise to levels that, according to other studies, might decrease muscle blood flow significantly." As a result, these researchers suggest that due to external pressure on the leg muscle there may be premature muscle fatigue because of local insufficient perfusion of the working muscle.

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14 Muscle relaxation pressure is defined as the pressure in the relaxed muscle between contractions during dynamic exercise.

Critique

All performance dynamic/game-like testing, except Coran's (1970) and Reed's (1979) research, have been conducted with low variability and high validity. Of these studies, the latter three (Marans et al., 1991 and Styf et al., 1992) provide the greatest insight of how a functional brace may perform during a game-like setting.

The one criticism with all these studies is that they focused on product testing (of available braces on the market) rather than on the athletic performance.

Kinematics

Kinematics involves the three-dimensional study of a subject's motion. Each limb segment (e.g. thigh, calf, foot) is identified in space and then the joint motion (connecting the limb segments) is calculated. In ideal circumstances all six degrees of freedom (three rotations and three translations) of a joint would be calculated from the data. In the knee, the three rotations would be flexion-extensions, varus-valgus, and internal-external rotation, while the translations would be anterior-posterior shear, medial-lateral shear, and joint compression-distraction (Branch & Hunter, 1990). However, according to these authors, it is very difficult to obtain all six degrees of freedom around a joint.

The earliest kinematic studies on absent ACL subjects concentrated on analysis to three rotations about the knee joint with the use of a three-degree-of-freedom goniometer. However, a goniometer has inherent restrictions imposed by the "umbilical cord" (Branch & Hunter, 1990). According to Branch and Hunter (1990), other researchers (Vailas et. al., 1990) have attempted to use high-speed cinematography, but have failed to duplicate those studies using electrogoniometers. However, with the development of video-based computerized motion analysis there should be greater accuracy in measuring unencumbered motion with six degrees of freedom.

Several studies, using an electrogoniometer on subjects with ACL-deficient knees and comparing them with normal controls, have demonstrated differences which occur with straight-ahead activity such as walking or running (Branch & Hunter, 1990). Kuntsen et al. (1987 & 1983), and Czerniecki et al. (1988), compared ACL-deficient subjects with normal control subjects during running. These authors used to goniometer which measured only three rotations. Both authors found flexion and total varus/valgus motion to be greater in the ACL-deficient group, while total rotation was less. However, none of the differences were statistically significant.

According to Branch and Hunter (1990), another study[15] analyzed walking using a goniometric technique whereby all six degrees were measured. These authors tested 20 ACL deficient and 30 control knees. There results also found no statistical difference between the two groups in flexion-extension, varus-valgus, internal-external rotation, medial-lateral shear, or joint compression-distraction. However, a statistically significant increase in anterior translation in the ACL-deficient subjects during the gait cycle was noted--with a mean difference of 4.7 mm. Unfortunately, the authors did not provide the KT-1000 results or make it clear at which walking or running phase(s) the increase in translation occurred.

Tibone et al. (1986), repeated the above study but used high-speed cinematography and hand digitization. There focus was on walking, running,

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15 Authors not identified by Branch & Hunter (1990).

and stair climbing. Again, as above, their results failed to show any statistical differences.

Further kinematic research between braced versus unbraced subjects was conducted at the University of Florida. Branch & Hunter (1990), used a Motion Analysis Corporation three-dimensional tracking system to measure compensatory kinematic changes between ACL-deficient subjects and normal (non ACL-deficient) subjects during a side-step maneuver. The ACL-deficient subjects kinematics were also analyzed using the strap-type (Don Joy) brace and the shell-type (CTi) brace.

Branch & Hunter (1990), reported that ACL-deficient subjects appear to hold their pelvis more anteriorly tilted, their hips slightly less abducted, and in slightly more external rotation on the stance or planted limb compared with normals. At the same time, the ACL-deficient subjects planted or stance side knee was measured to be in more varus and external rotation than normals. Furthermore, their stance or planted side ankle also was more externally rotated than in normals.

From this data, Branch & Hunter (1990), felt that "the cumulative external rotation of the hip, knee, and ankle in the ACL-deficient subject translates to a compensatory early turning of the body towards the cut" (see Figures 1.0 and 2.0). However, these authors found this data not to be statistically significant. As a result, Branch & Hunter (1990), suggested that subjects may employ a complex set of kinematic changes and thus the above measures may be statistical significant as a set but not individually.

Moreover, these authors produced some interesting kinematic changes in braced ACL-deficient subjects. The use of the ACL derotational knee brace during running produced a statistically significant decrease in knee flexion, by 22% (in the sagittal plane) during the swing phase and 13%

during the stance phase. Furthermore, a reduction in mediolateral varus-valgus movements of
24% as well as a 35% reduction in total tibial rotation was also documented.

Devita et al. (1992), conducted a study to assess the biomechanical effects of a functional knee brace on kinematics, ground reaction forces, as well as joint moments of force and joint powers in the lower extremity during the stance phase of running.

An unnamed brace made with an uniaxial hinge, post, and strap design was used for this study. The brace weighed 8.0 Newton. The aluminum tibial cuff was molded to fit each subject's leg, while the thigh was secured by a nonrigid cross-cell. Two groups were tested--non-injured subjects (termed healthy subjects) and injured subjects (termed ACL subjects). The ACL subjects were tested with and without the functional knee brace while, the healthy subjects were tested without the brace.

Kinematics analysis exhibited three results:

(1) The brace did not affect the kinematics of the ACL subjects

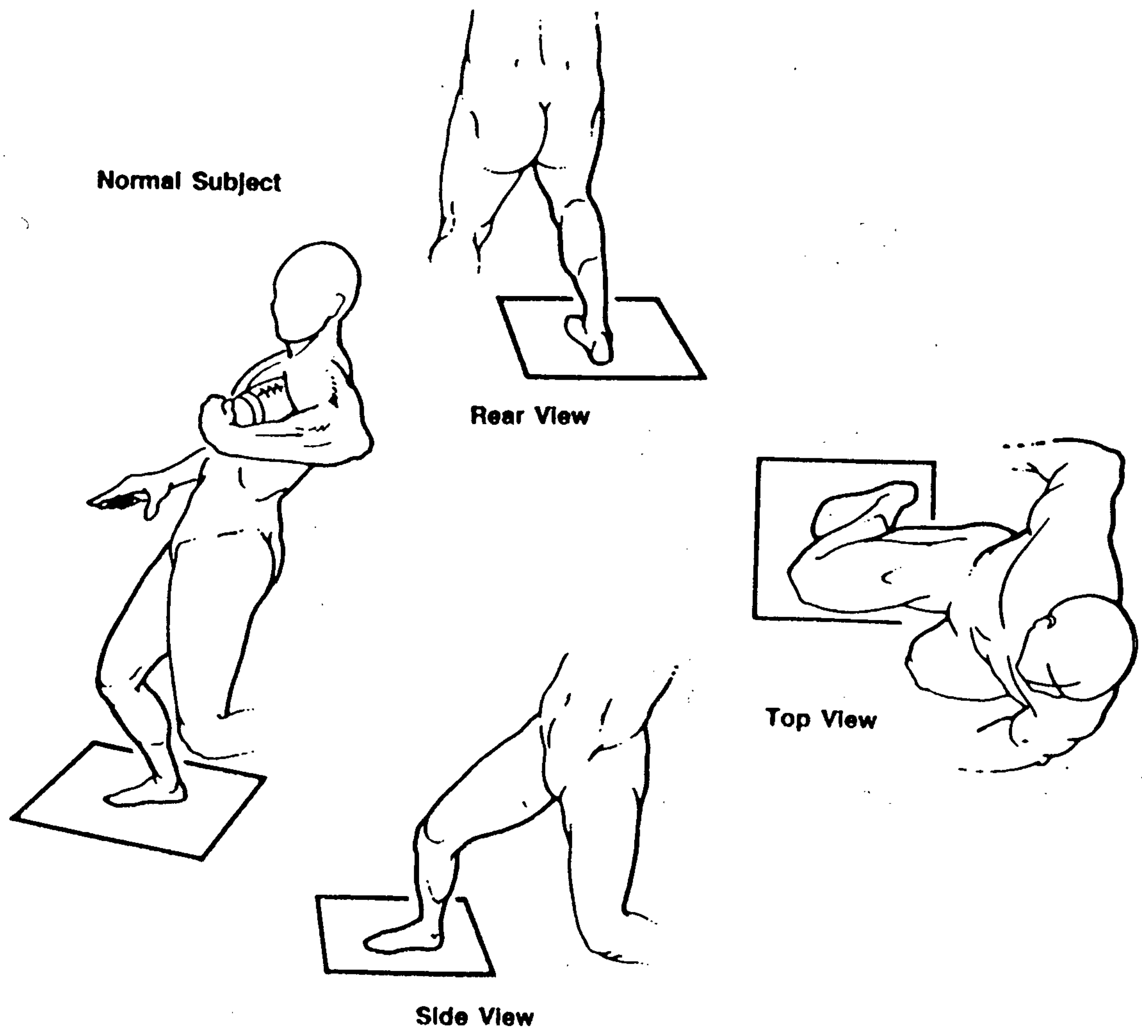
(2) In comparison to ACL subjects, the healthy runners flexed about 8° to 11° more at the hip and knee, respectively, throughout the stance phase

(3) The kinematic measure identified a more erect running style for the ACL-deficient subjects.

Critique

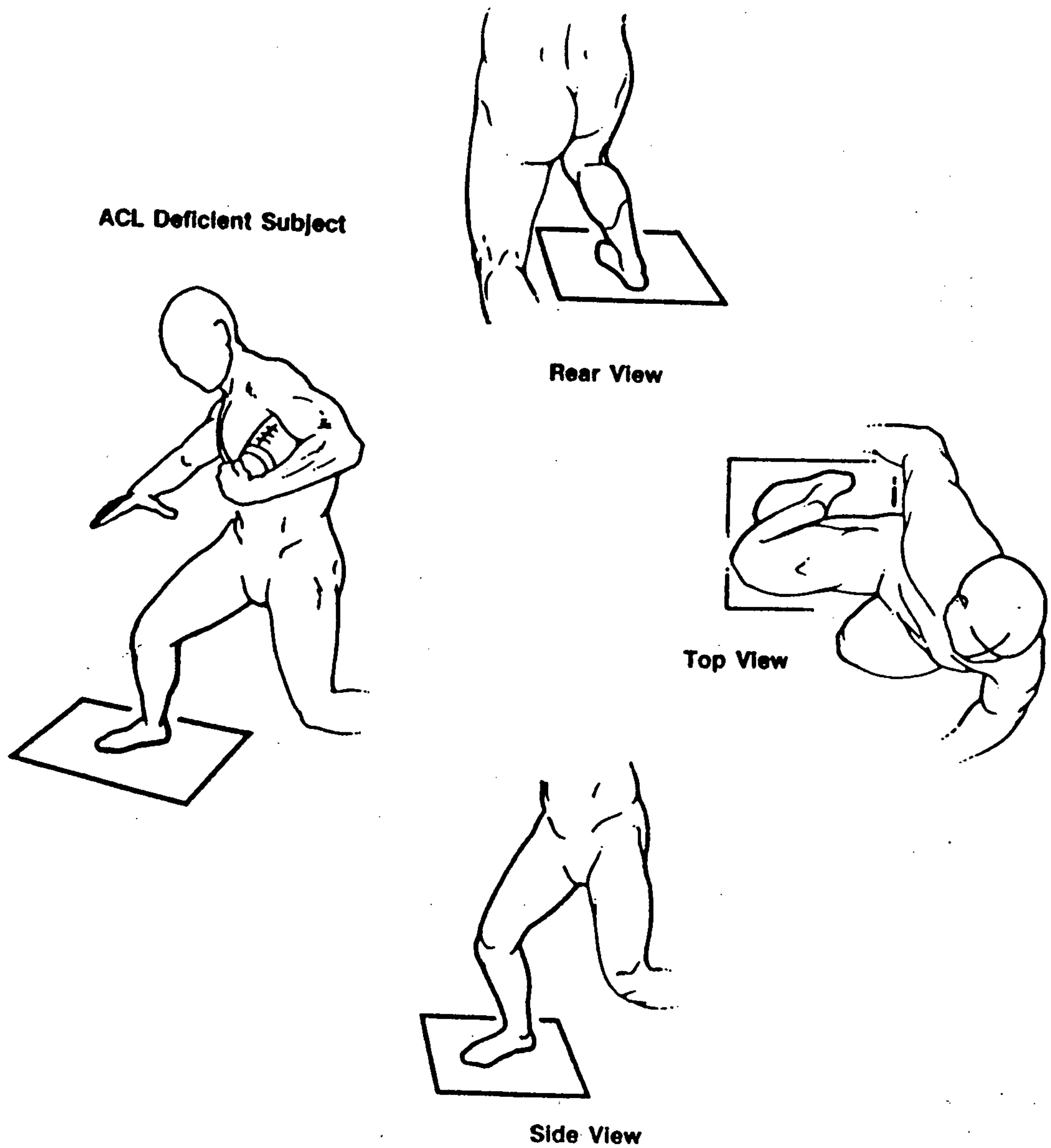
All kinematic data has been very valuable as it has allowed researchers to visualize how the body segments behave in a particular segment of time. For the research cited, studies performed using

Figure 1.0.
Kinematics Found During Mid-Stance of a Side-Step Cut in Normal Healthy Subjects.



From Branch and Hunter, 1990.

Figure 2.0
Kinematics Found During Mid-Stance of a Side-Step Cut in an ACL-Deficient Subject.



ACL-deficient subjects appear to hold their pelvis more anteriorly tilted, their hips slightly less abducted, and in slightly more external rotation on the stance or planted limb compared with normals. At the same time, the ACL-deficient subjects planted or stance side knee was measured to be in more varus and external rotation than normals. Furthermore, their stance side ankle also was more externally rotated than in normals (Branch and Hunter, 1990).

high-speed cinematography and hand digitization provide the best results. Studies done prior to the development of this technology were limited only by the equipment.

Force Plate Analysis

Ground reaction forces are those forces passed on to the foot by the ground during an activity. These forces are composed of the multiple forces generated by the body as a system during an event. The common forces measured are vertical forces, fore-aft shear and medial-lateral shear. Typical graphs of vertical forces and medial-lateral shear seen (respectively) are demonstrated in Figures 3.0 and 4.0.

In 1986, Tibone et al., studied 20 subjects with documented absent ACLs. They noted significant increase in stance vertical force (F2) in the involved limb during fast walking, and significant increase in the roll-off vertical force (F3) in the uninvolved limb during running. They surmised that the increased force during walking was an attempt to decrease forces across the joint (Figures 5.0 and 6.0). The decreased force on the involved knee joint was thought to be related to a midstance subluxation episode.

When subjects performed a cutting maneuver on the force platform several characteristics were noted. A cross-cut/closed-cut[16] with the uninvolved limb demonstrated a significantly sharper angle than with the involved limb (29.7° to 40.8°). Decreased lateral shear was noted in the involved limb during both the cross-cut and the side-step cut/open cut[17].

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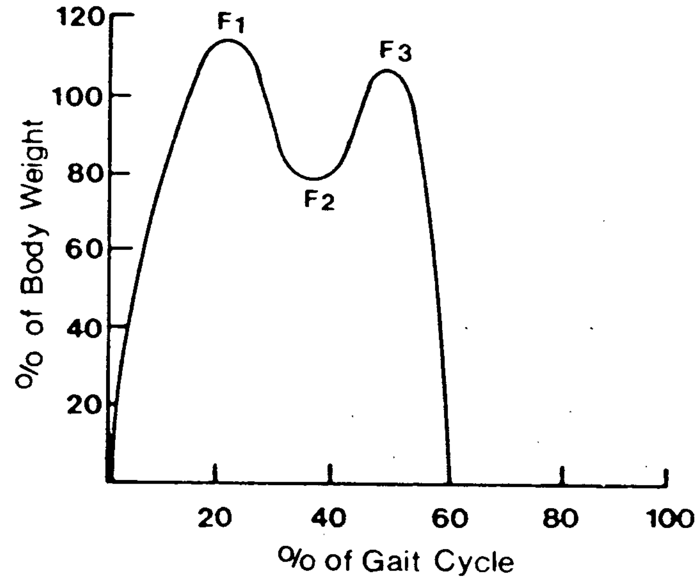
16 Cross-cut/closed-cut involves planting with the reference limb (ie. injured limb) and cutting to the planted side.

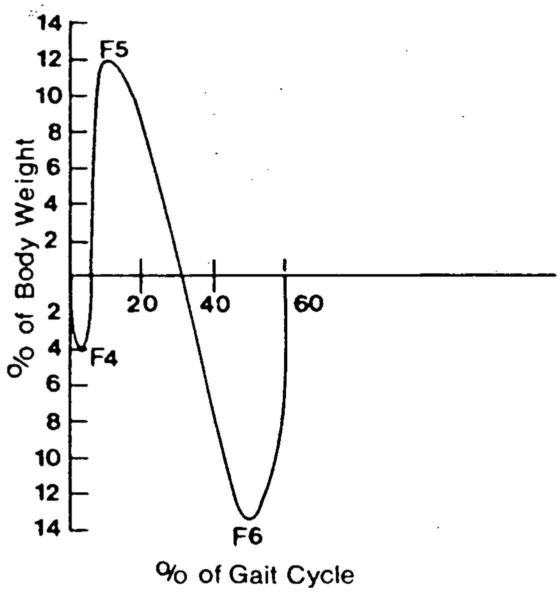
17 Side-step/open cut involves planting with the reference limb (ie. the injured limb) and cutting away from the planted limb.

Figure 3.0
Typical Vertical Forces Passed on to the Foot by the Ground During Activity



F1: Loading response **F2**: Single stance valley **F3**: Roll off peak

Figure 4.0
Typical Medial-Lateral Shear Forces Passed on to the Foot by the Ground During Activity



F4: Initial contact **F5**: Loading response maximum **F6**: Roll-off maximum. From Tibone et al. (1986).

Figure 5.0
Vertical Forces of ACL-Deficient and Normal Subjects During Fast Walking

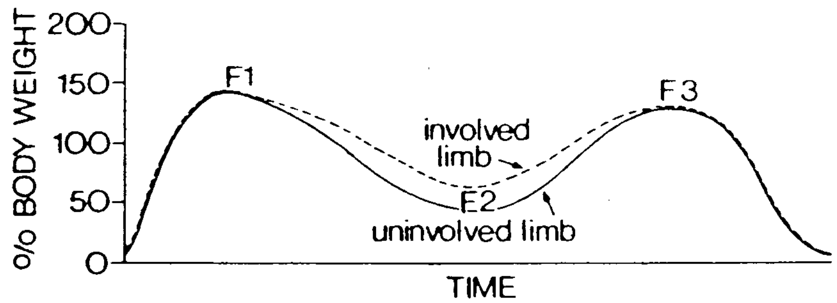
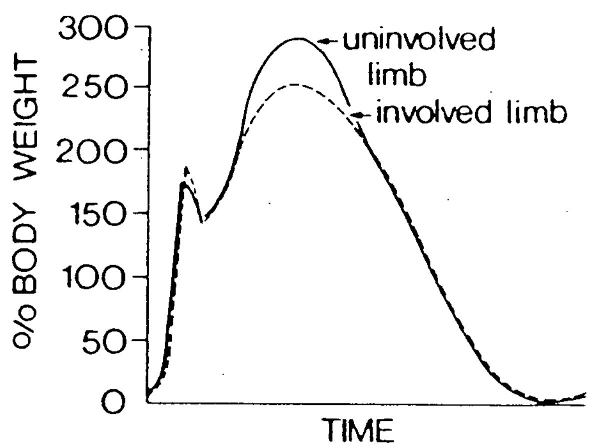


Figure 6.0.
Vertical Forces of ACL-Deficient and Normal Subjects During Running



From Tibone et al. (1986).

Vertical forces were significantly decreased during the side-step cut but not during the cross-cut, whereas a decrease in fore-aft shear was seen during the cross-cut but not the side-step cut. Using a nondimensional parameter called the "cutting index[18]," Tibone et al. (1986), were able to identify significant differences between the involved and the uninvolved limb. This cutting index attempted to take into account the multiple techniques that an individual with an ACL-deficient knee uses to survive a cut without subluxing the knee. These authors reported that the subjects with an ACL-deficient knee tend to utilize a slower approach to the cut, spend more time in the stance (plant) phase of the cut, reduce the angle of the cut, or exert less force on the planted leg during the cut.

Kunsten et al. (1987), performed a similar study however, they restricted their activity to straight-ahead running. Seven normal (control) subjects were compared with seven ACL-deficient subjects with and with-out two shell-type braces. When looking at vertical forces brace use increased the time to the first maximum force by 9% and the first minimum force by 13%. Furthermore, braced subjects had an increase in maximum braking force of 7% with an increase in time to maximum braking force of 6% was recorded. During the first three quarters of the stance phase, brace wearers had a statistically significantly increase in their total excursion of medial-lateral forces. In summary, in this study most of the significant changes occurred during the impact/loading phase of the supported period. The

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18 Cutting index defined.

$$\frac{(X)\ (Y)\ (Z\ \text{FORCES})(\text{ANGLE})}{(\text{TIME ON FORCE PLATFORM})\ (\text{APPROACH TIME})}$$

author's speculated that "these changes reflected a significant alteration in the gait pattern when wearing a brace." (Kunsten et al., 1987).

Devita et al. (1992), reported that functional knee braces did not effect the ground reaction forces of ACL subjects (subject with an ACL injury). The ACL subjects had larger maximum impact force in both conditions (braced and unbraced) compared with healthy runners, however, the differences were not statistically significant

In the most recent study Vailas et al. (1993), tested a placebo knee sleeve. A statistically significant decrease in torque was found when non-injured subjects used a derotational brace. ACL-deficient subjects had less torque on the involved limb than on the uninvolved limb when wearing a brace. No difference was noted when wearing the placebo. They did not report a statistical difference in torque between the braced and unbraced involved limbs.

Critique

As with kinematic studies, force platform studies have also provided excellent visual information on the forces encountered by the injured or non-injured and the braced or non-braced limbs. Furthermore, studies conducted by the four researchers above has provided a complete understanding of the limb(s) functional capabilities.

BRACE DESIGN[19]

The Generation II Poli-Axial knee brace consists of two shells/cuffs (thigh and gastrocnemius), with a laterally placed Poli-Axial hinge connecting them. The shell is constructed with a polyurethane material called (Pebax). The interior of the shell is lined by a closed-cell foam. This foam is blown with an inert gas to reduce the incidence of skin irritation.

The Poli-Axial hinge is made of high carbon steel (i.e. having the same characteristics as a car spring). The hinge consists of four different axes and can follow the anatomical movements of the knee with greater accuracy. Thus, the tibial portion of the brace (during flexion and extension) does not have to piston and migrate to accommodate for the short comings of the hinge. This allows for the "cuffs" of the brace to have greater contact with the brace. This hinge can be placed into an extra varus position and consequently place the knee into greater varus angulation. This will allow the knee to have greater protection from valgus injuries as the increased varus angulation has to be decreased before the medial structures are damaged.

As for hyperextension injuries, there is a three point pressure mechanism; proximal structure, distal structure, and the mid-strap located posterior to the popliteal fossa along with the hyperextension stop located within the hinge. The latter two structures especially, inhibit injurious movements.

The brace consists of four straps. First, the proximal cuff encircling the quadriceps muscles--the only elastic strap. Second, the mid-strap, in

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[19] All information was provided by Mr. D. Taylor, President of Generation II Orthotics, during an interview secession on March 16, 1992.

addition to hyperextension (as discussed above), is also utilized for vertical suspension of the brace. Third, the distal non-elastic strap, to secure the distal cuff on to the lower limb. Last, is the diagonal strap/dynamic force strap/the derotation strap.

The role of this derotation strap (as the name of the implies) is to limit derotation, especially on anteromedial rotatory motions. For posterolateral rotatory injuries the straps can be adjust in the opposite direction. The other two rotatory instabilities (anterolateral and posteromedial) are also inhibited by this brace.

PURPOSE OF THE STUDY

Significance of the Study

Previous knee bracing research has concentrated on one of the following parameters:

1) Subjective testing

2) Subjective and objective testing

3) Clinical testing

4) Cadaveric experiments

5) Biomechanical testing.

As Marans et al. (1991) states, the problem common to all these studies is the lack of testing in a dynamic setting. The degree of objective instability found by clinical testing has never been correlated to the degree of functional instability[20] that the patient may suffer from. Furthermore, most studies have concentrated their research to the effectiveness of one type of brace or a comparison between available braces (Marans et al., 1991). Moreover, all studies to date, involving functional knee braces, have focused on ACL deficient subjects or subjects with reconstructed (surgical intervention) ACL's.

Many authors state (Marans et al., 1991; Zetterlund et al., 1986; and Houston and Goenans, 1982) that knee bracing can impede an athlete's performance. The above authors felt that this may be the major factor in the

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[20] Functional instability is defined as instability during sporting activities or activities of daily living.

noncompliance often seen in high-caliber athletes. However, all the above studies showed only minimal time differences, one hundredths of a second slower in straight ahead running, (Marans et al., 1991) between the unbraced, ACL-deficient and the braced, ACL-deficient subjects. For all other tests the braced, ACL-deficient subjects performed better than the unbraced, ACL-deficient knee.

As a result of an increase in knee injuries in sports, researchers have to focus on preventing knee injuries instead of focusing on post-injury knee bracing. We know that the musculature is not able to react in time to protect the knee complex (Pope et al., 1979). Therefore, with the application of a functional brace perhaps the brace will help disperse the impact and/or stabilize the knee complex until the musculature surrounding the knee can offer further protection. If braces are able to offer this protection then we should not only see a reduction in knee injury severity, but possibly, a decrease in the rate of knee injuries.

Statement of the Problem

Before one can start to exploit the above hypothesis researchers have to convince the athlete how little functional braces restrict function. Therefore, the purposes of this study are to:

1) Evaluate the effect of functional knee bracing on athletic performance during dynamic testing in a non-injured knee.

2) Under dynamic testing, measure the effects of functional knee bracing on an ACL-deficient knee.

DEFINITION OF TERMS

Prophylactic knee braces	--	Usually an off the shelf brace with a lateral metal support. These braces are usually taped and/or strapped on to the limb.
Functional knee braces	--	Are custom-fit knee braces with either a lateral or lateral and medial support structures. These braces are usually only strapped on to the limb.
KT-1000 Arthrometer	--	An apparatus used to quantify knee joint laxity.
Dynamic tests	--	Tests which place an athlete in a "game-like" setting. Tests which allow the knee joint to be experience forces similar to those during a game. This does not include any externally applied forces to the knee joint.
Non-injured athletes	--	Athletes who do not have a prior knee injury.
ACL-deficient athletes	--	Athletes who have had an ACL injury. No other knee structure or ligament is damaged at the time of testing.
Competitive athlete	--	Athlete competing is sport(s) a minimum of 4 days per week--minimum 10 hours of activity per week.

LIMITATIONS, DELIMITATIONS, AND ASSUMPTIONS

This study was limited to the performance ability of each athlete. Although, each athlete was screened to be competitive, athletes may differ in their individual intensity and motivational levels. For the non-injured group, brace familiarization may have been a factor as some athletes may not have worn their functional brace during the brace accommodation period--10 days prior to the testing date. Also, for the non-injured group, the 'ideal' custom made brace provided some limitations, but could not be avoided. Lastly, the pooling of males and females subjects may have resulted in greater variance but could not be avoided due to the limited number of ACL-deficient subjects available for the study.

METHODOLOGY

Size/Sample Description

A total of sixty athletes were tested. Thirty (16 males and 14 females) non-injured 'normal/stable knee' and thirty (17 males and 13 females) ACL-deficient athletes were tested. The non-injured were fitted with an 'ideal' custom made Generation II Poli-Axial Knee Cage brace. The ACL-deficient athlete utilized their personal custom made Generation II Poli-Axial Knee Cage brace. The non-injured athletes had no history of knee or ankle injury (determined by consultation with the team Physician) as all non-injured athletes were varsity athletes. The ACL-deficient group included athletes with clinically diagnosed ACL injury that had not had reconstructive surgery. Diagnosis of ACL injury (a positive Anterior Drawer, Lachman, and pivot glide or shift) was determined by an orthopedic surgeon and/or primary care sports medicine specialist. For the ACL-deficient group, only athletes with a history of a Grade II (5-10 mm laxity [Booher & Thibodeau, 1985]) ACL injury were selected for the study. Prior to dynamic testing, each athlete had their knee laxity quantified as per KT-1000 testing[21] . All ACL-deficient athletes had rehabilitated their injury for a minimum of 6 months[22] and had utilized their brace (in competition a minimum of 4 days per week and 10 hours per week) also for a minimum of 6 months to a maximum of 24 months. In addition, athletes from both groups had no history of meniscal or PCL damage.

The groups were matched for:

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21 KT-1000 testing procedures will be identical to Branch & Hunter, 1990.

22 In allowing subjects to rehabilitate their knee for 6 months will allow the MCL to recuperate to functional levels and thus, testing will concentrate on the ACL.

(a) Age

(b) Activity level

(c) Weight

(d) Height[23]

Parameters Measured

Each athlete performed five functional tests--with and without the functional knee brace (Table 8.0). These functional activities were quantified by measuring performance times or performance distance.

Although straight-ahead running does not specifically stress the ACL, it was considered an important objective measure of the possible "restriction factor" that the brace might impose on the athlete's ability to perform at full capacity. The figure-of-eight running, slalom circuit, the hop test, and the running down the stairs test place the unstable ACL-deficient knee under maximal stress (Fonseca et al., 1992 and Marans et al., 1991) therefore, the same must be true for the stable knee.

As stated above, the straight-ahead running test was utilized only as a measure of possible constraints that the brace might impose on the athlete's ability. The hop test and running down a staircase tested the ACL under single plane forces. The last two tests, the figure-of-eight run and the slalom circuit tests were the most beneficial of all the tests as the ACL was placed under rotary stress (Tegner et al. 1986, Marans et al. 1991, and Fonseca et al. 1992).

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23 According to Generation II Orthotics Inc., height is a factor as the brace length has to be varied to provide the maximum amount of support to the knee.

Measurement Technique

Each athlete performed each functional test 8 times--2 submaximal practice runs (1 braced and 1 non-braced) to familiarize themselves with each test. The athletes then performed each test 3 times with a brace and 3 times without a brace (randomly assigned braced or unbraced trails). A three minute rest period was provided between each trial while a five minute rest

Table 8.0. Dynamic Functional Tests[24]

Straight Run
Figure-of-Eight Test
Slalom Test
Hop Test
Running Down Stairs

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24 Test descriptions.

1) Straight running -- A 10 meter straight running track with a minimum space of 10 meters allowed for deceleration after the end line.

2) Figure-of-eight -- 20 meter figure-of-eight running track with a diameter of both curves equal to 4 meters (Figure 7.0). Curves were equipped with high markers to prevent the subjects from executing turns in a shorter diameter than that stipulated. The starting point was located 2 meters from the beginning of one of the straight lines of the track.

3) Slalom circuit -- The slalom circuit is 20 meters long with 5 turns in its running track (Figure 8.0). The distance between turns will be 4 meters. Markers were carefully placed in each turn at an angle of 60°).

4) Hop test -- The hop test was performed over an area of 4 meters. A line will be used as a starting point. Standing on one leg with the toes behind the line (not in contact with any part of the line) the subject hops once to determine distance when landing on the same leg.

5) Running down stairs -- The athlete will start from the top step and descend 15 steps, one step a time. The athlete had side rails for support if needed.

Figure 7.0. Figure-of Eight Running Track

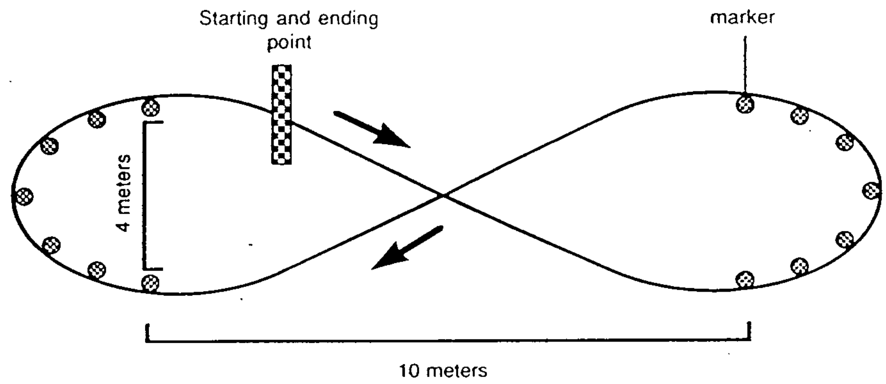
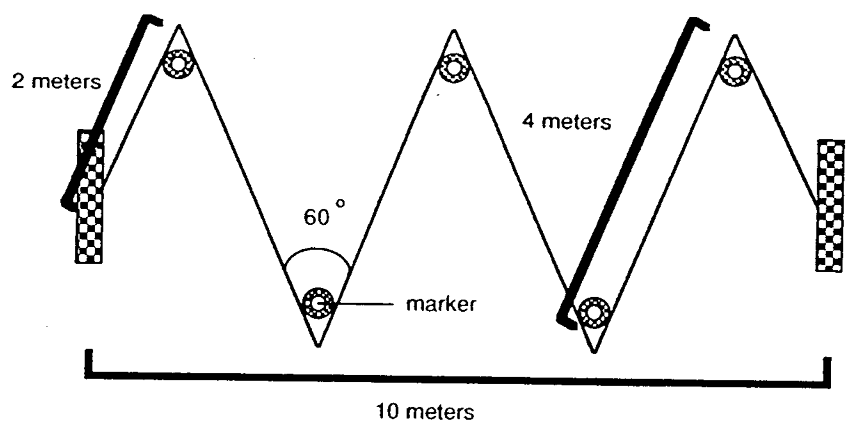


Figure 8.0. Slalom Circuit



period was provided between each test. All subjects followed the identical testing sequence--as listed in Table 8.0. All running times were recorded by using a hand held stopwatch. The mean performance time/distance over the three trials and well as the best performance time/distance achieved by the athlete were used for data collection and analysis. Also, recorded was the trail, for each test, the best performance time/distance was achieved on.

Statistical Analysis

A total of 60 athletes participated in the study. Three statistical analyses were conducted. First, a 2X3, repeated measure on both factors, ANOVA was conducted, on the means of tests and respective trials, during the accommodation phase. Second, a single factor ANOVA analysis was conducted on the best performance measures, after accommodation to the functional knee brace had occurred. The five above tests were performed between the "Non-Braced" and "Braced" conditions. Significance was determined at the $p<0.05$ level.

Third, a correlation analysis was conducted to investigate a possible relationship between the measured knee joint laxity of ACL-deficient athletes (via the KT-1000) and their performance times/distance.

The dependent variable was the performance times at each of the tests. The independent variable was effectiveness of the bracing mechanism.

Possible threats to internal validity were:

Measuring instruments --	Reduced as all instruments were calibrated (using a test specimen).
Differential selection of subjects --	All subjects were competitive athletes.
Maturation Process --	Eliminated as all subjects were between 19-35-years of age.

Pretesting procedures --

Efforts were made to eliminate this. All non-injured subjects were fitted with an 'ideal' custom fit brace. This brace was provided to them 10 days prior to the testing. All subjects were encouraged to utilized the brace for all activities, except when sleeping.

RESULTS

A total of 260 (215 injured and 45 non-injured) potential subjects were contacted from which 100 (59 injured and 41 non-injured) were tested. The other 169 were not tested because 93 subjects did not reply back to the initial letter mailed to them while the remaining 76 individuals had had surgery. Data from 40 subjects was not used for this study due to subjects dropping out (not being comfortable with the testing--33), reporting a PCL injury during or after completing the functional testing (6), and due to brace malfunction (1). Data from the 33 subjects that dropped out of the study was not used because 15 subjects did not complete the required number of trials for each test,[25] 13 subjects could not stay for the entire testing session due to prior commitments, and 5 subjects could not complete all the tests due to fatigue.

The mean age for both groups (females [Table 9.0] and males [Table 10.0] was 25 $\pm$ 2 years. The mean height for the females in the non-injured group was 170 $\pm$ 3.5 centimeters while females in the injured group were 3.0 $\pm$ 4.5 centimeters shorter. For males, the mean height (174 $\pm$ 3.0 centimeters) for subjects in the non-injured group was 2.0 $\pm$ centimeters greater than the mean height for subjects in the injured group (172 $\pm$ 4.0 centimeters). The mean weight for injured females subjects was 72 $\pm$ 3.0 kilograms while the mean weight for the non-injured females was 4.0 kilograms less (68 $\pm$ 3.5 kilograms). For the males, the mean weight of the injured group was 2.0 kilograms less (81 $\pm$ 4.0 kilograms) than the mean

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[25] Nine subjects missed one or more steps when performing the running down the stairs test, 5 subjects tripped over a pylon, and 1 subject tripped for an unknown reason.

weight for the non-injured group (83 $\pm$ 3.5 kilograms). None of the above descriptive statistics were statistically different.

Table 9.0. Mean Age, Height, and Weight for Females in Both Groups (N=27)

	AGE			HEIGHT[26]			WEIGHT[27]		
	High	**Low**	**Mean**	**High**	**Low**	**Mean**	**High**	**Low**	**Mean**
INJURED (N=13)	35	19	25	185	162	167	81	61	72
NON-INJURED (N=14)	30	22	25	172	154	170	72	60	68

Table 10.0. Mean Age, Height, and Weight for Males in Both Groups (N=33)

	AGE			HEIGHT			WEIGHT		
	High	**Low**	**Mean**	**High**	**Low**	**Mean**	**High**	**Low**	**Mean**
INJURED (N=17)	35	20	25	179	159	172	113	73	81
NON-INJURED (N=16)	35	21	25	186	162	174	99	73	83

Accommodation Phase

Interaction Within Non-Injured Subjects

Braced Versus Non-Braced

A 2X3, repeated measures on both factors, ANOVA analysis was

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26 All height measurements are in centimetres.

27 All weight measurements are in kilograms.

performed at the ($p < 0.05$) level. Statistically significant difference was noted within the braced and the non-braced groups for four of the five functional tests--only running down the stairs test showed no statistically significant difference (Table 11.0 and Figures 9.0 - 13.0). The differences in performance means within these two groups was as follows: (1) 10 meter dash--0.041 seconds; (2) figure-of-eight test--0.113 seconds; (3) slalom test--0.095 seconds; (4) hop test--(-0.024) centimeters; and (5) running down stairs--(0.003) seconds. In three of the four statistically significant tests (10 meter dash, figure-of-eight test, and the slalom test) subjects' performance was slower when wearing the brace; while in the hop test (the fourth statistically significant test) subjects' performance was superior when wearing the brace. For running down the stairs test, the two groups had almost identical performance times.

Table 11.0. Interaction of Means Within Non-Injured, Braced and Non-Injured, Non-Braced Subjects

TEST	p Value	Significant/Non-Significant
10 Meter Run	0.004	Significant
Figure-of-Eight Run	0.001	Significant
Slalom Run	0.006	Significant
Hop Test	0.007	Significant
Running Down Stairs	0.928	Non-Significant

Interaction Within Injured Subjects Braced Versus Non-Braced

Statistical significance ($p < 0.05$ level) was also noted within the braced and the non-braced groups for all five functional tests (Table 12.0 and

Figures 9.0 - 13.0). The differences within the means of braced and non-braced groups was as follows: (1) 10 meter dash--(-0.10) seconds; (2) figure-of-eight test--(-0.141) seconds;(3) slalom test--(-0.133) seconds; (4) hop test--0.046) centimeters; and (5) running down stairs--(-0.076) seconds. For all tests, braced subjects' performance was better with the brace than without the brace.

FIGURE 9.0. 10 METER DASH--COMPARISON BETWEEN NON-INJURED, BRACED/NON-BRACED AND INJURED, BRACED/NON-BRACED MEANS

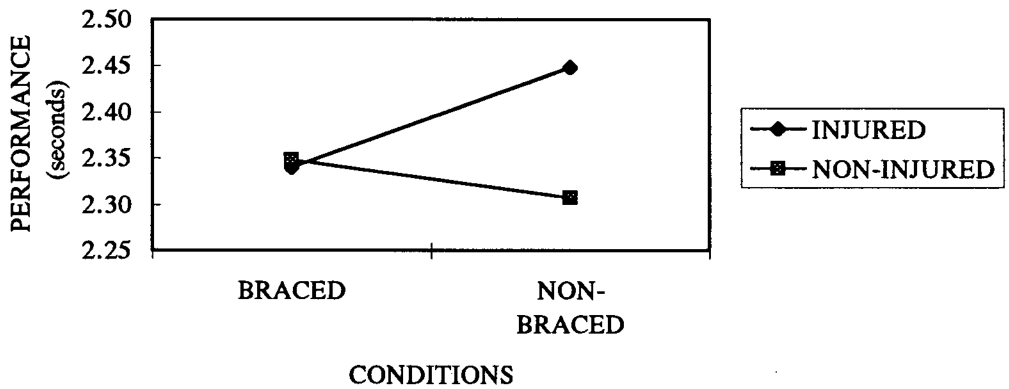


Table 12.0. Interaction of Means Within Injured, Braced and Injured, Non-Braced Subjects

TEST	p Value	Significant/Non-Significant
10 Meter Run	0.016	Significant
Figure-of-Eight Run	0.013	Significant
Slalom Run	0.001	Significant
Hop Test	0.005	Significant
Running Down Stairs	0.001	Significant

FIGURE 10.0. FIGURE-OF-EIGHT RUN--COMPARISON BETWEEN NON-INJURED BRACED/NON-BRACED AND INJURED BRACED/NON-BRACED MEANS

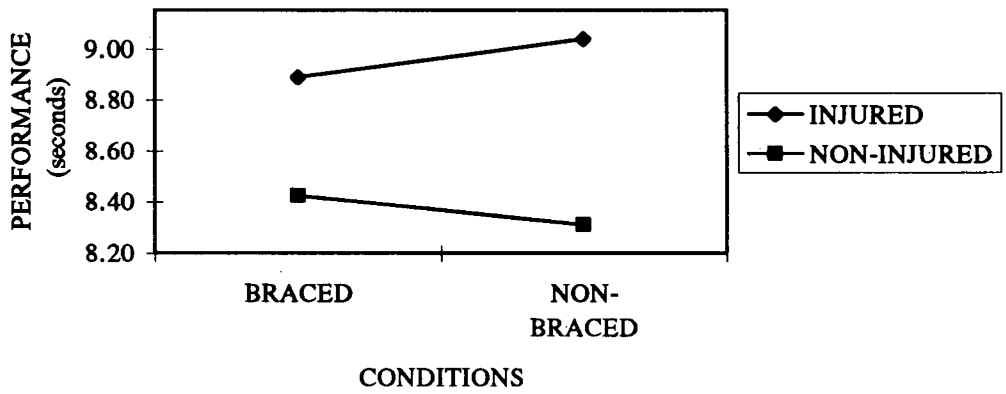


FIGURE 11.0. SLALOM RUN--COMPARISON BETWEEN NON-INJURED BRACED/NON-BRACED AND INJURED BRACED/NON-BRACED MEANS

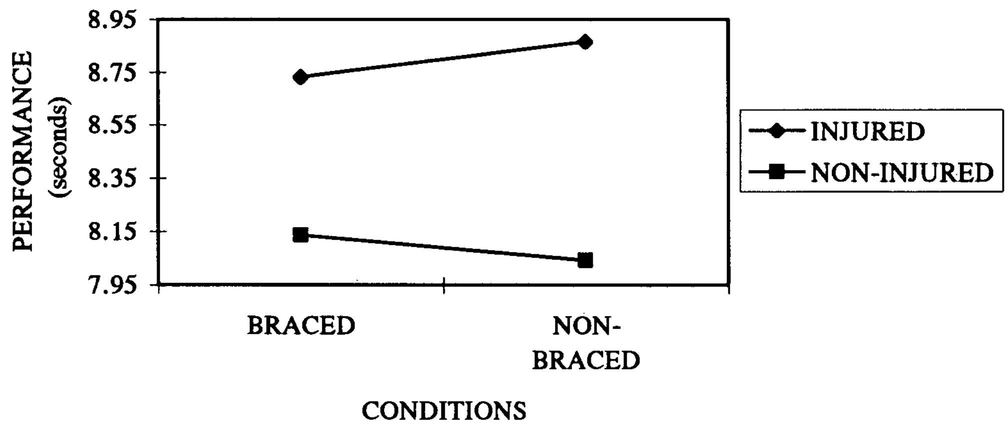


FIGURE 12.0. HOP TEST--COMPARISON BETWEEN NON-INJURED BRACED/NON-BRACED AND INJURED BRACED/NON-BRACED MEANS

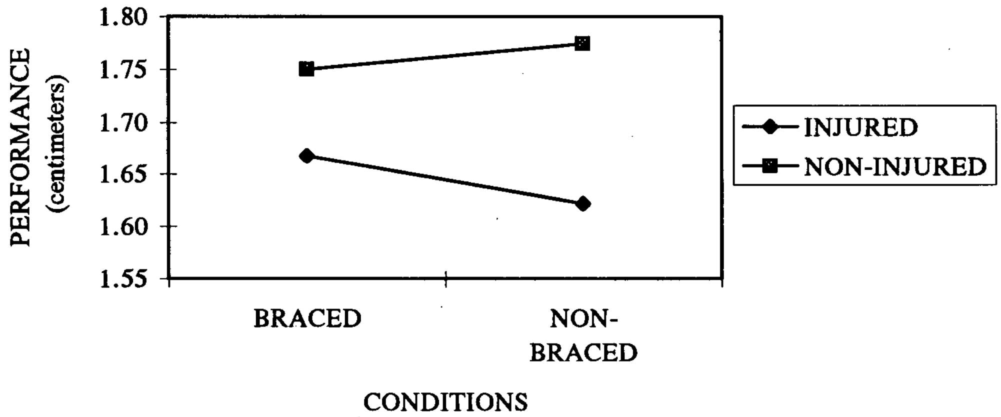
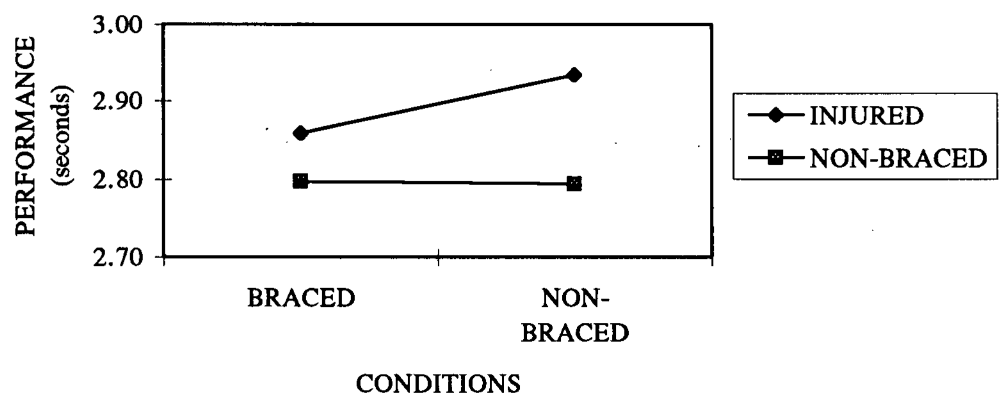


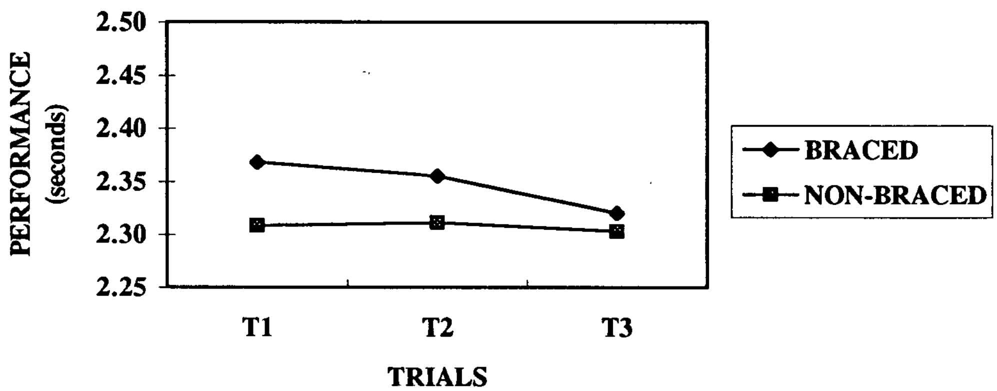
FIGURE 13.0. RUNNING DOWN STAIRS--COMPARISON BETWEEN NON-INJURED BRACED/NON-BRACED AND INJUREDBRACED/NON-BRACED MEANS



Interaction Within Non-Injured Subjects and Trials
Braced Versus Non-Braced

Statistically significant differences ($p<0.05$) were also found within the means of braced and non-braced conditions when interacted with respective trials. This statistical significant difference was noted for three of the five functional tests. The 10 meter run and the hop test were found to be non-significant (Table 13.0 and Figures 14.0 - 18.0). In the 10 meter dash, the best performance time for both groups was recorded on trial three; 2.320 seconds for the braced group and 2.303 for the non-braced group (Figure 14.0). In the figure-of-eight run, again, the best performance times were

FIGURE 14.0. 10 METER RUN--INTERACTION WITHIN NON-INJURED BRACED/NON-BRACED CONDITIONS AND TRAILS



noted on the third trial and there was only 0.013 seconds differences between the two groups (Figure 15.0). In the slalom test, the braced group followed the above trend of recording the best performance time on the third trial, while the best performance time for the non-braced group was recorded on the second trial. The best performance time between the two groups differed by 0.039 seconds (Figure 16.0). In the last two tests, braced subjects performed better than the non-braced subjects (Figures 17.0 and 18.0 respectively). In the hop test, braced subjects jumped 0.004 centimeters further than their counter parts--the braced subjects recorded their best jump on the third trial while the non-braced subjects recorded their best jump on the first trial. In the running down the stairs test, the best performance time for braced subjects was on the third trial (2.373 seconds), as compared to the second trial for non-braced subjects (2.793 seconds)--braced subjects performing better by 0.048 seconds.

FIGURE 15.0. FIGURE-OF-EIGHT RUN--INTERACTION WITHIN NON-INJURED BRACED/NON-BRACED CONDITIONS AND TRIALS

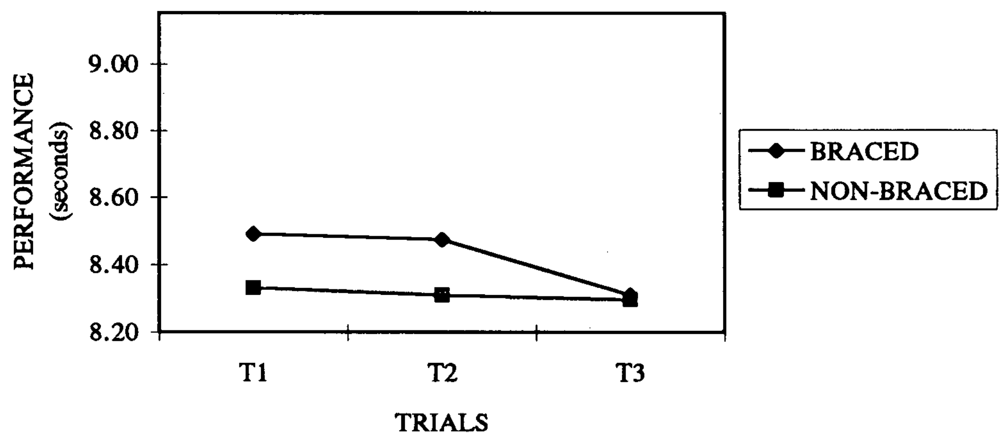


FIGURE 16.0. SLALOM RUN--INTERACTION WITHIN NON-INJURED BRACED/NON-BRACED CONDITIONS AND TRIALS

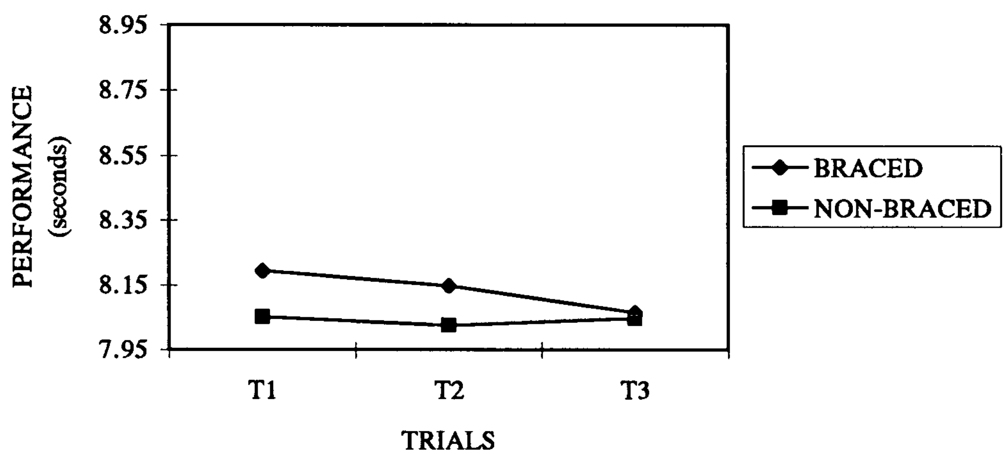
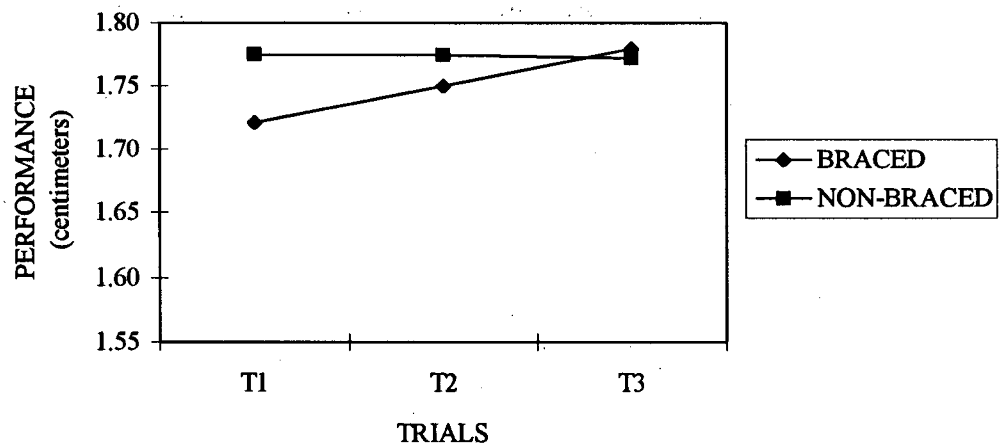


Table 13.0. Interaction of Means Within Non-Injured, Braced and Non-Injured, Non-Braced Conditions and Trials

TEST	p Value	Significant/Non-Significant
10 Meter Run	0.274	Non-Significant
Figure-of-Eight Run	0.006	Significant
Slalom Run	0.008	Significant
Hop Test	0.052	Non-Significant
Running Down Stairs	0.031	Significant

FIGURE 17.0. HOP TEST–INTERACTION BETWEEN NON-INJURED BRACED/NON-BRACED CONDITIONS AND TRIALS



Interaction Within Injured Subjects and Trials Braced Versus Non-Braced

When means of braced and non-braced subjects were statistically analyzed, again at the ($p<0.05$ level) over their respective trials, no statistically significant difference was noted. This result was observed for all five functional tests (Table 14.0 and Figures 19.0 - 23.0). The mean value for both the braced and non-braced subjects follows a general trend for performance enhancement with respect to the number of trials as the best performances were recorded on the third trial. The only exceptions to this trend are the subjects in the 10 meter run and the slalom test where the best performances are noted on trial two.

FIGURE 18.0. RUNNING DOWN STAIRS--INTERACTION WITHIN NON-INJURED BRACED/NON-BRACED CONDITIONS AND TRAILS

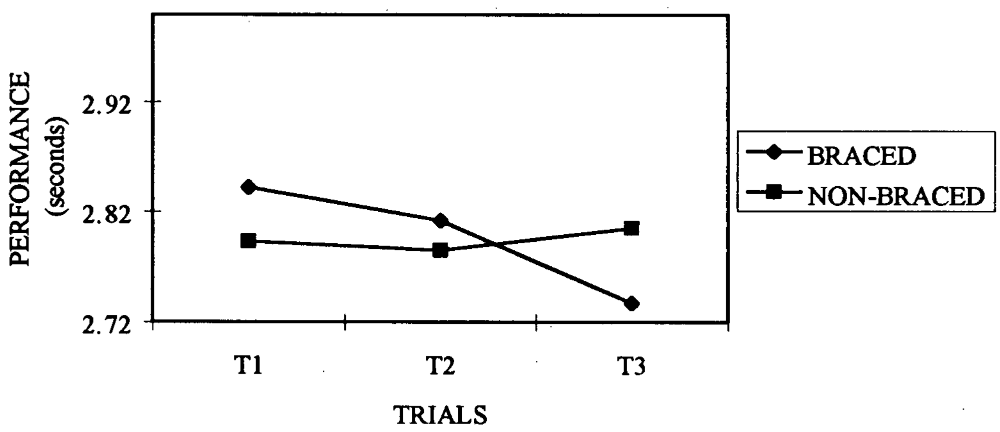


Table 14.0. Interaction of Means Within Injured, Braced and Injured, Non-Braced Conditions and Trials

TEST	p Value	Significant/Non-Significant
10 Meter Run	0.103	Non-Significant
Figure-of-Eight Run	0.210	Non-Significant
Slalom Run	0.345	Non-Significant
Hop Test	0.292	Non-Significant
Running Down Stairs	0.887	Non-Significant

FIGURE 19.0. 10 METER RUN--INTERACTION WITHIN INJURED BRACED/NON-BRACED CONDITIONS AND TRAILS

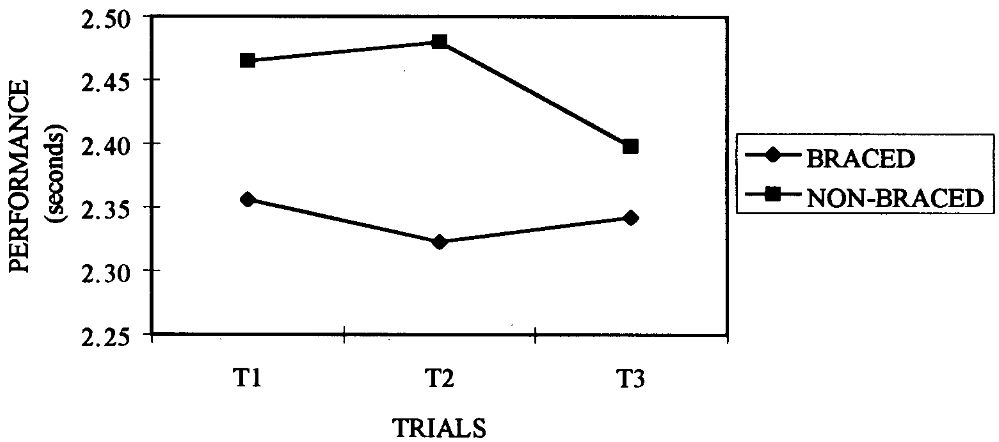


FIGURE 20.0. FIGURE-OF-EIGHT RUN--INTERACTION WITHIN INJURED BRACED/NON-BRACED CONDITIONS AND TRIALS

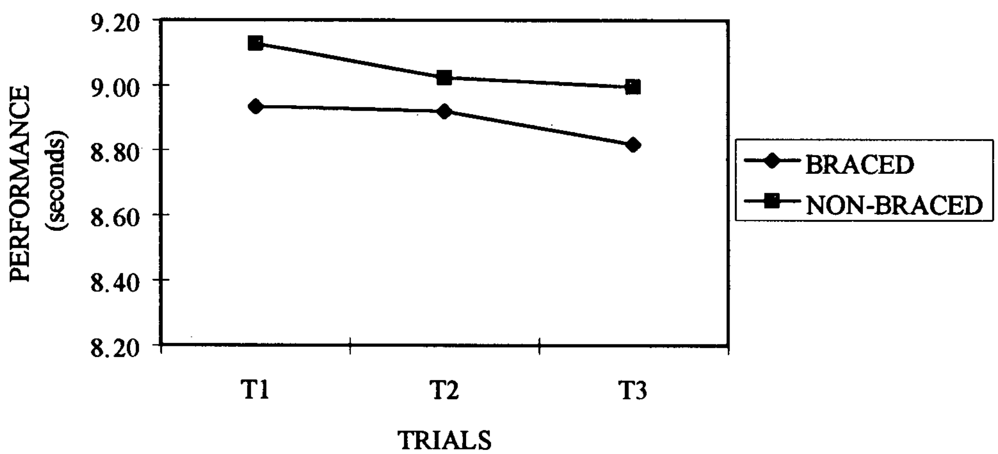


FIGURE 21.0. SLALOM RUN--INTERACTION WITHIN INJURED BRACED/NON-BRACED CONDITIONS AND TRIALS

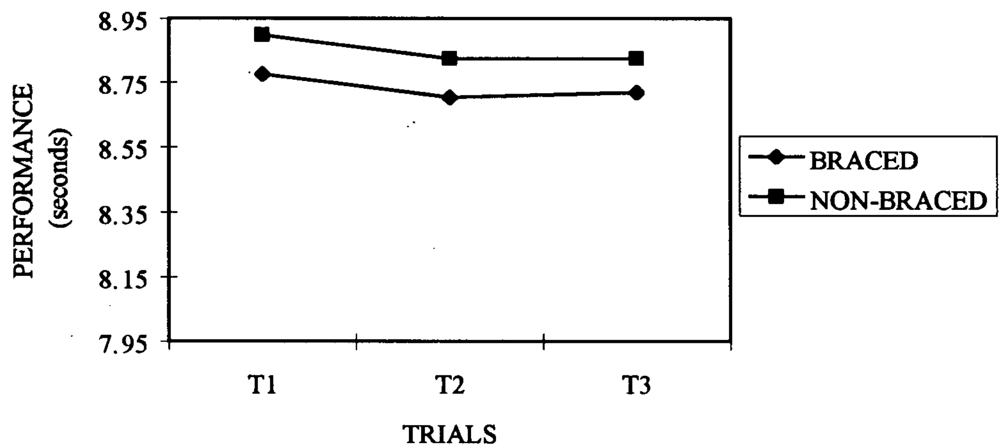


FIGURE 22.0. HOP TEST--INTERACTION BETWEEN INJURED BRACED/NON-BRACED CONDITIONS AND TRIALS

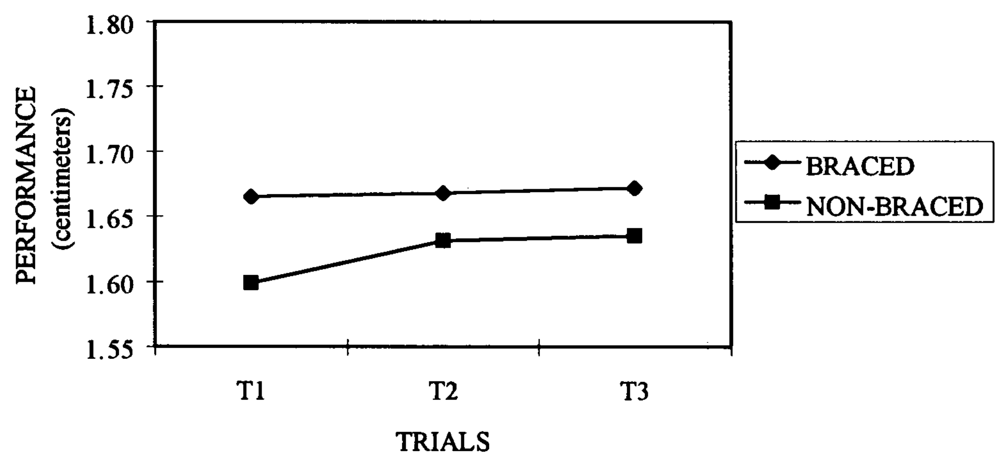
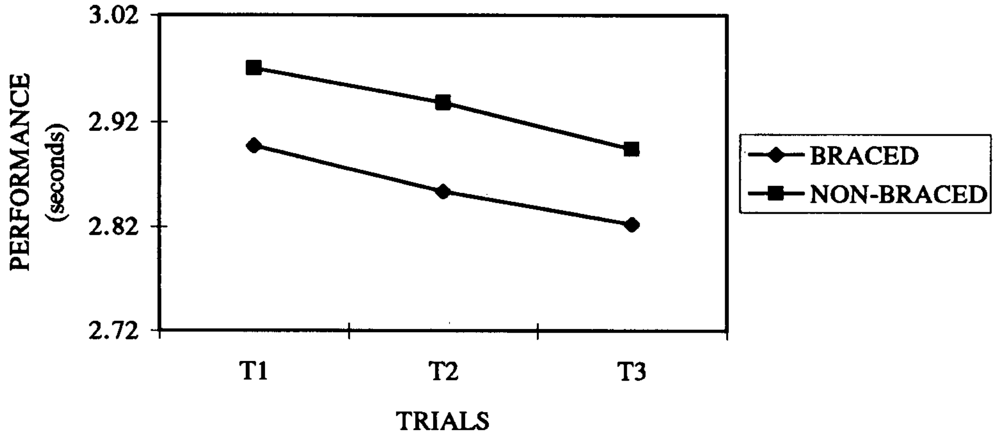


FIGURE 23.0. RUNNING DOWN STAIRS--INTERACTION WITHIN INJURED BRACED/NON-BRACED CONDITIONS AND TRAILS



Post Accommodation Phase

Best Performance Analysis

Non-Injured Subjects--Braced and Non-Braced

A single factor ANOVA statistical analysis was performed ($p<0.05$) on the best performance times and distance of braced and non-braced groups for the five functional tests. No statistically significant differences were found between the two groups (Table 15.0). In the 10 meter dash, the figure-of-eight test, and the slalom test, the non-braced group had superior performance times of 0.24, 0.08, and 0.06 seconds, respectively, when compared with performance times of the braced group. In the hop test and the running down the stairs test the braced group had superior performance than the non-braced group. The performance differences were 0.01 centimeters and 0.13 seconds, respectively.

Injured Subjects--Braced and Non-Braced

A second single factor ANOVA statistical analysis was performed ($p<0.05$) on the best performance times or distance of braced and non-braced subjects for all five functional tests. Again, no statistically significant differences were found between the two groups (Table 16.0). For all tests the braced group achieved superior results. In the 10 meter dash, a 0.09 seconds difference was noted; 0.16 seconds for the figure-of-eight test; 0.20 seconds for the slalom test; 0.05 centimeters for the hop test; and 0.12 seconds for running down the stairs test.

Table 15.0. Best Performance Time and Distance Interaction Between Non-Injured, Braced and Non-Injured, Non-Braced Subjects

TEST	p Value	Significant/Non-Significant
10 Meter Run	0.193	Non-Significant
Figure-of-Eight Run	0.618	Non-Significant
Slalom Run	0.682	Non-Significant
Hop Test	0.958	Non-Significant
Running Down Stairs	0.174	Non-Significant

Table 16.0. Best Performance Time and Distance Interaction Between Injured, Braced and Injured, Non-Braced Subjects

TEST	p Value	Significant/Non-Significant
10 Meter Run	0.173	Non-Significant
Figure-of-Eight Run	0.392	Non-Significant
Slalom Run	0.453	Non-Significant
Hop Test	0.479	Non-Significant
Running Down Stairs	0.357	Non-Significant

Correlation Between Knee Joint Laxity and Performance

There was no significant correlation found between knee joint laxity and performance times or distance for either the injured, braced (Table 17.0 and Figures 24.0 to 28.0) the injured, non-braced groups (Table 18.0 and

Figures 29.0 to 33.0).

Table 17.0. Correlation Between Knee Joint Laxity and Performance For Injured, Braced Subjects

TEST	r Value
10 Meter Run	0.145134
Figure-of-Eight Run	0.137367
Slalom Run	0.232399
Hop Test	-0.256056
Running Down Stairs	0.017487

FIGURE 24.0. 10 METER DASH--CORRELATION BETWEEN KNEE JOINT LAXITY AND INJURED BRACED SUBJECTS

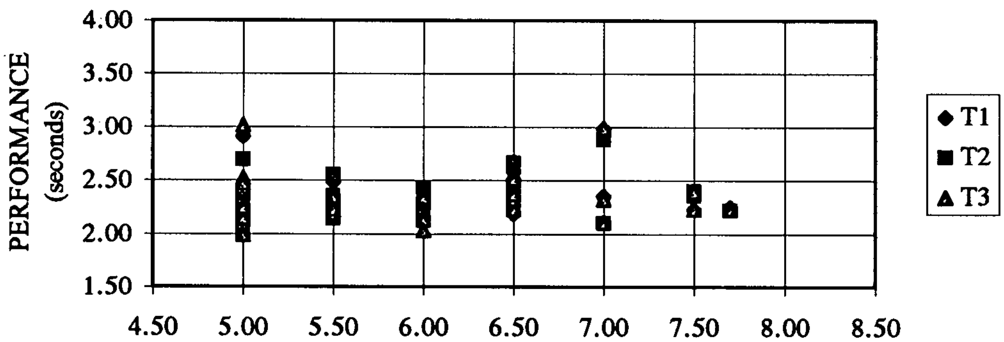


FIGURE 25.0. FIGURE-OF-EIGHT-RUN--CORRELATION BETWEEN KNEE JOINT LAXITY AND INJURED BRACED SUBJECTS

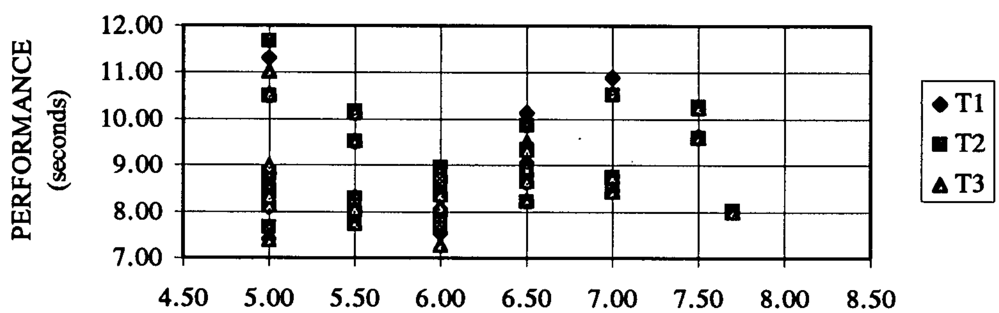


FIGURE 26.0. SLALOM RUN--CORRELATION BETWEEN KNEE JOINT LAXITY AND INJURED BRACED SUBJECTS

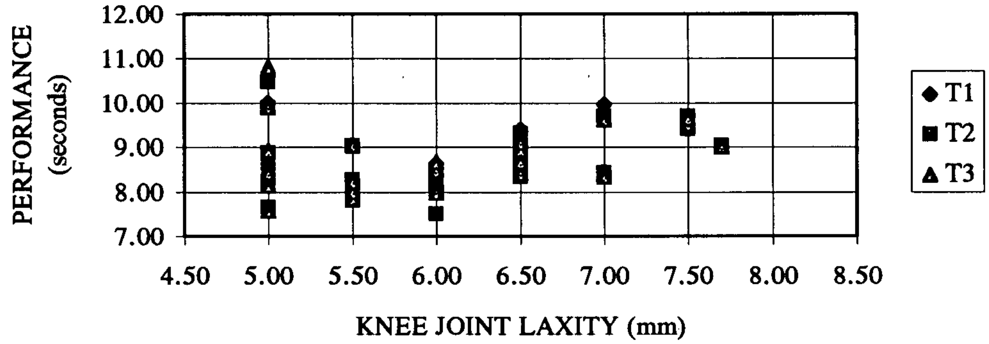


FIGURE 27.0. HOP TEST--CORRELATION BETWEEN KNEE JOINT AND INJURED BRACED SUBJECTS

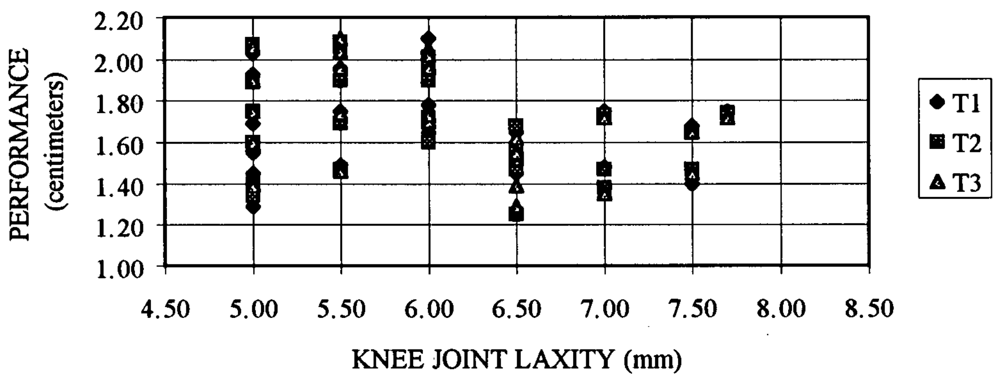
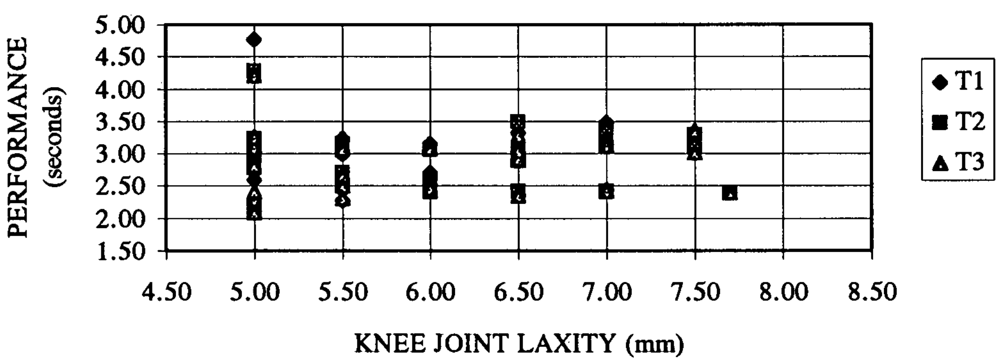


FIGURE 28.0. RUNNING DOWN STAIRS--CORRELATION BETWEEN KNEE JOINT LAXITY AND INJURED BRACED SUBJECTS



For the injured, braced group, the highest positive correlation was noted for the slalom test (r = 0.23) while the lowest was noted for the running down the stairs test (r = 0.02). The hop test recorded the only negative correlation, which also was the highest correlation value (r = -0.25).

For the injured, non-braced group, all correlation values were higher than those recorded for the injured, braced group. The correlation value for the slalom test was again the highest positive correlation (r = 0.31) while the lowest correlation was observed for the figure-of-eight test (r = 0.03). Again, a negative, as well as the highest, correlation value was observed for the hop test (r = -0.31).

Table 18.0. Correlation Between Knee Joint Laxity and Performance For Injured, Non-Braced Subjects

TEST	r Value
10 Meter Run	0.253335
Figure-of-Eight Run	0.032457
Slalom Run	0.310966
Hop Test	-0.316147
Running Down Stairs	0.055447

FIGURE 29.0. 10 METER RUN--CORRELATION BETWEEN KNEE JOINT LAXITY AND INJURED NON-BRACED SUBJECTS

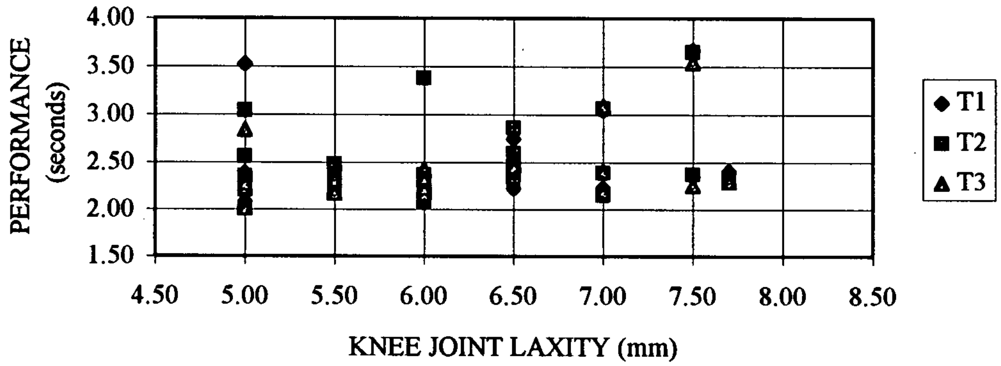


FIGURE 30.0. FIGURE-OF-EIGHT RUN--CORRELATION BETWEEN KNEE JOINT LAXITY AND INJURED NON-BRACED SUBJECTS

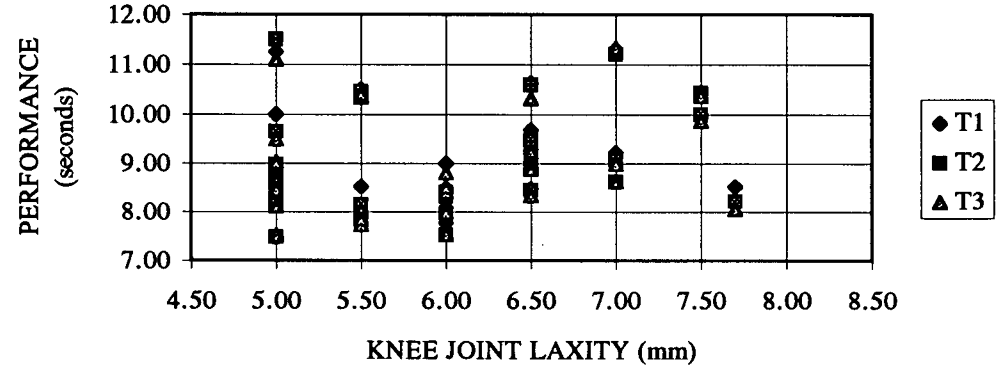


FIGURE 31.0. SLALOM RUN--CORRELATION BETWEEN KNEE JOINT LAXITY AND INJURED NON-BRACED SUBJECTS

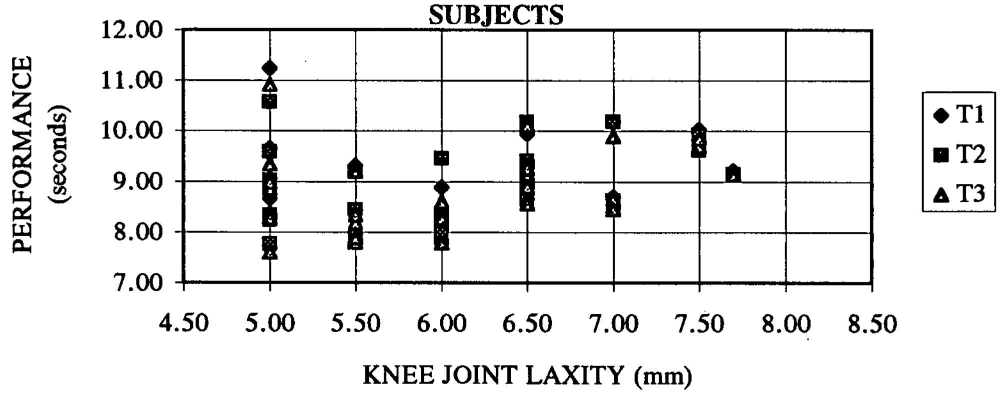


FIGURE 32.0. HOP TEST--CORRELATION BETWEEN KNEE JOINT LAXITY AND INJURED NON-BRACED SUBJECTS

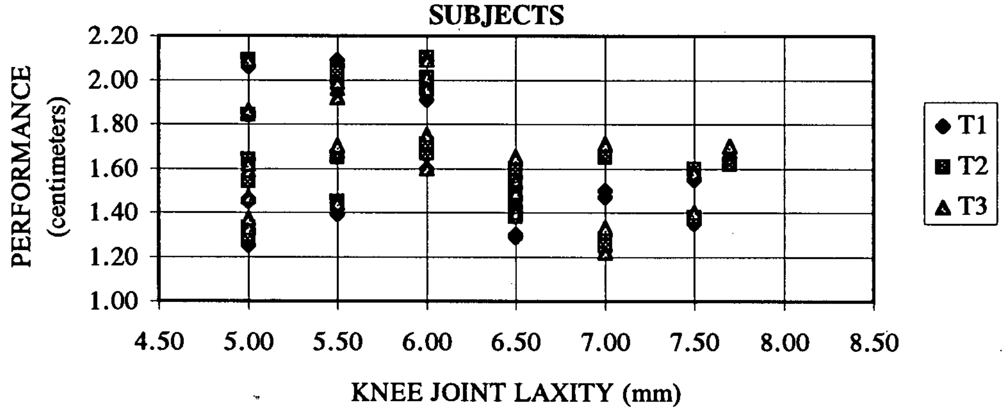
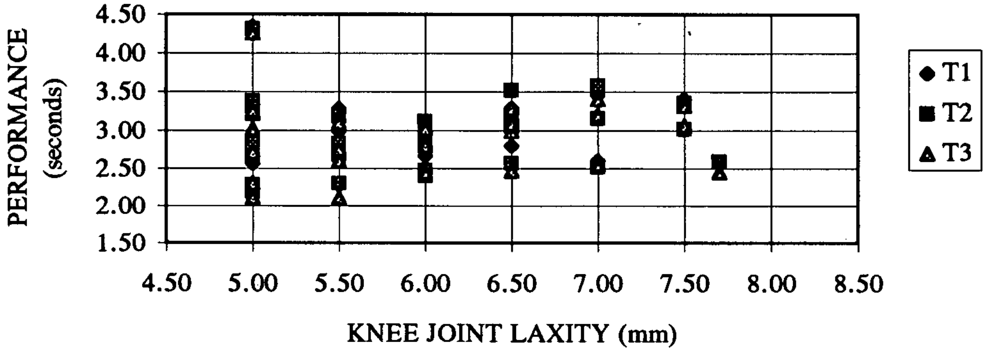


FIGURE 33.0 RUNNING DOWN STAIRS--CORRELATION BETWEEN KNEE JOINT LAXITY AND INJURED NON-BRACED SUBJECTS



DISCUSSION

The primary design and function of a functional knee brace is to provide stability to the unstable (injured) knee joint. To evaluate the effectiveness of this primary role has been the goal of many researchers over the past decade. As a result, the literature available on this topic is extensive and provides both the possible positive and the negative factors associated with functional knee bracing. In 1991, Marans et al., reported that braced ACL-deficient subjects performed better than unbraced, ACL-deficient subjects. Other researchers still contend that functional knee braces hinder performance and may even increase the risk of injury (France et al., 1987; Zetterlund et al., 1986; and Houston & Goenans, 1982). However, most researchers feel that the non-compliance often seen in high-caliber athletes toward the use of functional knee braces is due to the possibility of the brace hindering one's performance. Before athletes comply to the use of functional knee braces, data on these braces must be provided which evaluates this possible hindrance in performance. The goals of this study were to evaluate the effect of functional knee bracing on athletic performance during dynamic testing in a non-injured knee and to measure the effects of functional knee bracing, under dynamic testing, on an ACL-deficient knee.

In comparing the interaction of means within non-injured, braced and non-injured, non-braced subjects (during the accommodation phase), statistically significant differences were noted for the 10 meter dash, the figure-of-eight test, the slalom test, and the hop test. No statistically significant difference was recorded for the running down the stairs test. Of the four statistically significant tests, performance levels were inferior, while utilizing a functional knee brace, in the first three tests while in the hop test, (fourth test) performance level was enhanced by 2.4 centimeters when using the functional knee brace. This enhanced performance could be attributed to proprioception--if a subject feels greater sense of security at the knee joint then one will perform with greater confidence as was illustrated by Vailas et al., (1993), when he used a placebo knee sleeve and found no statistically significant difference in torque between the involved knee and the uninvolved knee. Decrease in performance for the non-injured, braced subjects, for the first three tests, ranged from three one hundredths of a second (10 meter dash) to one tenth of a second difference (figure-of-eight test).

As was expected, statistical significance was noted for all five tests within the injured, braced and the injured, non-braced subjects during the accommodation phase. For all tests, the injured, braced group performed better than the injured, non-braced group. Increase in performance for the injured, braced group ranged from seven one hundredths of a second (running down stairs test) to fourteenth of a second difference (figure-of-eight test).

When non-injured, braced and non-injured, non-braced groups were analyzed with respect to the three trials, three of the five functional tests had statistically significant differences--the 10 meter dash and the hop test showed no statistically significant difference. This indicates that during the accommodation phase, for the three statistically significant tests, variability was present over the respective three trials within the two groups.

This variability could have been caused by two factors, learning effect or fatigue. If the cause was learning effect then one would expect that the majority of the subjects would record their best performance times on the last trial. If the cause was fatigue then the best performance times would result on the first trial after which performance levels would decrease. In this study, for the three significant tests (figure-of-eight test, slalom test, and the running down the stairs test) majority of the non-injured, braced subjects (60%, 50% and 70% respectively) recorded their best performance on the third trial. For the non-injured, non-braced subjects, with respect to the same tests, their best performances were distributed evenly over the three trials.

Injured, braced and injured, non-braced groups were also statistically analyzed over the three trials but no statistically significant differences were found for any of the functional tests during the accommodation phase. This indicates that there was low variability (learning effect or fatigue was not a factor) over the three trials. Percentages for both groups illustrate this point as 74% (37% for each trial) of the best performances were distributed between the first trials or the third trials.

As illustrated above, fatigue was not a factor when performing the five dynamic tests as only the anaerobic cardiovascular system was utilized. However, as illustrated by Highgenboten et al., 1991, the use of a functional knee brace will produce an elevated metabolic cost (increase of 3% to 8% after two weeks of using a functional knee brace). However, with continued use of a functional knee brace (greater than two weeks) one should see the cardiovascular system adapting to meet the demands being placed on this system.

A single factor ANOVA was performed on the best performance times/distance to detect if a statistically significant difference could be ascertained between the non-injured, braced and the non-injured, non-braced groups once these groups had accommodated to the brace. No statistically significant differences were found for any of the five tests. This data suggests that once an individual had accommodated to the functional knee brace non-injured, braced individuals are able to execute these five functional tests at performance levels that are very close to (10 meter dash, figure-of-eight-test, slalom test, and running down the stairs test) or better (hop test) than the non-injured, non-braced subjects. Therefore, contrary to previous studies, functional knee bracing did not statistically impair performance which is an important a consideration when counseling individuals who are contemplating using a functional knee brace for possible prophylactic benefit.

Of note are the best performance results of injured, braced and injured, non-braced subjects. A single factor ANOVA failed to show any statistically significant differences between the two groups for any of the five tests. These results may be attributed to placing less force on the injured knee joint and to knee laxity levels (these two factors are detailed below).

Contrary to my hypothesis and to previous studies (Marans et al., 1991 and Veldhuizen et al., 1991), performance improvements were statistically more apparent in the hop test and the running down the stairs test than the 10 meter dash test. Again, proprioception may have played a factor in this outcome.

A strong correlation was not evident between knee joint laxity and performance times/distance for either the injured, braced or the injured, non-braced groups. This maybe attributed to a large number of injured subjects having knee joint laxity between 5.0 and 6.0 millimeters and thus knee stability may not have been a large factor for these subjects. As a result, these subjects were able to compensate their running style by placing less force on the injured knee joint--similar to the biomechanical finding of Branch and Hunter, 1986. If subjects in this study had had greater knee joint laxity, 7.0 millimeters or greater, (closer to the mean anterior displacement shown by Daniel et al., 1988) a stronger correlation may have been noted because the injured subjects would have had a less stable knee joint. This greater knee joint laxity may have resulted in decreased performance as subjects might not have been able to compensate their running style in the above fashion.

However, a general correlation trend was observed for both groups and over all five tests--the greater the knee joint laxity exhibited, the slower or lower the performance level recorded.

CONCLUSIONS

As was expected, the results of this study did show that non-injured, braced subjects were able to accommodate to a functional knee brace. During the accommodation period, the non-injured, braced athletes showed a slight decrease in performance in three of the five functional tests; one test (running down the stairs test) showed no difference in performance levels, while the hop test showed an enhancement in performance. Once the non-injured, braced subjects had accommodated to a functional knee brace they were able to perform the five functional tests at almost the same level as or better than the non-injured, non-braced subjects. One should take note of two important factors:

1) That the accommodation period for the brace was only ten days--had the accommodation period been longer the performance levels may have been similar to those after the accommodation period as illustrated by Veldhuizen et al., 1991.

2) The braces utilized by the non-injured subjects were "ideal-fitting" braces, not custom fit braces. Again, had the braces been custom fit to each subject, accommodation may have occurred quicker.

As hypothesized, injured athlete's performance was enhanced when a functional knee brace was utilized. However, when only the best performance results were compared between the two groups the performance levels were not significantly different. This latter finding maybe attributed to the subjects's compensating by placing less forces on the injured knee joint as illustrated by Branch and Hunter, 1990.

Contrary to the hypothesis, a strong correlation was not evident between the athlete's knee joint laxity and performance levels for any of the five functional tests. As stated earlier, this may have been attributed to the large number of athletes' having only moderate knee joint laxity. However, a general trend was observed--the greater the knee laxity the lower the performance level.

In the present study, no effects of fatigue were exhibited by any of the participants as only the anaerobic cardiovascular system was utilized while performing the five functional tests. According to available literature there is an increased metabolic cost of using a functional knee brace (increase of 3% to 8%) when the aerobic cardiovascular system is utilized; which could be a factor when using a functional knee brace for a full game setting. However, with long term use of a functional knee brace (greater than 21 days) the cardiovascular system should be able to adapt to meet the demands.

The results of the present study show that a functional knee brace only marginally hinders performance during the accommodation phase when performing the 10 meter, figure-of-eight, and the slalom tests; is not a factor when performing the running down the stairs test; and improves performance during the hop test. However, once accommodation to the brace has occurred, performance levels of non-injured, braced subjects are similar (in the 10 meter, figure-of-eight and the slalom tests) or better (in the hop and running down the stairs tests) than the non-injured, non-braced subjects.

Future research needs to concentrate on longitudinal testing of the functional knee brace in a dynamic setting similar to Jackson's et al. (1991), study involving prophylactic knee braces. This will allow researchers to determine what effect on performance, if any, a functional knee brace can play if it is used for prophylactic benefit.

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APPENDIX A

Performance Times/Distance For Non-Injured, Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
1	10 METER DASH	2.19	2.24	2.11
	FIGURE-OF-EIGHT	8.70	8.57	8.66
	SLALOM RUN	7.21	7.09	7.18
	SINGLE HOP	1.65	1.66	1.72
	RUNNING DOWN STAIRS	2.89	2.99	3.02
2	10 METER DASH	2.19	2.19	2.16
	FIGURE-OF-EIGHT	8.83	8.74	8.52
	SLALOM RUN	6.49	6.92	6.36
	SINGLE HOP	1.73	1.85	1.85
	RUNNING DOWN STAIRS	2.99	3.23	3.15
3	10 METER DASH	2.29	2.15	2.27
	FIGURE-OF-EIGHT	7.68	7.44	7.60
	SLALOM RUN	7.62	7.41	7.49
	SINGLE HOP	1.94	1.74	1.96
	RUNNING DOWN STAIRS	2.68	2.74	2.64
4	10 METER DASH	2.27	2.17	2.22
	FIGURE-OF-EIGHT	7.69	7.74	7.60
	SLALOM RUN	7.67	7.41	7.64
	SINGLE HOP	1.95	1.89	1.95
	RUNNING DOWN STAIRS	2.77	2.57	2.65
5	10 METER DASH	2.35	2.27	2.38
	FIGURE-OF-EIGHT	7.64	7.74	7.67
	SLALOM RUN	8.50	8.32	8.00
	SINGLE HOP	1.88	1.77	1.89
	RUNNING DOWN STAIRS	2.66	2.44	2.43

APPENDIX A CONTINUED

Performance Times/Distance For Non-Injured, Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
6	10 METER DASH	2.24	2.30	2.03
	FIGURE-OF-EIGHT	8.13	7.80	7.53
	SLALOM RUN	7.50	7.61	7.59
	SINGLE HOP	1.62	1.77	1.80
	RUNNING DOWN STAIRS	2.82	2.85	2.80
7	10 METER DASH	2.55	2.45	2.45
	FIGURE-OF-EIGHT	8.06	7.91	7.51
	SLALOM RUN	7.80	7.99	7.59
	SINGLE HOP	2.13	2.08	2.10
	RUNNING DOWN STAIRS	2.57	2.56	2.99
8	10 METER DASH	2.55	2.76	2.28
	FIGURE-OF-EIGHT	8.20	8.15	8.19
	SLALOM RUN	8.38	8.54	8.22
	SINGLE HOP	1.93	1.96	1.97
	RUNNING DOWN STAIRS	3.21	3.25	2.94
9	10 METER DASH	2.65	2.39	2.31
	FIGURE-OF-EIGHT	8.26	8.06	8.08
	SLALOM RUN	8.31	8.02	8.06
	SINGLE HOP	1.82	2.00	2.02
	RUNNING DOWN STAIRS	2.71	2.47	2.74
10	10 METER DASH	2.44	2.34	2.25
	FIGURE-OF-EIGHT	8.41	8.40	8.03
	SLALOM RUN	8.21	8.07	8.13
	SINGLE HOP	1.91	2.00	2.15
	RUNNING DOWN STAIRS	2.63	2.41	2.40

APPENDIX A CONTINUED

Performance Times/Distance For Non-Injured, Braced Subjects (N=30)

SUB NO.	TESTS	TRIALS 1	2	3
11	10 METER DASH	2.13	2.27	2.13
	FIGURE-OF-EIGHT	8.04	7.84	7.73
	SLALOM RUN	7.81	7.84	7.80
	SINGLE HOP	1.89	1.92	2.07
	RUNNING DOWN STAIRS	3.03	2.61	2.35
12	10 METER DASH	2.32	2.28	2.26
	FIGURE-OF-EIGHT	8.23	8.20	8.15
	SLALOM RUN	8.18	8.15	8.12
	SINGLE HOP	1.88	1.92	1.98
	RUNNING DOWN STAIRS	2.99	3.00	2.94
13	10 METER DASH	2.12	1.96	2.17
	FIGURE-OF-EIGHT	8.07	9.61	8.14
	SLALOM RUN	8.07	8.61	8.14
	SINGLE HOP	1.95	1.65	1.70
	RUNNING DOWN STAIRS	3.06	2.84	2.91
14	10 METER DASH	2.22	2.42	2.45
	FIGURE-OF-EIGHT	8.68	8.87	8.34
	SLALOM RUN	8.39	8.22	8.34
	SINGLE HOP	1.76	1.80	1.85
	RUNNING DOWN STAIRS	3.06	3.02	2.91
15	10 METER DASH	2.29	2.27	2.25
	FIGURE-OF-EIGHT	8.25	8.16	8.10
	SLALOM RUN	8.12	8.12	8.06
	SINGLE HOP	1.75	1.77	1.85
	RUNNING DOWN STAIRS	2.40	2.33	2.30

APPENDIX A CONTINUED

Performance Times/Distance For Non-Injured, Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
16	10 METER DASH	2.20	2.18	2.15
	FIGURE-OF-EIGHT	8.12	8.11	8.09
	SLALOM RUN	8.26	8.30	8.24
	SINGLE HOP	1.90	1.90	1.83
	RUNNING DOWN STAIRS	2.45	2.38	2.37
17	10 METER DASH	2.37	2.41	2.41
	FIGURE-OF-EIGHT	9.46	9.24	9.24
	SLALOM RUN	7.00	7.07	6.89
	SINGLE HOP	1.46	1.40	1.57
	RUNNING DOWN STAIRS	3.24	3.20	3.07
18	10 METER DASH	2.49	2.52	2.54
	FIGURE-OF-EIGHT	8.79	8.73	8.52
	SLALOM RUN	8.56	8.56	8.28
	SINGLE HOP	1.44	1.75	1.40
	RUNNING DOWN STAIRS	2.99	3.45	2.48
19	10 METER DASH	2.38	2.45	2.49
	FIGURE-OF-EIGHT	9.74	9.71	9.31
	SLALOM RUN	8.98	8.91	8.73
	SINGLE HOP	1.19	1.35	1.36
	RUNNING DOWN STAIRS	2.81	2.78	2.66
20	10 METER DASH	2.53	2.49	2.48
	FIGURE-OF-EIGHT	8.89	8.84	8.80
	SLALOM RUN	8.35	8.35	8.30
	SINGLE HOP	1.54	1.55	1.56
	RUNNING DOWN STAIRS	2.68	2.63	2.45

APPENDIX A CONTINUED

Performance Times/Distance For Non-Injured, Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
21	10 METER DASH	2.45	2.43	2.55
	FIGURE-OF-EIGHT	8.29	8.09	8.09
	SLALOM RUN	8.274	8.20	8.20
	SINGLE HOP	1.70	1.75	1.90
	RUNNING DOWN STAIRS	3.09	2.99	2.92
22	10 METER DASH	2.44	2.28	2.28
	FIGURE-OF-EIGHT	8.45	8.58	8.30
	SLALOM RUN	8.30	8.20	8.40
	SINGLE HOP	1.66	1.70	1.63
	RUNNING DOWN STAIRS	2.63	2.99	2.59
23	10 METER DASH	2.42	2.59	2.47
	FIGURE-OF-EIGHT	8.66	8.52	8.25
	SLALOM RUN	8.76	8.34	8.36
	SINGLE HOP	1.69	1.76	1.72
	RUNNING DOWN STAIRS	2.86	2.83	2.80
24	10 METER DASH	2.52	2.50	2.54
	FIGURE-OF-EIGHT	9.60	9.65	9.56
	SLALOM RUN	9.68	9.59	9.61
	SINGLE HOP	1.65	1.70	1.71
	RUNNING DOWN STAIRS	3.27	3.02	3.09
25	10 METER DASH	2.47	2.48	2.50
	FIGURE-OF-EIGHT	8.96	9.02	9.03
	SLALOM RUN	8.62	8.31	8.54
	SINGLE HOP	1.53	1.57	1.54
	RUNNING DOWN STAIRS	3.01	3.00	3.03

APPENDIX A CONTINUED

Performance Times/Distance For Non-Injured, Braced Subjects (N=30)

SUB NO.	TESTS	TRIALS 1	2	3
26	10 METER DASH	2.63	2.66	2.51
	FIGURE-OF-EIGHT	9.14	9.01	8.74
	SLALOM RUN	9.07	8.74	8.51
	SINGLE HOP	1.79	1.78	1.84
	RUNNING DOWN STAIRS	2.68	2.64	2.57
27	10 METER DASH	2.32	2.29	2.18
	FIGURE-OF-EIGHT	8.38	8.16	8.20
	SLALOM RUN	8.31	8.54	8.38
	SINGLE HOP	1.62	1.78	1.82
	RUNNING DOWN STAIRS	2.98	3.01	2.95
28	10 METER DASH	2.44	2.34	2.33
	FIGURE-OF-EIGHT	8.45	8.35	8.31
	SLALOM RUN	8.27	8.25	8.21
	SINGLE HOP	1.65	1.69	1.75
	RUNNING DOWN STAIRS	3.21	3.11	3.00
29	10 METER DASH	2.29	2.27	2.25
	FIGURE-OF-EIGHT	8.68	8.81	8.76
	SLALOM RUN	8.67	8.36	8.17
	SINGLE HOP	1.43	1.46	1.38
	RUNNING DOWN STAIRS	2.34	2.45	2.44
30	10 METER DASH	2.30	2.30	2.23
	FIGURE-OF-EIGHT	8.28	8.21	8.22
	SLALOM RUN	8.50	8.42	8.41
	SINGLE HOP	1.58	1.59	1.51
	RUNNING DOWN STAIRS	2.56	2.56	2.52

APPENDIX B

Performance Times/Distance For Non-Injured, Non-Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
1	10 METER DASH	2.24	2.24	2.29
	FIGURE-OF-EIGHT	8.34	8.39	8.54
	SLALOM RUN	7.02	7.14	7.00
	SINGLE HOP	1.84	1.74	1.61
	RUNNING DOWN STAIRS	3.05	2.93	2.93
2	10 METER DASH	2.21	1.94	2.04
	FIGURE-OF-EIGHT	8.54	8.65	8.74
	SLALOM RUN	6.74	7.06	6.71
	SINGLE HOP	1.86	1.74	1.75
	RUNNING DOWN STAIRS	3.22	3.16	3.15
3	10 METER DASH	2.13	2.13	2.13
	FIGURE-OF-EIGHT	7.46	7.36	7.25
	SLALOM RUN	7.35	7.29	7.20
	SINGLE HOP	1.91	1.93	2.00
	RUNNING DOWN STAIRS	2.77	2.65	2.79
4	10 METER DASH	2.16	2.31	2.13
	FIGURE-OF-EIGHT	7.67	7.81	7.59
	SLALOM RUN	7.60	7.59	7.43
	SINGLE HOP	1.90	1.95	2.05
	RUNNING DOWN STAIRS	2.55	2.74	2.99
5	10 METER DASH	2/24	2.37	2.35
	FIGURE-OF-EIGHT	7.59	7.59	7.76
	SLALOM RUN	8.13	8.10	8.10
	SINGLE HOP	1.91	1.90	1.81
	RUNNING DOWN STAIRS	2.47	2.40	2.38

APPENDIX B CONTINUED

Performance Times/Distance For Non-Injured, Non-Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
6	10 METER DASH	2.09	2.28	2.08
	FIGURE-OF-EIGHT	7.42	7.30	7.61
	SLALOM RUN	7.49	7.59	7.57
	SINGLE HOP	1.70	1.79	1.80
	RUNNING DOWN STAIRS	2.77	2.81	2.82
7	10 METER DASH	2.41	2.35	2.34
	FIGURE-OF-EIGHT	7.92	7.52	7.60
	SLALOM RUN	7.58	7.59	7.79
	SINGLE HOP	2.13	2.06	2.08
	RUNNING DOWN STAIRS	2.99	3.04	3.03
8	10 METER DASH	2.30	2.3	2.43
	FIGURE-OF-EIGHT	8.23	8.25	8.10
	SLALOM RUN	7.99	8.10	8.23
	SINGLE HOP	1.93	1.97	1.92
	RUNNING DOWN STAIRS	3.02	2.81	3.13
9	10 METER DASH	2.32	2.21	2.33
	FIGURE-OF-EIGHT	8.10	8.10	8.00
	SLALOM RUN	8.00	7.82	8.09
	SINGLE HOP	1.83	2.00	1.89
	RUNNING DOWN STAIRS	2.62	2.73	2.67
10	10 METER DASH	2.34	2.40	2.25
	FIGURE-OF-EIGHT	7.98	7.98	7.98
	SLALOM RUN	7.91	7.87	7.95
	SINGLE HOP	2.12	2.16	2.03
	RUNNING DOWN STAIRS	2.57	2.67	2.87

APPENDIX B CONTINUED

Performance Times/Distance For Non-Injured, Non-Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
11	10 METER DASH	2.27	2.27	2.35
	FIGURE-OF-EIGHT	8.00	7.85	7.75
	SLALOM RUN	7.93	7.63	7.75
	SINGLE HOP	1.89	2.04	2.02
	RUNNING DOWN STAIRS	2.52	2.76	2.46
12	10 METER DASH	2.30	2.31	2.27
	FIGURE-OF-EIGHT	8.13	8.18	8.20
	SLALOM RUN	8.10	8.15	8.20
	SINGLE HOP	1.99	1.94	1.99
	RUNNING DOWN STAIRS	2.93	2.93	2.94
13	10 METER DASH	2.25	2.14	2.10
	FIGURE-OF-EIGHT	8.00	8.02	7.97
	SLALOM RUN	7.77	7.78	7.88
	SINGLE HOP	1.70	1.70	1.68
	RUNNING DOWN STAIRS	3.09	2.77	2.88
14	10 METER DASH	2.27	2.33	2.27
	FIGURE-OF-EIGHT	8.76	8.68	8.58
	SLALOM RUN	8.92	8.53	8.51
	SINGLE HOP	1.97	1.90	1.90
	RUNNING DOWN STAIRS	2.90	2.94	3.26
15	10 METER DASH	2.24	2.35	2.30
	FIGURE-OF-EIGHT	8.07	8.15	8.13
	SLALOM RUN	8.01	8.10	8.10
	SINGLE HOP	1.84	1.83	1.84
	RUNNING DOWN STAIRS	2.31	2.37	2.35

APPENDIX B CONTINUED

Performance Times/Distance For Non-Injured, Non-Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
16	10 METER DASH	2.16	2.18	2.16
	FIGURE-OF-EIGHT	8.02	8.11	8.10
	SLALOM RUN	8.21	8.31	8.30
	SINGLE HOP	1.85	1.89	1.87
	RUNNING DOWN STAIRS	2.40	2.35	2.45
17	10 METER DASH	2.38	2.25	2.44
	FIGURE-OF-EIGHT	9.13	9.16	9.20
	SLALOM RUN	6.89	7.20	6.85
	SINGLE HOP	1.47	1.46	1.47
	RUNNING DOWN STAIRS	3.22	3.16	3.07
18	10 METER DASH	2.44	2.31	2.27
	FIGURE-OF-EIGHT	8.64	8.41	8.24
	SLALOM RUN	8.23	8.55	8.09
	SINGLE HOP	1.67	1.57	1.56
	RUNNING DOWN STAIRS	2.30	2.36	2.36
19	10 METER DASH	2.30	2.40	2.56
	FIGURE-OF-EIGHT	9.34	9.17	9.10
	SLALOM RUN	8.67	8.61	8.97
	SINGLE HOP	1.36	1.28	1.37
	RUNNING DOWN STAIRS	2.77	2.66	2.62
20	10 METER DASH	2.61	2.63	2.45
	FIGURE-OF-EIGHT	8.76	8.65	8.83
	SLALOM RUN	8.25	8.23	8.23
	SINGLE HOP	1.54	1.48	1.50
	RUNNING DOWN STAIRS	2.69	2.77	2.55

APPENDIX B CONTINUED

Performance Times/Distance For Non-Injured, Non-Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
21	10 METER DASH	2.42	2.42	2.52
	FIGURE-OF-EIGHT	8.06	8.02	8.00
	SLALOM RUN	8.00	8.00	8.15
	SINGLE HOP	1.82	1.85	1.79
	RUNNING DOWN STAIRS	3.03	2.90	2.95
22	10 METER DASH	2.31	2.25	2.37
	FIGURE-OF-EIGHT	8.48	8.57	8.59
	SLALOM RUN	8.24	8.19	8.15
	SINGLE HOP	1.77	1.69	1.75
	RUNNING DOWN STAIRS	2.53	2.99	2.76
23	10 METER DASH	2.52	2.40	2.38
	FIGURE-OF-EIGHT	8.60	8.50	8.15
	SLALOM RUN	8.19	8.21	8.13
	SINGLE HOP	1.67	1.66	1.68
	RUNNING DOWN STAIRS	2.95	2.74	2.82
24	10 METER DASH	2.52	2.42	2.52
	FIGURE-OF-EIGHT	9.36	9.33	9.32
	SLALOM RUN	9.41	9.36	9.36
	SINGLE HOP	1.73	1.73	1.75
	RUNNING DOWN STAIRS	3.13	3.23	3.30
25	10 METER DASH	2.41	2.41	2.41
	FIGURE-OF-EIGHT	9.34	9.27	9.29
	SLALOM RUN	8.25	7.77	8.11
	SINGLE HOP	1.57	1.60	2.58
	RUNNING DOWN STAIRS	3.29	3.17	3.20

APPENDIX B CONTINUED

Performance Times/Distance For Non-Injured, Non-Braced Subjects (N=30)

SUB NO.	TESTS	TRIALS 1	2	3
26	10 METER DASH	2.53	2.58	2.47
	FIGURE-OF-EIGHT	8.91	9.09	8.95
	SLALOM RUN	9.13	8.74	9.04
	SINGLE HOP	1.80	1.83	1.82
	RUNNING DOWN STAIRS	2.57	2.56	2.52
27	10 METER DASH	2.15	2.37	2.15
	FIGURE-OF-EIGHT	8.15	8.12	8.34
	SLALOM RUN	8.31	8.25	8.34
	SINGLE HOP	1.86	1.78	1.79
	RUNNING DOWN STAIRS	3.01	2.94	3.02
28	10 METER DASH	2.32	2.32	2.34
	FIGURE-OF-EIGHT	8.29	8.30	8.28
	SLALOM RUN	8.22	8.25	9.26
	SINGLE HOP	1.71	1.73	1.73
	RUNNING DOWN STAIRS	3.05	3.07	3.09
29	10 METER DASH	2.19	2.16	2.17
	FIGURE-OF-EIGHT	8.41	8.58	8.48
	SLALOM RUN	8.60	8.38	8.45
	SINGLE HOP	1.34	1.48	1.57
	RUNNING DOWN STAIRS	2.53	2.30	3.24
30	10 METER DASH	2.21	2.20	2.22
	FIGURE-OF-EIGHT	8.17	8.20	8.18
	SLALOM RUN	8.39	8.40	8.46
	SINGLE HOP	1.58	1.55	1.55
	RUNNING DOWN STAIRS	3.55	2.65	2.56

APPENDIX C

Performance Times/Distance For Injured, Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
31	10 METER DASH	2.15	2.13	2.39
	FIGURE-OF-EIGHT	8.06	7.88	8.03
	SLALOM RUN	8.41	8.33	8.08
	SINGLE HOP	1.68	1.60	1.70
	RUNNING DOWN STAIRS	2.60	2.48	2.48
32	10 METER DASH	2.16	2.35	2.33
	FIGURE-OF-EIGHT	8.00	7.79	7.75
	SLALOM RUN	7.84	7.92	7.85
	SINGLE HOP	1.95	2.06	2.10
	RUNNING DOWN STAIRS	2.99	2.67	2.66
33	10 METER DASH	2.41	2.14	2.13
	FIGURE-OF-EIGHT	7.54	7.65	7.28
	SLALOM RUN	8.16	7.52	8.15
	SINGLE HOP	2.10	2.01	2.05
	RUNNING DOWN STAIRS	2.51	2.40	2.47
34	10 METER DASH	2.48	2.55	2.39
	FIGURE-OF-EIGHT	8.31	8.29	8.07
	SLALOM RUN	7.98	8.28	8.09
	SINGLE HOP	1.90	1.90	1.96
	RUNNING DOWN STAIRS	2.43	2.70	2.49
35	10 METER DASH	2.15	2.14	2.19
	FIGURE-OF-EIGHT	8.06	7.99	7.74
	SLALOM RUN	8.24	7.81	8.01
	SINGLE HOP	1.96	2.08	2.04
	RUNNING DOWN STAIRS	2.28	2.49	2.31

APPENDIX C CONTINUED

Performance Times/Distance For Injured, Braced Subjects (N=30)

SUB NO.	TESTS	TRIALS 1	2	3
36	10 METER DASH	2.30	2.43	2.31
	FIGURE-OF-EIGHT	8.39	8.36	8.13
	SLALOM RUN	8.42	8.16	8.48
	SINGLE HOP	1.95	1.90	2.02
	RUNNING DOWN STAIRS	3.15	3.06	3.07
37	10 METER DASH	2.14	2.02	1.99
	FIGURE-OF-EIGHT	7.42	7.66	7.39
	SLALOM RUN	7.68	7.68	7.59
	SINGLE HOP	2.03	2.04	2.06
	RUNNING DOWN STAIRS	2.29	2.21	2.30
38	10 METER DASH	2.30	2.02	1.99
	FIGURE-OF-EIGHT	7.42	7.66	7.39
	SLALOM RUN	8.63	8.44	8.66
	SINGLE HOP	1.78	1.96	1.96
	RUNNING DOWN STAIRS	2.70	2.51	2.41
39	10 METER DASH	2.38	2.37	2.52
	FIGURE-OF-EIGHT	8.78	8.81	8.98
	SLALOM RUN	8.63	8.85	8.54
	SINGLE HOP	1.45	1.40	1.40
	RUNNING DOWN STAIRS	2.59	2.77	2.41
40	10 METER DASH	2.91	2.69	3.01
	FIGURE-OF-EIGHT	11.31	11.67	11.02
	SLALOM RUN	10.63	10.46	10.80
	SINGLE HOP	1.29	1.34	1.39
	RUNNING DOWN STAIRS	4.77	4.27	4.20

APPENDIX C CONTINUED

Performance Times/Distance For Injured, Braced Subjects (N=30)

SUB NO.	TESTS	TRIALS 1	2	3
41	10 METER DASH	2.34	2.27	2.47
	FIGURE-OF-EIGHT	8.09	8.12	8.13
	SLALOM RUN	8.54	8.20	8.40
	SINGLE HOP	1.93	1.89	1.90
	RUNNING DOWN STAIRS	2.90	2.81	2.81
42	10 METER DASH	2.58	2.66	2.56
	FIGURE-OF-EIGHT	9.01	8.89	8.92
	SLALOM RUN	8.78	8.75	8.74
	SINGLE HOP	1.55	1.25	1.29
	RUNNING DOWN STAIRS	2.35	2.41	2.35
43	10 METER DASH	2.11	2.10	2.11
	FIGURE-OF-EIGHT	8.76	8.74	8.75
	SLALOM RUN	8.42	8.43	8.45
	SINGLE HOP	1.75	1.73	1.72
	RUNNING DOWN STAIRS	2.43	2.41	2.41
44	10 METER DASH	2.20	2.22	2.20
	FIGURE-OF-EIGHT	8.53	8.49	8.50
	SLALOM RUN	8.90	8.87	8.92
	SINGLE HOP	1.55	1.60	1.59
	RUNNING DOWN STAIRS	2.12	2.08	2.09
45	10 METER DASH	2.25	2.22	2.23
	FIGURE-OF-EIGHT	7.99	8.02	8.00
	SLALOM RUN	9.00	9.03	9.01
	SINGLE HOP	1.75	1.74	1.72
	RUNNING DOWN STAIRS	2.39	2.39	2.39

APPENDIX C CONTINUED

Performance Times/Distance For Injured, Braced Subjects (N=30)

SUB NO.	TESTS	TRIALS 1	2	3
46	10 METER DASH	2.16	2.25	2.23
	FIGURE-OF-EIGHT	8.24	8.21	8.26
	SLALOM RUN	8.36	8.34	8.40
	SINGLE HOP	1.65	1.68	1.62
	RUNNING DOWN STAIRS	2.93	2.89	2.89
47	10 METER DASH	2.15	2.16	2.15
	FIGURE-OF-EIGHT	8.33	8.33	8.33
	SLALOM RUN	8.24	8.20	8.14
	SINGLE HOP	1.69	1.75	1.75
	RUNNING DOWN STAIRS	3.05	3.02	3.09
48	10 METER DASH	2.66	2.39	2.51
	FIGURE-OF-EIGHT	9.06	8.91	8.95
	SLALOM RUN	9.33	9.27	8.94
	SINGLE HOP	1.50	1.50	1.56
	RUNNING DOWN STAIRS	3.16	3.01	3.13
49	10 METER DASH	2.98	2.88	2.99
	FIGURE-OF-EIGHT	10.89	10.52	10.56
	SLALOM RUN	9.97	9.70	9.61
	SINGLE HOP	1.48	1.47	1.36
	RUNNING DOWN STAIRS	3.48	3.35	3.20
50	10 METER DASH	2.33	2.20	2.34
	FIGURE-OF-EIGHT	8.72	8.67	8.37
	SLALOM RUN	8.09	8.17	7.99
	SINGLE HOP	1.65	1.72	1.73
	RUNNING DOWN STAIRS	2.66	2.47	2.45

APPENDIX C CONTINUED

Performance Times/Distance For Injured, Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
51	10 METER DASH	2.67	2.45	2.40
	FIGURE-OF-EIGHT	10.14	9.86	9.50
	SLALOM RUN	9.39	9.31	9.27
	SINGLE HOP	1.45	1.47	1.39
	RUNNING DOWN STAIRS	3.32	3.49	3.28
52	10 METER DASH	2.31	2.30	2.34
	FIGURE-OF-EIGHT	10.12	10.16	10.14
	SLALOM RUN	8.20	8.18	8.23
	SINGLE HOP	1.75	1.69	1.74
	RUNNING DOWN STAIRS	3.23	3.15	3.11
53	10 METER DASH	2.39	2.40	2.39
	FIGURE-OF-EIGHT	10.25	10.26	10.24
	SLALOM RUN	9.41	9.41	9.49
	SINGLE HOP	1.68	1.65	1.65
	RUNNING DOWN STAIRS	3.33	3.29	3.35
54	10 METER DASH	2.23	2.36	2.23
	FIGURE-OF-EIGHT	9.63	9.59	9.59
	SLALOM RUN	9.70	9.70	9.63
	SINGLE HOP	1.40	1.47	1.45
	RUNNING DOWN STAIRS	3.10	3.09	3.01
55	10 METER DASH	2.21	2.22	2.23
	FIGURE-OF-EIGHT	9.50	9.52	9.53
	SLALOM RUN	9.05	9.03	9.04
	SINGLE HOP	1.49	1.47	1.46
	RUNNING DOWN STAIRS	3.00	3.05	3.07

APPENDIX C CONTINUED

Performance Times/Distance For Injured, Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
56	10 METER DASH	2.25	2.27	2.25
	FIGURE-OF-EIGHT	10.49	10.49	10.53
	SLALOM RUN	10.01	9.89	9.89
	SINGLE HOP	1.60	1.60	1.60
	RUNNING DOWN STAIRS	3.23	3.23	3.25
57	10 METER DASH	2.35	2.33	2.33
	FIGURE-OF-EIGHT	9.36	9.32	9.35
	SLALOM RUN	9.06	9.10	9.12
	SINGLE HOP	1.55	1.54	1.52
	RUNNING DOWN STAIRS	2.91	2.89	2.92
58	10 METER DASH	2.35	2.33	2.33
	FIGURE-OF-EIGHT	9.32	9.33	9.34
	SLALOM RUN	9.10	9.06	9.09
	SINGLE HOP	1.53	1.55	1.54
	RUNNING DOWN STAIRS	2.89	2.89	2.91
59	10 METER DASH	2.39	2.37	2.37
	FIGURE-OF-EIGHT	8.63	8.65	8.66
	SLALOM RUN	8.69	8.65	8.66
	SINGLE HOP	1.53	1.57	1.55
	RUNNING DOWN STAIRS	3.00	3.02	3.04
60	10 METER DASH	2.35	2.34	2.32
	FIGURE-OF-EIGHT	8.48	8.45	8.43
	SLALOM RUN	8.40	8.36	8.34
	SINGLE HOP	1.38	1.38	1.35
	RUNNING DOWN STAIRS	3.16	3.12	3.12

APPENDIX D

Performance Times/Distance For Injured, Non-Braced Subjects (N=30)

SUB NO.	TESTS	TRIALS 1	2	3
31	10 METER DASH	2.22	2.26	2.42
	FIGURE-OF-EIGHT	8.09	7.96	7.99
	SLALOM RUN	8.40	8.24	8.24
	SINGLE HOP	1.60	1.67	1.60
	RUNNING DOWN STAIRS	2.45	2.45	2.42
32	10 METER DASH	2.24	2.31	2.17
	FIGURE-OF-EIGHT	7.95	7.95	7.73
	SLALOM RUN	7.87	7.81	7.80
	SINGLE HOP	2.06	1.96	1.96
	RUNNING DOWN STAIRS	2.99	2.67	2.75
33	10 METER DASH	2.19	3.38	2.16
	FIGURE-OF-EIGHT	7.77	7.53	7.53
	SLALOM RUN	7.78	7.88	7.78
	SINGLE HOP	2.09	2.10	2.09
	RUNNING DOWN STAIRS	2.46	2.40	2.42
34	10 METER DASH	2.41	2.48	2.38
	FIGURE-OF-EIGHT	8.53	8.16	8.03
	SLALOM RUN	7.97	7.94	8.19
	SINGLE HOP	2.09	2.01	1.99
	RUNNING DOWN STAIRS	2.62	2.82	2.60
35	10 METER DASH	2.33	2.16	2.17
	FIGURE-OF-EIGHT	7.75	7.82	7.85
	SLALOM RUN	7.84	7.91	7.88
	SINGLE HOP	2.07	2.03	1.92
	RUNNING DOWN STAIRS	2.28	2.30	2.11

APPENDIX D CONTINUED

Performance Times/Distance For Injured,NON-Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
36	10 METER DASH	2.33	2.36	2.32
	FIGURE-OF-EIGHT	8.45	8.31	8.42
	SLALOM RUN	8.47	8.36	8.35
	SINGLE HOP	1.97	2.01	2.01
	RUNNING DOWN STAIRS	2.98	3.11	3.02
37	10 METER DASH	2.08	2.01	2.01
	FIGURE-OF-EIGHT	7.48	7.47	7.51
	SLALOM RUN	7.69	7.76	7.60
	SINGLE HOP	2.06	2.09	2.09
	RUNNING DOWN STAIRS	2.56	2.28	2.31
38	10 METER DASH	2.13	2.01	2.01
	FIGURE-OF-EIGHT	7.48	7.47	7.51
	SLALOM RUN	7.69	7.76	7.60
	SINGLE HOP	2.06	2.09	2.09
	RUNNING DOWN STAIRS	2.56	2.28	2.31
39	10 METER DASH	2.34	2.56	2.34
	FIGURE-OF-EIGHT	8.91	8.98	9.03
	SLALOM RUN	8.66	8.89	8.86
	SINGLE HOP	1.46	1.54	1.47
	RUNNING DOWN STAIRS	2.66	2.65	2.71
40	10 METER DASH	3.53	3.04	2.84
	FIGURE-OF-EIGHT	11.25	11.50	11.10
	SLALOM RUN	11.24	10.56	10.92
	SINGLE HOP	1.32	1.33	1.33
	RUNNING DOWN STAIRS	4.34	4.31	4.26

APPENDIX D CONTINUED

Performance Times/Distance For Injured, Non-Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
41	10 METER DASH	2.35	2.34	2.27
	FIGURE-OF-EIGHT	8.17	8.16	8.12
	SLALOM RUN	8.32	8.24	8.37
	SINGLE HOP	1.85	1.84	1.86
	RUNNING DOWN STAIRS	2.84	2.82	3.02
42	10 METER DASH	2.61	2.59	2.56
	FIGURE-OF-EIGHT	9.29	9.47	8.94
	SLALOM RUN	8.87	9.00	8.86
	SINGLE HOP	1.39	1.38	1.41
	RUNNING DOWN STAIRS	2.80	2.56	2.46
43	10 METER DASH	2.23	2.15	2.19
	FIGURE-OF-EIGHT	9.23	9.07	8.99
	SLALOM RUN	8.70	8.62	8.65
	SINGLE HOP	1.50	1.65	1.71
	RUNNING DOWN STAIRS	2.60	2.51	2.55
44	10 METER DASH	2.31	2.26	2.21
	FIGURE-OF-EIGHT	8.80	8.72	8.64
	SLALOM RUN	9.06	9.02	8.98
	SINGLE HOP	1.45	1.54	1.59
	RUNNING DOWN STAIRS	2.23	2.18	2.11
45	10 METER DASH	2.40	2.33	2.29
	FIGURE-OF-EIGHT	8.53	8.22	8.06
	SLALOM RUN	9.22	9.15	9.13
	SINGLE HOP	1.65	1.62	1.70
	RUNNING DOWN STAIRS	2.59	2.59	2.45

APPENDIX D CONTINUED

Performance Times/Distance For Injured, Non-Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
46	10 METER DASH	2.22	2.32	2.39
	FIGURE-OF-EIGHT	8.49	8.45	8.34
	SLALOM RUN	8.66	8.65	8.56
	SINGLE HOP	1.30	1.55	1.65
	RUNNING DOWN STAIRS	3.15	3.05	2.99
47	10 METER DASH	2.21	2.20	2.23
	FIGURE-OF-EIGHT	8.56	8.50	8.46
	SLALOM RUN	8.35	8.33	8.27
	SINGLE HOP	1.64	1.64	1.62
	RUNNING DOWN STAIRS	3.35	3.30	3.22
48	10 METER DASH	2.46	2.51	2.48
	FIGURE-OF-EIGHT	9.68	9.33	9.22
	SLALOM RUN	9.31	9.41	9.16
	SINGLE HOP	1.56	1.60	1.54
	RUNNING DOWN STAIRS	3.05	3.20	3.20
49	10 METER DASH	3.05	3.05	3.09
	FIGURE-OF-EIGHT	11.30	11.20	11.33
	SLALOM RUN	10.18	10.18	9.89
	SINGLE HOP	1.47	1.28	1.22
	RUNNING DOWN STAIRS	3.46	3.58	3.40
50	10 METER DASH	2.33	2.35	2.33
	FIGURE-OF-EIGHT	8.50	8.41	8.52
	SLALOM RUN	8.14	7.99	8.27
	SINGLE HOP	1.66	1.71	1.75
	RUNNING DOWN STAIRS	2.66	2.47	2.45

APPENDIX D CONTINUED

Performance Times/Distance For Injured,Non-Braced Subjects (N=30)

		TRIALS		
SUB NO.	TESTS	1	2	3
51	10 METER DASH	2.75	2.86	2.41
	FIGURE-OF-EIGHT	10.62	10.59	10.32
	SLALOM RUN	9.94	10.17	10.12
	SINGLE HOP	1.47	1.40	1.45
	RUNNING DOWN STAIRS	3.29	3.52	3.29
52	10 METER DASH	2.45	2.43	2.39
	FIGURE-OF-EIGHT	10.50	10.45	10.34
	SLALOM RUN	8.43	8.44	8.33
	SINGLE HOP	1.65	1.65	1.70
	RUNNING DOWN STAIRS	3.28	3.18	3.19
53	10 METER DASH	3.67	3.65	3.54
	FIGURE-OF-EIGHT	10.45	10.42	10.37
	SLALOM RUN	9.80	9.62	9.70
	SINGLE HOP	1.55	1.60	1.58
	RUNNING DOWN STAIRS	3.40	3.35	3.32
54	10 METER DASH	2.35	2.37	2.25
	FIGURE-OF-EIGHT	9.98	9.99	9.87
	SLALOM RUN	10.03	9.93	9.86
	SINGLE HOP	1.35	1.38	1.40
	RUNNING DOWN STAIRS	3.01	3.01	3.06
55	10 METER DASH	2.45	2.30	2.31
	FIGURE-OF-EIGHT	10.45	10.33	10.40
	SLALOM RUN	9.32	9.18	9.20
	SINGLE HOP	1.39	1.45	1.44
	RUNNING DOWN STAIRS	3.26	3.19	3.10

APPENDIX D CONTINUED

Performance Times/Distance For Injured, Non-Braced Subjects (N=30)

SUB NO.	TESTS	TRIALS 1	2	3
56	10 METER DASH	2.39	2.35	2.27
	FIGURE-OF-EIGHT	10.00	9.65	9.50
	SLALOM RUN	9.65	9.58	9.34
	SINGLE HOP	1.25	1.29	1.37
	RUNNING DOWN STAIRS	3.33	3.38	3.26
57	10 METER DASH	2.45	2.40	2.43
	FIGURE-OF-EIGHT	9.20	9.10	9.01
	SLALOM RUN	9.39	9.31	9.24
	SINGLE HOP	1.29	1.49	1.43
	RUNNING DOWN STAIRS	3.25	3.10	3.05
58	10 METER DASH	2.50	2.40	2.43
	FIGURE-OF-EIGHT	9.20	9.05	9.01
	SLALOM RUN	9.35	9.24	9.27
	SINGLE HOP	1.30	1.45	1.49
	RUNNING DOWN STAIRS	3.09	3.09	3.05
59	10 METER DASH	2.59	2.50	2.45
	FIGURE-OF-EIGHT	8.94	8.90	8.89
	SLALOM RUN	8.95	8.88	8.90
	SINGLE HOP	1.39	1.47	1.40
	RUNNING DOWN STAIRS	3.15	3.12	3.09
60	10 METER DASH	2.39	2.38	2.40
	FIGURE-OF-EIGHT	8.65	8.63	8.63
	SLALOM RUN	8.50	8.50	8.45
	SINGLE HOP	1.23	1.25	1.33
	RUNNING DOWN STAIRS	3.19	3.15	3.20

APPENDIX E

Best Performance Times/Distance Non-Injured, Braced and Non-Injured, Non-Braced Subjects (N=30)

		CONDITION	
SUB NO.	TESTS	BRACED	UN-BRACED
1	10 METER DASH	2.11	2.24
	FIGURE-OF-EIGHT	8.57	8.34
	SLALOM RUN	7.09	7.00
	SINGLE HOP	1.72	1.84
	RUNNING DOWN STAIRS	2.89	2.93
2	10 METER DASH	2.16	1.99
	FIGURE-OF-EIGHT	8.52	8.54
	SLALOM RUN	6.36	6.71
	SINGLE HOP	1.85	1.86
	RUNNING DOWN STAIRS	2.99	3.15
3	10 METER DASH	2.15	2.13
	FIGURE-OF-EIGHT	7.44	7.25
	SLALOM RUN	7.41	7.20
	SINGLE HOP	1.96	2.00
	RUNNING DOWN STAIRS	2.64	2.65
4	10 METER DASH	2.17	2.13
	FIGURE-OF-EIGHT	7.60	2.59
	SLALOM RUN	7.41	7.43
	SINGLE HOP	1.95	2.05
	RUNNING DOWN STAIRS	2.57	2.55
5	10 METER DASH	2.27	2.24
	FIGURE-OF-EIGHT	7.64	7.59
	SLALOM RUN	8.00	8.10
	SINGLE HOP	1.89	1.91
	RUNNING DOWN STAIRS	2.43	2.38

APPENDIX E CONTINUED

Best Performance Times/Distance Non-Injured, Braced and Non-Injured, Non-Braced Subjects (N=30)

SUB NO.	TESTS	CONDITIONS BRACED	UN-BRACED
6	10 METER DASH	2.03	2.08
	FIGURE-OF-EIGHT	7.53	7.30
	SLALOM RUN	7.50	7.49
	SINGLE HOP	1.80	1.80
	RUNNING DOWN STAIRS	2.80	2.77
7	10 METER DASH	2.45	2.34
	FIGURE-OF-EIGHT	7.51	7.52
	SLALOM RUN	7.59	7.58
	SINGLE HOP	2.13	2.13
	RUNNING DOWN STAIRS	2.56	3.03
8	10 METER DASH	2.28	2.30
	FIGURE-OF-EIGHT	8.15	8.10
	SLALOM RUN	8.22	7.99
	SINGLE HOP	1.97	1.97
	RUNNING DOWN STAIRS	2.94	2.81
9	10 METER DASH	2.31	2.21
	FIGURE-OF-EIGHT	8.06	8.00
	SLALOM RUN	8.02	7.82
	SINGLE HOP	2.02	2.00
	RUNNING DOWN STAIRS	2.47	2.62
10	10 METER DASH	2.25	2.25
	FIGURE-OF-EIGHT	8.03	7.98
	SLALOM RUN	8.07	7.87
	SINGLE HOP	2.15	2.16
	RUNNING DOWN STAIRS	2.40	2.57

APPENDIX E CONTINUED

Best Performance Times/Distance Non-Injured, Braced and Non-Injured, Non-Braced Subjects (N=30)

SUB NO.	TESTS	CONDITIONS BRACED	UN-BRACED
11	10 METER DASH	2.13	2.27
	FIGURE-OF-EIGHT	7.73	7.75
	SLALOM RUN	7.80	7.63
	SINGLE HOP	2.07	2.04
	RUNNING DOWN STAIRS	2.35	2.46
12	10 METER DASH	2.26	2.27
	FIGURE-OF-EIGHT	8.15	8.13
	SLALOM RUN	8.12	8.10
	SINGLE HOP	1.98	1.99
	RUNNING DOWN STAIRS	2.94	2.93
13	10 METER DASH	1.96	2.10
	FIGURE-OF-EIGHT	8.07	7.97
	SLALOM RUN	8.07	7.77
	SINGLE HOP	1.95	1.70
	RUNNING DOWN STAIRS	2.84	2.77
14	10 METER DASH	2.22	2.27
	FIGURE-OF-EIGHT	8.34	8.53
	SLALOM RUN	8.22	8.51
	SINGLE HOP	1.85	1.97
	RUNNING DOWN STAIRS	2.91	2.90
15	10 METER DASH	2.25	2.24
	FIGURE-OF-EIGHT	8.10	8.07
	SLALOM RUN	8.06	8.01
	SINGLE HOP	1.85	1.84
	RUNNING DOWN STAIRS	2.30	2.31

APPENDIX E CONTINUED

Best Performance Times/Distance Non-Injured, Braced and Non-Injured, Non-Braced Subjects (N=30)

		CONDITIONS	
SUB NO.	TESTS	BRACED	UN-BRACED
16	10 METER DASH	2.15	2.16
	FIGURE-OF-EIGHT	8.09	8.02
	SLALOM RUN	8.24	8.21
	SINGLE HOP	1.90	1.89
	RUNNING DOWN STAIRS	2.37	2.35
17	10 METER DASH	2.37	2.25
	FIGURE-OF-EIGHT	9.24	9.13
	SLALOM RUN	6.89	6.85
	SINGLE HOP	1.57	1.47
	RUNNING DOWN STAIRS	3.07	3.07
18	10 METER DASH	2.49	2.27
	FIGURE-OF-EIGHT	8.52	8.24
	SLALOM RUN	8.28	8.09
	SINGLE HOP	1.75	1.67
	RUNNING DOWN STAIRS	2.48	2.30
19	10 METER DASH	2.38	2.30
	FIGURE-OF-EIGHT	9.31	9.10
	SLALOM RUN	8.73	8.61
	SINGLE HOP	1.36	1.37
	RUNNING DOWN STAIRS	2.66	2,62
20	10 METER DASH	2.48	2.45
	FIGURE-OF-EIGHT	8.80	8.65
	SLALOM RUN	8.30	8.23
	SINGLE HOP	1.56	1.54
	RUNNING DOWN STAIRS	2.45	2.55

APPENDIX E CONTINUED

Best Performance Times/Distance Non-Injured, Braced and Non-Injured, Non-Braced Subjects (N=30)

		CONDITIONS	
SUB NO.	TESTS	BRACED	UN-BRACED
21	10 METER DASH	2.43	2.42
	FIGURE-OF-EIGHT	8.09	8.00
	SLALOM RUN	8.20	8.00
	SINGLE HOP	1.90	1.85
	RUNNING DOWN STAIRS	2.92	2.90
22	10 METER DASH	2.28	2.25
	FIGURE-OF-EIGHT	8.30	8.48
	SLALOM RUN	8.20	8.15
	SINGLE HOP	1.70	1.77
	RUNNING DOWN STAIRS	2.59	2.53
23	10 METER DASH	2.42	2.38
	FIGURE-OF-EIGHT	8.25	8.15
	SLALOM RUN	8.34	8.13
	SINGLE HOP	1.76	1.68
	RUNNING DOWN STAIRS	2.80	2.74
24	10 METER DASH	2.50	2.42
	FIGURE-OF-EIGHT	9.56	9.32
	SLALOM RUN	9.59	9.36
	SINGLE HOP	1.71	1.75
	RUNNING DOWN STAIRS	3.02	3.13
25	10 METER DASH	2.47	2.41
	FIGURE-OF-EIGHT	9.02	9.27
	SLALOM RUN	8.31	7.77
	SINGLE HOP	1.57	1.60
	RUNNING DOWN STAIRS	3.00	3.17

APPENDIX E CONTINUED

Best Performance Times/Distance Non-Injured, Braced and Non-Injured, Non-Braced Subjects (N=30)

		CONDITIONS	
SUB NO.	TESTS	BRACED	UN-BRACED
26	10 METER DASH	2.51	2.47
	FIGURE-OF-EIGHT	8.74	8.91
	SLALOM RUN	8.51	8.74
	SINGLE HOP	1.84	1.83
	RUNNING DOWN STAIRS	2.57	2.52
27	10 METER DASH	2.18	2.15
	FIGURE-OF-EIGHT	8.16	8.12
	SLALOM RUN	8.31	8.25
	SINGLE HOP	1.82	1.86
	RUNNING DOWN STAIRS	2.95	2.94
28	10 METER DASH	2.33	2.32
	FIGURE-OF-EIGHT	8.31	8.28
	SLALOM RUN	8.21	8.22
	SINGLE HOP	1.75	1.73
	RUNNING DOWN STAIRS	3.00	3.05
29	10 METER DASH	2.25	2.16
	FIGURE-OF-EIGHT	8.68	8.41
	SLALOM RUN	8.17	8.45
	SINGLE HOP	1.46	1.57
	RUNNING DOWN STAIRS	2.34	2.24
30	10 METER DASH	2.23	2.20
	FIGURE-OF-EIGHT	8.21	8.17
	SLALOM RUN	8.41	8.39
	SINGLE HOP	1.59	1.58
	RUNNING DOWN STAIRS	2.52	2.55

APPENDIX F

Best Performance Times/Distance Injured, Braced and Injured, Non-Braced Subjects (N=30)

		CONDITIONS	
SUB NO.	TESTS	BRACED	UN-BRACED
31	10 METER DASH	2.13	2.22
	FIGURE-OF-EIGHT	7.88	7.96
	SLALOM RUN	8.08	8.24
	SINGLE HOP	1.70	1.67
	RUNNING DOWN STAIRS	2.48	2.42
32	10 METER DASH	2.16	2.17
	FIGURE-OF-EIGHT	7.75	7.73
	SLALOM RUN	7.84	7.80
	SINGLE HOP	2.10	2.06
	RUNNING DOWN STAIRS	2.66	2.67
33	10 METER DASH	2.13	2.16
	FIGURE-OF-EIGHT	7.28	7.53
	SLALOM RUN	7.52	7.78
	SINGLE HOP	2.10	2.10
	RUNNING DOWN STAIRS	2.40	2.40
34	10 METER DASH	2.39	2.38
	FIGURE-OF-EIGHT	8.07	8.16
	SLALOM RUN	7.98	7.97
	SINGLE HOP	1.96	2.09
	RUNNING DOWN STAIRS	2.43	2.60
35	10 METER DASH	2.14	2.16
	FIGURE-OF-EIGHT	7.74	7.75
	SLALOM RUN	7.81	7.85
	SINGLE HOP	2.08	2.07
	RUNNING DOWN STAIRS	2.28	2.11

APPENDIX F CONTINUED

Best Performance Times/Distance Injured, Braced and Injured, Non-Braced Subjects (N=30)

		CONDITIONS	
SUB NO.	TESTS	BRACED	UN-BRACED
36	10 METER DASH	2.30	2.32
	FIGURE-OF-EIGHT	8.13	8.31
	SLALOM RUN	8.16	8.36
	SINGLE HOP	2.02	2.01
	RUNNING DOWN STAIRS	3.06	2.98
37	10 METER DASH	1.99	2.01
	FIGURE-OF-EIGHT	7.39	7.47
	SLALOM RUN	7.59	2.60
	SINGLE HOP	2.07	2.09
	RUNNING DOWN STAIRS	2.21	2.28
38	10 METER DASH	2.03	2.08
	FIGURE-OF-EIGHT	8.10	8.38
	SLALOM RUN	8.44	8.59
	SINGLE HOP	1.96	1.96
	RUNNING DOWN STAIRS	2.41	2.77
39	10 METER DASH	2.37	2.34
	FIGURE-OF-EIGHT	8.78	8.91
	SLALOM RUN	8.54	8.66
	SINGLE HOP	1.45	1.54
	RUNNING DOWN STAIRS	2.41	2.65
40	10 METER DASH	2.69	2.84
	FIGURE-OF-EIGHT	11.02	11.10
	SLALOM RUN	10.46	10.56
	SINGLE HOP	1.39	1.33
	RUNNING DOWN STAIRS	4.20	4.26

APPENDIX F CONTINUED

Best Performance Times/Distance Injured, Braced and Injured, Non-Braced Subjects (N=30)

		CONDITIONS	
SUB NO.	TESTS	BRACED	UN-BRACED
41	10 METER DASH	2.28	2.27
	FIGURE-OF-EIGHT	8.09	8.12
	SLALOM RUN	8.20	8.24
	SINGLE HOP	1.93	1.86
	RUNNING DOWN STAIRS	2.81	2.82
42	10 METER DASH	2.56	2.56
	FIGURE-OF-EIGHT	8.89	8.94
	SLALOM RUN	8.74	8.86
	SINGLE HOP	1.55	1.41
	RUNNING DOWN STAIRS	2.35	2.46
43	10 METER DASH	2.10	2.15
	FIGURE-OF-EIGHT	8.74	8.99
	SLALOM RUN	8.42	8.62
	SINGLE HOP	1.75	1.71
	RUNNING DOWN STAIRS	2.41	2.51
44	10 METER DASH	2.20	2.21
	FIGURE-OF-EIGHT	8.49	8.64
	SLALOM RUN	8.87	8.98
	SINGLE HOP	1.60	1.59
	RUNNING DOWN STAIRS	2.08	2.11
45	10 METER DASH	2.22	2.29
	FIGURE-OF-EIGHT	7.99	8.06
	SLALOM RUN	9.00	9.13
	SINGLE HOP	1.75	1.70
	RUNNING DOWN STAIRS	2.39	2.45

APPENDIX F CONTINUED

Best Performance Times/Distance Injured, Braced and Injured, Non-Braced Subjects (N=30)

		CONDITIONS	
SUB NO.	TESTS	BRACED	UN-BRACED
46	10 METER DASH	2.19	2.22
	FIGURE-OF-EIGHT	8.21	8.34
	SLALOM RUN	8.34	8.56
	SINGLE HOP	1.68	1.65
	RUNNING DOWN STAIRS	2.89	2.99
47	10 METER DASH	2.15	2.20
	FIGURE-OF-EIGHT	8.33	8.46
	SLALOM RUN	8.14	8.27
	SINGLE HOP	1.75	1.64
	RUNNING DOWN STAIRS	3.02	3.22
48	10 METER DASH	2.39	2.46
	FIGURE-OF-EIGHT	8.91	9.22
	SLALOM RUN	8.94	9.16
	SINGLE HOP	1.56	1.60
	RUNNING DOWN STAIRS	3.01	3.05
49	10 METER DASH	2.88	3.05
	FIGURE-OF-EIGHT	10.52	11.20
	SLALOM RUN	9.61	9.89
	SINGLE HOP	1.48	1.28
	RUNNING DOWN STAIRS	3.20	3.40
50	10 METER DASH	2.20	2.33
	FIGURE-OF-EIGHT	8.37	8.41
	SLALOM RUN	7.99	7.99
	SINGLE HOP	1.73	1.75
	RUNNING DOWN STAIRS	2.45	2.45

APPENDIX F CONTINUED

Best Performance Times/Distance Injured, Braced and Injured, Non-Braced Subjects (N=30)

SUB NO.	TESTS	CONDITIONS BRACED	UN-BRACED
51	10 METER DASH	2.40	2.41
	FIGURE-OF-EIGHT	9.50	10.32
	SLALOM RUN	9.27	9.94
	SINGLE HOP	1.47	1.47
	RUNNING DOWN STAIRS	3.28	3.29
52	10 METER DASH	2.31	2.39
	FIGURE-OF-EIGHT	10.12	10.34
	SLALOM RUN	8.18	8.33
	SINGLE HOP	1.75	1.70
	RUNNING DOWN STAIRS	3.11	3.18
53	10 METER DASH	2.39	3.54
	FIGURE-OF-EIGHT	10.24	10.37
	SLALOM RUN	9.41	9.62
	SINGLE HOP	1.68	1.0
	RUNNING DOWN STAIRS	3.29	3.32
54	10 METER DASH	2.23	2.25
	FIGURE-OF-EIGHT	9.59	9.87
	SLALOM RUN	9.63	9.86
	SINGLE HOP	1.47	1.40
	RUNNING DOWN STAIRS	3.01	3.01
55	10 METER DASH	2.21	2.30
	FIGURE-OF-EIGHT	9.50	10.33
	SLALOM RUN	9.03	9.18
	SINGLE HOP	1.47	1.45
	RUNNING DOWN STAIRS	3.00	3.10

APPENDIX F CONTINUED

Best Performance Times/Distance Injured, Braced and Injured, Non-Braced Subjects (N=30)

		CONDITIONS	
SUB NO.	TESTS	BRACED	UN-BRACED
56	10 METER DASH	2.25	2.27
	FIGURE-OF-EIGHT	10.49	9.50
	SLALOM RUN	9.89	9.34
	SINGLE HOP	1.60	1.37
	RUNNING DOWN STAIRS	3.23	3.26
57	10 METER DASH	2.33	2.40
	FIGURE-OF-EIGHT	9.32	9.01
	SLALOM RUN	9.06	9.24
	SINGLE HOP	1.55	1.49
	RUNNING DOWN STAIRS	2.89	3.05
58	10 METER DASH	2.33	2.40
	FIGURE-OF-EIGHT	9.32	9.01
	SLALOM RUN	9.06	9.24
	SINGLE HOP	1.55	1.49
	RUNNING DOWN STAIRS	2.89	3.05
59	10 METER DASH	2.37	2.45
	FIGURE-OF-EIGHT	8.63	8.89
	SLALOM RUN	8.65	8.88
	SINGLE HOP	1.57	1.47
	RUNNING DOWN STAIRS	3.00	3.09
60	10 METER DASH	2.32	2.38
	FIGURE-OF-EIGHT	8.43	8.63
	SLALOM RUN	8.34	8.45
	SINGLE HOP	1.38	1.33
	RUNNING DOWN STAIRS	3.12	3.15

APPENDIX G

KT-1000 Measurements (At 30 Pounds) For Injured Subjects (N=30)

SUB. NO.	MEASUREMENT (mm)
31	6.00
32	5.50
33	6.00
34	5.50
35	5.50
36	6.00
37	5.00
38	6.00
39	5.00
40	5.00
41	5.00
42	6.50
43	7.00
44	5.00
45	7.70
46	6.50
47	5.00
48	6.50
49	7.00
50	6.00
51	6.50
52	5.50
53	7.50
54	7.50
55	5.50
56	5.00
57	6.50
58	6.50
59	6.50
60	7.00