The effect of material strength on prophylactic knee bracing

by

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A Thesis Submitted to the
Graduate Faculty in Partial Fulfillment of the
Requirements for the Degree of
MASTER OF SCIENCE

Interdepartmental Program: Biomedical Engineering
Major: Biomedical Engineering

Signatures have been redacted for privacy

Iowa State University
Ames, Iowa

1991

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INTRODUCTION

Since man first started walking upright, the knee has taken abuses it was not designed to handle. Daily, it is bent, twisted, pulled, and compressed. Forces on and within the joint during strenuous activity can reach several times the weight of the body. Today's amateur and professional athletes have pushed their bodies (knees prominently included) to the edge of physical limitations and beyond. The ligaments, which hold this largest and most complex joint of our body together, although very strong, are sometimes strained beyond their capabilities causing painful and debilitating injuries.

Knee bracing, in various forms, has become a common method over the past two decades of buttressing the knee joint that is weak or subjected to excessive forces. Braces have been used to rehabilitate injuries, assist "normal" functioning, and to prevent injury. Although potentially beneficial, knee braces have not yet been perfected, and some controversy exists over the possible harmful effects of bracing.

Many methods have been used to test the efficacy of knee braces, with varying degrees of success. They include in vivo knees, cadavers, and mechanical representations of the knee joint ranging from simple to complex. Each approach has its benefits and limitations. A major stumbling block for each method is the number of variables that must be dealt with and controlled.

Numerous researchers (Baker, et al., 1987; Beck, et al., 1986; France, et al., 1987; Knutzen, et al., 1987; Paulos, et al., 1987; Tegner, et al., 1988) have compared one brace style to another attempting to determine which provides better protection. These comparisons, however, bypass the more fundamental issue of what design variables make one brace more effective than another. To this end, this investigation tested the effect of one variable, material strength of the brace uprights, using one design. By eliminating or controlling all other variables, the benefit of one material over another in buttressing the knee joint was evaluated. The objective of this study was to determine if the material chosen for manufacturing bi-lateral prophylactic brace uprights influences the strength, stiffness, and, ultimately, the level of protection provided by the brace.

The following hypothesis reflected this objective at two levels of testing, low mass/momentum impacts (6.68 kg/15.03 kg*m/s) and high mass/momentum impacts (16.9 kg/38 kg*m/s). The two levels of testing were to determine if the results were dependent on the intensity of the impact.

Hypothesis one: would a decrease in the measured tension on the medial collateral ligament result from increasing the bending resistance of the brace uprights (without changing the brace design) during low momentum impacts?

Hypothesis two: would a decrease in the measured tension on the medial collateral ligament result from increasing the bending resistance of the brace uprights (without changing the brace design) during high momentum impacts, impacts which produce forces in a range that could injure the ligament?

LITERATURE REVIEW

Braces

Knee braces may be categorized into three general groups: functional, rehabilitative, and prophylactic; with some braces having characteristics in several groups.

Functional braces

Functional braces attempt to support and stabilize a knee that is weak or in some way unsteady (Millet, 1987; Podesta, 1988). The brace may be able to return near normal function to an abnormal knee and is typically worn after the rehabilitation period resulting from surgery or non-surgical trauma.

A functional knee brace typically has either (1) a hinge, double uprights, and shell configuration, or (2) a hinge, double uprights, and strap configuration (see fig. 1 and 2). The shell type provides more soft tissue contact area and a stiffer bridge between the uprights than do the strap designs. The hinges are designed to mimic the complex motion of the knee so as not to limit or constrain normal leg motion. Hinges vary widely in configuration and include simple, biaxial, geared-polycentric, multi-axial cam, and posterior offset designs (see fig. 3).

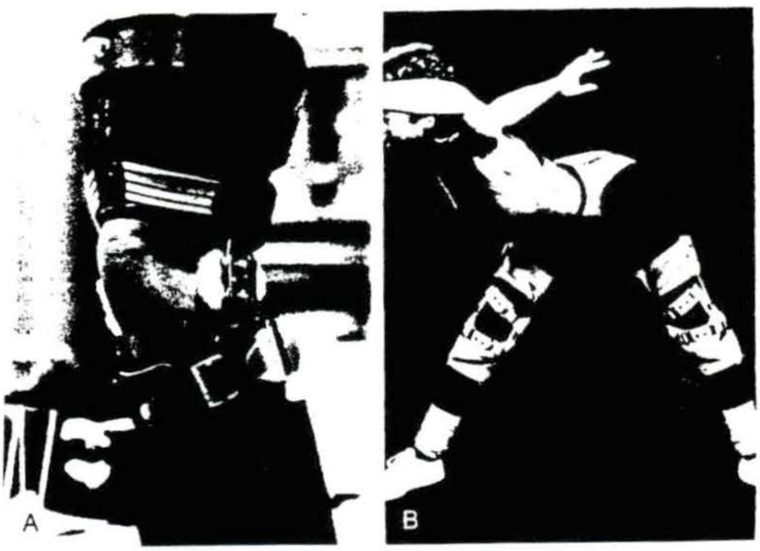

Figure 1. Functional knee braces. A - The hinge, double uprights and shell design. B - The hinge, double uprights and strap design (Podesta, 1988)

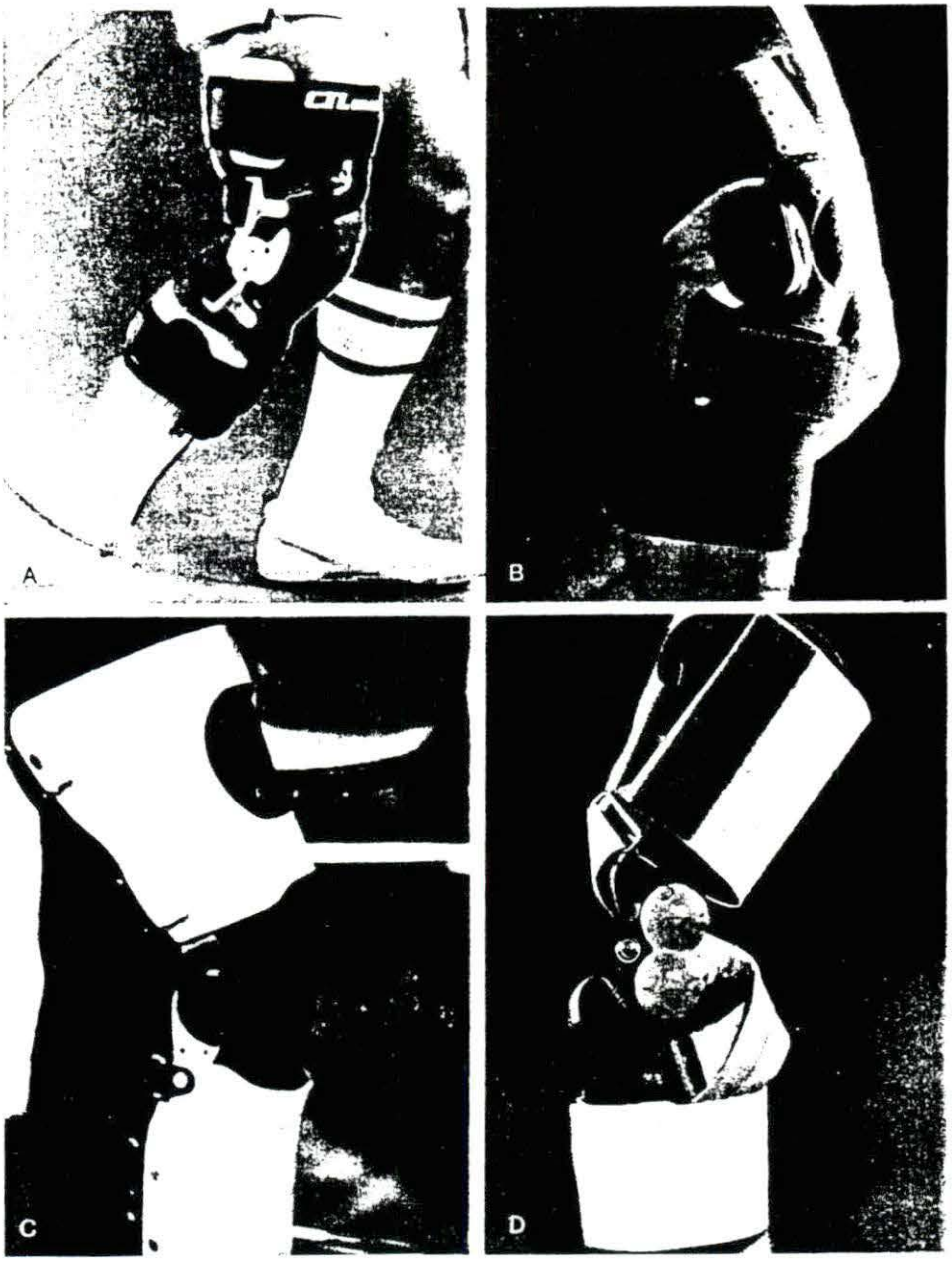

Figure 2. Examples of various functional knee braces (Podesta, 1988)

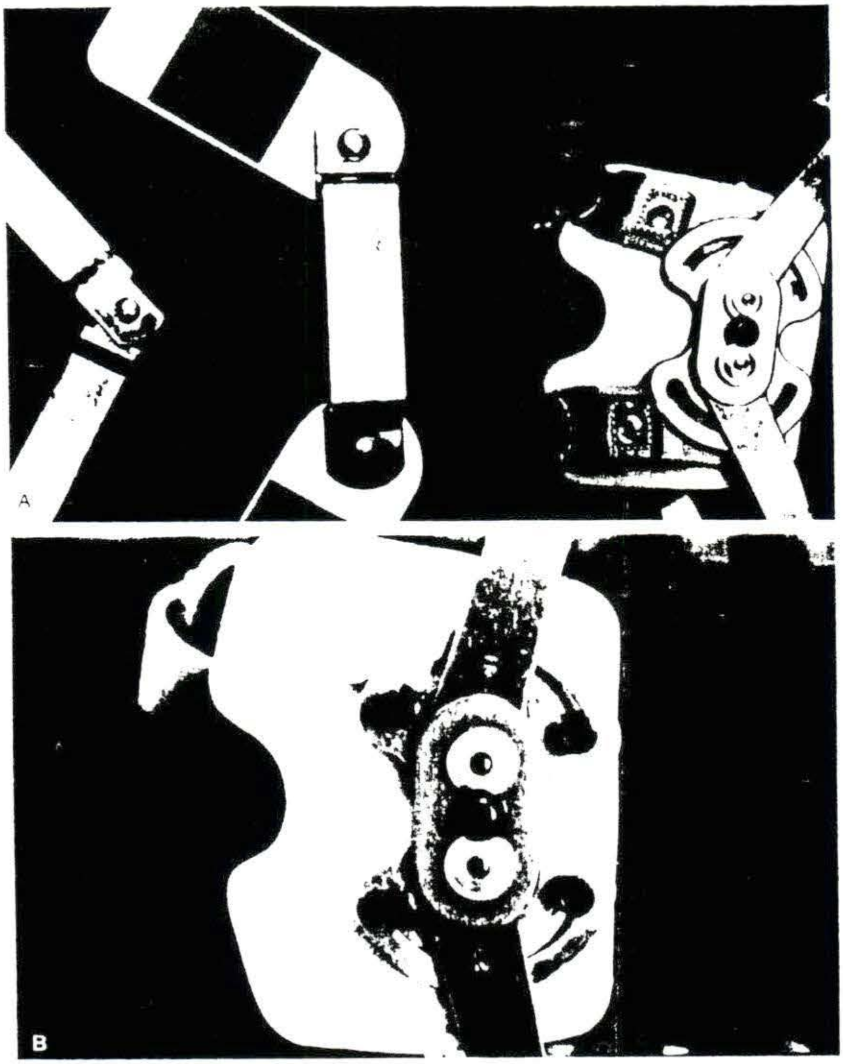

Figure 3. Several types of hinges for braces are shown.
A - Simple, biaxial, and polyaxial hinges. B -
Flexion/extension stops (Hunter, 1985)

Rehabilitative braces

Rehabilitative braces are generally worn during the period immediately following an injury or reconstructive surgery, the intent being to control the extent of motion and provide protection for the ligaments and cartilage as they heal (Millet, 1987 & 1988; Podesta, 1988).

Rehabilitative braces may be quite similar to functional knee braces in design. Common design factors include the double uprights, shells, straps, and various hinges. The rehabilitative brace, however, may have stops incorporated into the hinge design or possibly a foot plate. The hinge stops serve to limit the range of motion during recovery while the foot plate helps prevent slippage of the brace and restrains some rotary tendencies of the lower leg.

Prophylactic braces

Prophylactic braces are used to prevent injuries or lessen the extent of injuries that occur due to contact and non-contact sporting activities (Podesta, 1988; Millet, 1987 and 1988). Consequently, individuals with uninjured or otherwise normal knees are the primary users of this type of brace. Functional and rehabilitative braces also attempt to prevent further injury but are generally more bulky and cumbersome. The prophylactic brace, while preventing injuries, must also be lightweight, inexpensive, and not restrict movement or it will not be worn.

Anatomy of the Knee

The Ligaments

A ligament falls in the category of collagenous tissues, which also includes tendons and the skin. A ligament is composed of collagen, elastin, and reticulin fibers. The function of each component is strength, stretchability, and bulk, respectively. Collagen and elastin account for approximately 90% of collagenous tissue volume. In most ligaments, collagen fibers make up this 90% tissue volume by themselves with almost no elastin present. This disproportionate mixture, along with their nearly parallel fiber orientation, creates strength under tensile loads but forsakes elasticity (Frankel and Nordin, 1980).

Variation in the strength of ligaments is mainly due to their size and shape. The more fibers and greater the cross-sectional area of the bundle, the stronger the ligament is. The other crucial factor is external, ie. the rate of loading. When the rate of loading is high, such as in an impact situation, the load needed to rupture the ligament and the amount of elongation of the fibers is greater than for slow load rates which yield lower maximum loads and less fiber elongation before rupture. Slow rates generally cause avulsions at the ligament-bone insertion where collagen fibers mesh with fibrocartilage which, in turn, gradually mineralizes into the cortical bone. Fast rates generally tear the ligament itself (Frankel and Nordin, 1980; Cornwall, 1984; Paulos, et. al., 1987).

A few authors have reported actual breaking loads for the medial collateral ligament (MCL). Brown, et. al. (1986), found, on average, that valgus forces which produced greater than 12.6% elongation (strain, $\Delta l/l_0$) were enough to avulse (tear) the ligament. They reported 437 N average valgus load upon MCL rupture. However, Paulos, et. al. (1987), measured the peak ligament failure tension at 2346 N under a valgus load of 1058 N and stated the MCL was responsible for restraining 80% of the valgus load, ie. 846.4 N. Given the small sample size (n=3 for Brown, n=6 for Paulos), this difference is not unreasonable.

Consider the following analysis to draw together the similarities of the aforementioned studies. The medial collateral ligament is approximately 12 cm long and 1 cm wide (Nielsen, 1987). At the point just prior to MCL failure, it is strained 12.6%, $\Delta l/l_0$ (Brown, et. al., 1986). Therefore, the change in length, Δl, would be 0.126 times 12 cm, or 1.512 cm. The medial joint opening at MCL failure averages 1.57 cm as reported by Paulos, et. al. (1987), a difference of

3.7%. Another study, by Andriacchi, et. al. (1982), used springs having a constant of 1400 N/cm to mimic the force-elongation characteristics of the MCL. Multiplying 1400 N/cm by 1.512 cm (Δl) equals 2116.8 N of tension on the ligament to produce the 1.57 cm joint opening. The difference from the ligament failure tension of Paulos, et. al. (1987), is 9.8%. However, Brown measured just prior to failure and Paulos at failure which could account for this small difference.

The Knee Joint

The knee joint is the largest and considered the most complex of the body's synovial joints. It is usually thought of as a modified hinge type joint but has slight pivotal as well as gliding movement. The knee is able to move in three planes of motion; frontal, sagittal, and transverse; with flexion and extension in the sagittal plane accounting for the majority of motion. Figure 4 shows the three planes of motion. Normal motion of the knee is approximately 0-140 degrees sagittal, 30 degrees internal to 40 degrees external rotation in the transverse plane, and only a few degrees of either abduction or adduction in the frontal plane (Frankel & Nordin, 1980). This limited freedom of motion in the frontal plane is the underlying cause of medial collateral ligament injuries in contact sports today.

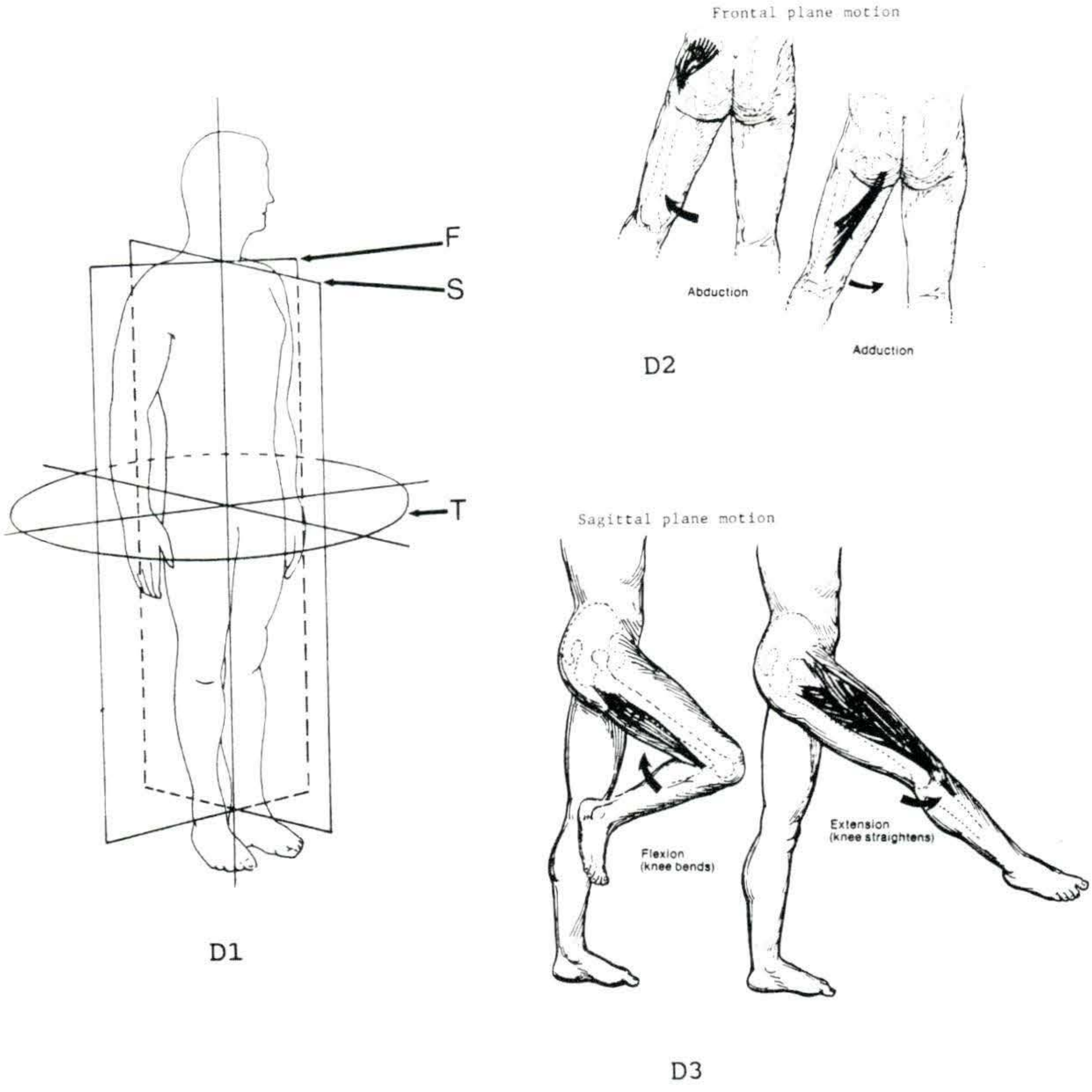

Figure 4. The body planes. F - frontal, S - sagittal, T - transverse (D1 - Frankel and Nordin, 1980; D2 and D3 - Hole, 1984)

The knee joint is held together by a complex pattern of ligaments and tendons. Damage or rupture of these connective tissues causes lack of contact in the joint and leads to abnormal knee motion (Blacharski, et. al., 1975). The major ligaments include the posterior cruciate (PCL), anterior cruciate (ACL), lateral collateral (LCL), medial collateral (MCL), and patellar; each of which can be further subdivided. The PCL functions to limit rotation and prevent forward slippage of the femur. The ACL limits rotation and backwards slippage. Side to side movement is controlled by the lateral ligaments. Adduction (inward bending) of the tibia relative to the femur is limited by the LCL while abduction (outward bending) is limited by the MCL (Hole, 1984; Crowninshield, et al., 1976). For example, a lateral impact in football causes adduction (medial movement) of the joint and a corresponding strain in the MCL. See figure 5 for anatomy of knee and ligaments.

Three separate articulating surfaces make up the knee joint; the femoropatellar and two tibiofemoral joints, each of which is covered by hyaline cartilage for protection. The distal portion of the tibiofemoral surfaces are formed by C-shaped menisci (fibrocartilaginous material), which help support and protect the tibial and femoral condyles from the continuous heavy body weight that the joint must withstand.

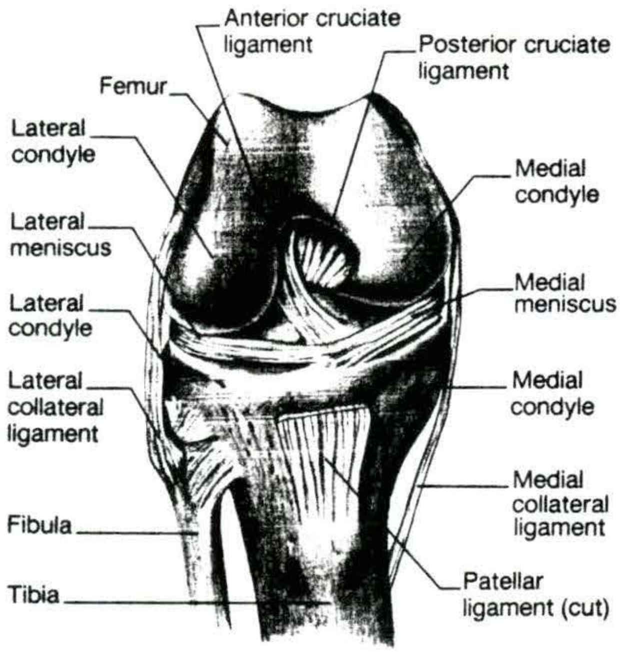

Figure 5. The basic anatomy of the knee joint (Hole, 1984)

Since the femoral condyles are rounded and the tibial "condyle" is basically a flat plateau, the joint is naturally unstable. Several internal and external structures help to correct this. Externally they include: a tough fibrous joint capsule (the body's strongest) which itself is strengthened by the iliotibial tract, tendons of the quadriceps, LCL and MCL, patellar tendon and ligament, and the popliteal ligaments. The ACL and PCL together form the internal support, crossing one another between the femoral condyles forming an "X" (hence the name Cruciate) and attaching to the central area of the head of the tibia and femur (Langley, et. al., 1974).

Knee Bracing

Although functional and rehabilitative knee bracing has been used for a long time, prophylactic knee bracing has been commonplace only since the 1970's, and there is yet much controversy over its benefits and detriments. One method employed to discover whether braces are effective has been through the use of injury rate studies (Anderson, et al., 1979; Grace, et al., 1988; Hansen, et al., 1985; Hewson, et al., 1986; Randall, et al., 1984; Rovere, et al., 1987; Teitz, et al., 1987). Researchers have also tested braces under the conditions in which they will be used, subjecting individual braces to various forces and impacts in order to compare them to one another and against a control situation of no brace

(Baker, et al., 1987; Beck, et al., 1986; Brown, et al., 1986; France, et al., 1987; Knutzen, et al., 1987; Paulos, et al., 1987; Tegner, et al., 1988). While each method has some benefits, no one test has yet provided the definitive answer.

Injury rate studies

The Anderson Knee Stabler One of the first brace usage studies was done by Anderson, et al. (1979), on the Anderson Knee Stabler. They advocated use of their brace "as a preventative device by athletes in vulnerable positions." Anderson and his co-authors concluded this brace prevented significant valgus stress, reduced anterior-posterior laxity, and provided "excellent support to an injured knee." Also, the athletes did not complain of discomfort or show a decrease in performance while wearing the brace. Unfortunately, these conclusions are based on a total of nine football players; one player having played six games with the brace in use, and the other eight players all having used the brace for five or less games over a period of two years.

Another study of the Anderson Knee Stabler was reported by Hansen, et al. (1985). Reviewing medical records from the previous four years at the University of Southern California, they noted "fewer knee injuries to players who used the brace" and concluded the brace helps reduce ligament and meniscal injuries. This study was based on a much larger data pool of

148 braced and 329 unbraced players. However, they still recommended the collection of more longitudinal data and comparisons to other football programs to further evaluate the Anderson Knee Stabler.

These comparisons were made in studies done at Wake Forest University (Rovere, et al., 1987) and another at the University of Arizona (Hewson, et al., 1986), both on the Anderson Knee Stabler. (The early studies on prophylactic knee bracing did not all use the Anderson brace. It was merely one of the first on the market having the benefit of good initial acceptance.) In their study at Wake Forest, Rovere, et al. (1987), looked at knee injuries occurring during a two year period when braces were worn by all players on the football team and the preceding two years when the braces were not used. They found knee injury rates were higher in the years the brace was used than before it was used. This led to the conclusion that the Anderson brace was "ineffective as a prophylactic device". At the University of Arizona, Hewson, et al. (1986), based their analysis on the number of "exposures". An exposure was defined as the number of individual players at a practice session or in a game. From the four years of records studied, this method yielded 28,191 braced "exposures" and 29,293 control "exposures". Hewson reported no statistical difference between the fifty one MCL injuries of the braced group and the forty seven injuries of the unbraced group despite impressions prior to the study that they had reduced numbers of injuries. One of their conclusions, therefore, was that the prophylactic knee brace used in the study did not improve knee injury prevention.

General brace studies Six hundred and ninety-four high school football players in two New Mexico school systems were studied by Grace, et al. (1988). Each player was matched with another of similar build and position on his own team. One player used a brace, a single upright design with either a simple hinge or a double hinge, and the other (control) was not braced. Grace, et al. (1988), found athletes who wore the single hinged brace were significantly more likely to have knee injuries, more severe injuries, and more surgical procedures than their control counterparts. The group wearing double hinged braces also experienced an increase in rate and severity of injuries but not to a statistically significant degree. They noted, however, an unusual number of other injuries to the lower extremity, particularly the ankle and lower leg.

During the first year of the study, nine injuries to the foot/ankle, including severe sprains and fractures, and four fractures of the fibula, one of which occurred at the distal edge of the brace support, were observed. The second year there were twenty eight lower extremity injuries not involving the knee. There was a "large and significant" difference in the rate of this type of injury between braced and non-braced players with braced players being three times more prone to injury. Grace, et al. (1988), hypothesized that since the biomechanical forces on the leg are altered by the brace, injury might be incurred elsewhere due to this transfer of force. They could not find any previous study that had documented this trend toward adjacent injuries. From their data, they concluded single hinge braces compounded the risk of injury and should not be used, and double hinge braces did not decrease the risk of injury.

A large pooling of records was attempted by Teitz, et al. (1987). They looked at seventy one NCAA Div-I schools in 1984 and sixty one in 1985. Teitz was able to gather the injury reports on 6307 players in 1984 and 5445 players in 1985 from these Div-I schools upon which to base their statistical analysis. They were aware, however, of the possible complicating factors involved with such a large and inter-program study. These factors included different coaching philosophies, playing surfaces, naturally fluctuating injury rates, rule changes, and player positions. They believed a large study population would mitigate these effects associated with comparing different schools and that the effects were likely to be small.

Teitz found that braced players in both 1984 and 1985 had a higher injury rate than non-braced players, and the results were statistically significant (1984: 11% vs. 6%; 1985: 9.4% vs. 6.4%). Interestingly, there was no significant difference in 1984 when rates were compared for individual player position. In the 1985 study, there were significant differences in some of the positions. They also found no association between playing surface and injury rates nor any difference in rates between types of braces. Overall, braced players had injury rates no better than non-braced players. Teitz and her colleagues concluded that "so-called preventive braces are not preventive and may in fact be harmful." This strong statement against prophylactic braces drew many responses.

In letters to the editor of The Journal of Bone and Joint Surgery (1987), several authorities questioned the article by Teitz, et al., 1987. Robert F. McDavid, Ph.D., and Lonnie E. Paulos, M.D., both took exception to the use of surveys and questionnaires in gathering scientific data. Derek Brock, R.P.T.-A.T.C., agreed with McDavid that variability in attachment of the brace by so many different players and trainers renders the data suspect. And Gilbert W. Gleim, Ph.D., stated that Teitz's own data shows injuries to be less severe when the braces are used based on the criteria of time lost from play. It appears the outpouring of response to this article was not only due to its content but also on the emphasis placed on it by The Journal of Bone and Joint Surgery. Paulos, in referring to the editorial preceding it (Cowell, 1987), questions the fact that Teitz's article garnered lead position in The Journal of Bone and Joint Surgery and that the editorial seemed to condemn the use of lateral knee braces.

Garrick, et al. (1987), reviewed six studies and outlined the criteria used by them in judging the effectiveness of their braces. They identified factors which could have led to incorrect conclusions in the individual studies, including incorrectly typing the injury, the number of injuries, the number of persons exposed to injury, and comparisons between different times and places. The studies Garrick reviewed also demonstrated knee injuries were associated with position played more than any other variable. Allowing individuals to choose or dictate who wears the brace along with how injuries are defined both influence the data and subsequent conclusions. For example, a player if given the choice, might elect to wear the brace only during practice for fear of hindering performance in the game; or a coach might only brace the players in positions regarded as "unskilled" and not brace those in positions such as receiver or running back for fear of impairing their performance. Either choice would bias the results. Some authorities (Rovere, et al., 1987; Garrick, et al., 1987) agree games are more dangerous than practices and certain positions (eg., linemen, running back, linebackers, tight ends) are more susceptible to injury than other positions. Defining and grading injuries can also change over time, as new medical methods are developed and as team physicians change, influencing, for example, how an injury as minor as a sprain is diagnosed.

Garrick, et al. (1987), commented the practical solution to bracing might be to use braces for whatever benefits they might provide. Some studies (Teitz, et al., 1987; Grace, et al., 1988), though, have shown that there are some possible negative side effects to using prophylactic braces. For this reason, they neither supported nor totally rejected using braces in contact sports.

<u>The Ampro Knee Brace</u> Another prophylactic brace, having two uprights (see figure 6), as opposed to one on the Anderson brace, has been used and evaluated at Iowa State University. Randall, et al. (1984), summarized the use of the Ampro Knee Guard, a semi-flexible nylon and copolymer brace. During tests of agility while wearing the brace they found no

Figure 6. The Ampro Knee Guard

more than a three percent loss over a timed agility course. The players indicated confidence in the brace and its comfort. Only one injury was reported during twenty days of spring practice, while nine injuries were reported the previous year of practice without the brace. No statistical conclusions were drawn from these numbers, only that more study was needed upon which to base valid conclusions.

A follow-up study was done by Brodersen and Symanowski (1988). They looked at time loss and injury severity recorded in ISU football medical files from 1979 to 1987 to judge the effectiveness of the Ampro brace. Their findings indicated an eighteen percent reduction in knee injuries along with some reduction in player time lost. From this, they concluded use of the Ampro brace offered "...a significant reduction in both the overall knee injury rate and the proportion of serious knee injuries."

<u>Uncontrollable variables</u> The various researchers in all of the previously mentioned studies noted other factors which complicated their analysis of knee injury rates. Rovere, et al. (1987), stated that coaching techniques and the types of offensive and defensive formations used can influence the rates of injury. The type of playing surface is also a factor, along with the choice of footwear. For example, natural grass is a safer surface than artificial turf

(Garrick, et al., 1987; Hewson, et al., 1986). The NCAA has continually changed and modified rules in an attempt to reduce injuries. All of these things constitute uncontrollable variables that confound the statistical results. Some have suggested that these variables are negated by large statistical populations and by long term, inter-program studies (Teitz, et al., 1987), but this remains to be proven. In addition, it is possible a favorable bias of results could occur due to the instituting of bracing after a disastrous season and, therefore, the tendency of the number of injuries to naturally revert to a more normal, lower value even without bracing. That is, the effect of regression of the sample population mean back to the level of the whole, or true, population mean regardless of the influence of factors such as knee bracing (Teitz, et al., 1987). Brodersen and Symanowski (1988), believed they had overcome and negated this phenomenon by the length of their study (eight years). Hewson, et al., (1986) however, concluded that knee injuries on a team are "a random event with rare occurrence" in light of the number of chances for an injury to occur. Rovere, et al. (1987) stated the naturally occurring frequency of injuries may not be discernable from a two year study, ie., the true average will not be found.

The problem of adequate sample size is another complicating factor. Due to the relative infrequency of MCL injuries, for example, statistical differences are often difficult to resolve. Garrick, et al. (1987), identified this statistical problem with sampling size.

> If we assume the overall rate of MCL injury to be that found in the WF study, for a 90% chance of finding a 50% reduction in MCL injuries (from 13.5 to 6.75 per 100 player-seasons) with a one-tailed test at the 5% level of significance, it would take 250 to 300 athletes in each group, or a total of 500 to 600 player-seasons of exposure. To detect smaller, perhaps more realistic reductions, much larger numbers would be required. Reliably detecting an increase in ACL injuries would be even more difficult, as they occur about one-fourth as often as MCL injuries. Player-seasons exposures in excess of 4,000 would be required to have a 90% chance of detecting a 50% increase at the 5% level of significance. For this reason alone, carefully designed, multicenter studies will be essential if these controversies are to be satisfactorily resolved.

Obviously, unless a new brace is dramatically better than the control, or injury rates increase drastically, these controversies will continue for some time to come.

Biomechanical testing

Prophylactic braces A biomechanical force test was done by Brown, et al. (1986), on the Ampro Knee Brace and a unilateral brace. Their test was designed to simulate a lateral (valgus) impact on the knee. Using cadavers, they fixed the foot (laced into an athletic shoe) to a rigid support and placed the hip in a two dimensional pivot. The knee joint was then struck by a concave impactor driven with a servohydraulic motor. A liquid metal strain gauge was sutured to the MCL which, when stretched, would produce linear voltage changes in the liquid metal (mercury). The strain measurements were then paired with the corresponding measurements of valgus load applied, knee deflection, and the testing machine command stroke signal (when the loading started and stopped). The test took approximately 370 milliseconds during which time the valgus load reached 625 Newtons and an MCL strain of about 7.1%.

The tests showed an average relative strain relief (unbraced strain minus braced strain divided by unbraced strain) of 18.3% for the unilateral brace and 25.1% for the bilateral brace. These were statistically significant from the unbraced leg but not statistically different from each other. There was no statistical difference between the apparent stiffness of the braces, defined as the change in force divided by the distance the knee was deflected. Brown concluded there was "reasonable evidence" the braces helped protect the knee to some extent.

Paulos, et al. (1987), also used cadaver legs for testing. Their goals were to determine if clinical static testing was relevant to actual use, what forces were necessary to damage the ligaments and related restraining structures, and to define the mechanical properties of several commercial braces. They believed the mechanical and material properties of the brace under static and dynamic loads were important to understanding its function in vivo. Using an Instron machine, specimens were subjected to static nondestructive and low rate destructive loads in a three point bending fixture. Low rate loads were calculated as 73% strain/sec., equivalent to a lateral impact of about two tenths of a mile per hour, and a somewhat higher rate (to test the load rate dependency of the MCL) of 856% strain/sec., or 2.35 mph.

Results showed substantial data variation between individuals but much less on repeated tests of a single sample. Static testing of the Anderson Knee Stabler and the McDavid Knee Guard showed little change in the engineering bending stiffness on the knee. However, Paulos, et al. (1987), concluded that static testing was sufficient to judge a braces on field performance, because impacts were longer in duration than the natural frequency of either the braces (most were greater than 100 Hz) or the knee (10 Hz).

Low rate failure testing by Paulos, et al. (1987), yielded peak ligament tensions of 1122 N for the ACL, 1406 N for the PCL, and 2346 N for the MCL at the point of rupture. Their contributions to valgus restraint were 11% (ACL), 9% (PCL), and 80% (MCL). The higher load rate produced higher ultimate rupture loads and higher stiffnesses in the MCL but decreased strain accommodation, ie., the ligaments ability to withstand strain.

Citing brace rigidity (those tested were less rigid than the knee), and joint line clearance, or its lack (ie. the space between the hinge and the knee), Paulos felt these braces were not effective in preventing harmful valgus forces. More study was needed on the combined biomechanical relationship between the knee and the brace with regard to its mechanical and material properties.

A follow-up study was done to access the impact response of the braced knee (France, et al., 1987). They tested several commercially available braces, including the two prophylactic braces used in the Paulos (1987) study, with impact loading. The impact tests were done on a complex mechanical knee joint/leg/lower torso apparatus designed to mimic the function of the human knee as closely as possible. The resulting "surrogate" knee was so unique, it was patented.

The surrogate limb of France, et al. (1987), was composed of cast aluminum bones with the exact shape of the head of the femur/tibia complex, along with steel cables representing the major groupings of tendons and ligaments in the knee joint. Each of the tendons and ligaments was instrumented to sense forces due to the impacts. Tests were done on the free standing surrogate under varying impactor masses, flexion angles, and with the hip and foot either constrained or unconstrained, using both braces and no brace for comparison.

Results showed the MCL tension was greater for fixed foot vs. free and straight leg vs. flexed when using a constant impact force. The braces (Anderson and McDavid) were most effective, in general, with higher mass/low velocity impacts (vs. low mass/high velocity impacts of the same energy), fixed foot and hip, and straight leg alignment. The braces were rated by an Impact Safety Factor (ISF), defined by France as

$$\frac{\text{MCL peak tension, unbraced/impact momentum, unbraced}}{\text{MCL peak tension, braced/impact momentum, braced}}$$

The Omni Anderson rated 1.29 ISF and the McDavid rated 1.18 ISF. France proposed a minimum ISF of 1.50 as a standard level of safety which equates to a 30% reduction in MCL load. Several functional braces that were tested did rate higher than the two prophylactic braces. Interestingly, the only brace that just met the minimum ISF was made by a company that supported this research project. France, et al. (1987), concluded that the current prophylactic braces available were biomechanically inadequate. They believed, however, based on further refinements of brace material properties and mechanical design, that prophylactic knee bracing could be made effective.

In a different study, which also used the Anderson Knee Stabler, the McDavid Knee Guard, and several functional braces on cadaver legs, Baker, et al. (1987), had similar results. They measured force in the MCL and ACL, and abduction angles with an electrogoniometer due to valgus loads at three angles of knee flexion. The two prophylactic braces demonstrated from 0 to 6% reduction in abduction angles at the various angles of flexion while the functional braces were somewhat better at 0 to 23%. The prophylactic braces did not reduce the measured force in the MCL and one, the Anderson Knee Stabler, seemed to increase the force on the ACL.

Baker concluded from his data that functional braces, especially those with more soft tissue containment, provided some protection for the MCL while prophylactic braces gave little or no protection.

<u>Functional</u> <u>braces</u> Several other studies dealing only with functional braces are noteworthy in the examination of prophylactic bracing if only because of the similarity in design features of the two types of braces.

Knutzen, et al. (1987), examined the Marquette Knee Stabilizer and the Generation II knee braces. An electrogoniometer was used to measure total knee movement in all three planes during running trials at a controlled speed.

They reported a reduction in varus/valgus motion of approximately four degrees (24%) for both braces.

Beck, et al. (1986), tested seven functional braces. They used a Stryker Knee Laxity Tester and a Medtronic KT-1000 device to measure anterior tibial displacement on three patients. Although some braces were reported as better than others, no statistical differences were evident due to the small sample size combined with the small amount of protection afforded by each brace.

Two studies evaluated the Lenox Hill brace (Colville, et al., 1986; Wojtys, et al., 1987). Colville looked at forty five patients with ACL deficiencies and compared the brace to no brace in anterior subluxation of the tibia, rotary instability, and lateral instability. They also used a subjective questionnaire to determine satisfaction with the brace. Their results showed some improvement in objective stability measurements but the perceived functional improvement by the patients was the most notable benefit.

Wojtys, et al. (1987), did their testing on four cadaver legs. They built a special apparatus to apply force to the tibia in both anterior and posterior directions. The distal and proximal ends of the leg were fixed rigidly in the device. Movement of the knee joint due to these applied forces was measured by a triaxial electrogoniometer attached rigidly to the bones. The data showed the Lenox Hill brace decreased anterior translation from 10.2 mm to 5.6 mm at thirty degrees of flexion under no axial load. The brace also decreased external rotation of the tibia an average of seven degrees after the ACL was sectioned. Under all other sets of conditions the Lenox Hill brace did not improve the protection of the knee.

Four derotation braces and an elastic knee cuff were evaluated by Tegner, et al. (1988), using an electrogoniometer to measure range of motion in all three planes. The braces' effect on strength was tested on a Cybex II isokinetic device and their effect on performance was tested by running a figure eight. In a slideboard test, simulating skating, all braces reduced abduction/adduction by about forty percent. The elastic cuff showed no effect. The strength and performance tests showed some reduction when wearing the braces. Tegner concluded that although the braces showed some benefits, nothing was proven concerning how much force they could resist, which would be an important factor.

Modeling Knee Joints

Mechanical models

Mechanical models of the human knee joint can be very complex to create owing to the complexity of the actual knee.

A benefit of mechanical models over cadaver testing, however, is working with fewer constraints. The model is readily available, can be built to exhibit as few or as many variables as needed, and does not come with regulations concerning its use as do cadavers. Quoting Nisell (1985), " . . . the knee biomechanical model is considered as a useful instrument for quantifying knee joint forces." Researchers are now faced with the task of determining how best to make the model and what materials to use.

A mechanical model has several basic components related to the basic anatomy of the leg. The bone structure provides the rigid shape of the model and may be cast from a material to exactly match the surface features of a real bone or it may be of a more general shape, ie. a metal rod or tube. Aluminum has been shown to have good characteristics relative to strength and bending that simulate the bones (Mason, et al., 1989). The material choice, however, is not critical so long as it has the general physical characteristics of bone.

The researcher also has the choice of how to fix or hold the ends of the model leg (or cadaver). The model can be limited to one plane of motion to minimize the number of variables or it can be mounted so as to duplicate the three dimensional motion of the actual hip and ankle joints (Inoue, et al.,1987; Wojtys, et al., 1987; Baker, et al., 1987).

Mason, et al. (1989), said that fixing the foot and hip effectively makes "..the leg into a rigid beam with the knee as the weak spot.". This scenario would obviously be the worst case, more likely to cause injury than if the body could deflect with the motion of the impact. The surrogate knee of Mason, et al. (1989), was also capable of a free standing position, which allowed impact tests on a deflectable target.

The knee joint itself has been represented as a simple hinge or shaped to form the condyles of the femur and tibia. The hinge limits the degrees of freedom of motion and so restricts the information that can be obtained from it. The more natural condylar surface, if properly formed with anatomically correct ligament placement, can produce the most lifelike results. However, the natural shape is difficult to duplicate and also is very hard to hold together in its natural position (Mason, et al., 1989). The more lifelike results are necessarily more complex, though, and therefore more difficult to interpret. Other designs fall between these extremes including ball and socket types, saddle joints, or gliding joints (Hole, 1984).

The ligaments themselves are variously made of springs, cables, elastic bands or combinations of these or other connective materials. A simple model used by Smith, et al. (1988), relied on springs alone to constrain the knee joint.

No data on the physical characteristics of these springs was provided. A much more complex model combining sheathed steel cables connected to springs of known load constants was used by Mason, et al. (1989). The individual spring associated with each cable, which in turn represented an individual ligament or tendon, was chosen to have the correct strain characteristics of the element it represented. The force/elongation properties of the cables were likewise known and combined with that of the springs. As was stated before, however, the exact material characteristics are not critical and do not have to perfectly match their anatomical counterparts; they merely have to have known values and be reproducible in repeated test procedures (Mason, et al., 1989).

When constructing a mechanical model with the intent and purpose being to apply braces to it, the tissue over the bone must also be considered. The force on the brace is meant to be transferred to the leg and away from the knee joint. Pressure on the skin surface from the brace is transmitted in a diffuse way through the tissues to the internal bone structure (and ligaments connecting them) which resists the applied force (Brace, et al., 1977). Mason, et al. (1989), suggested a "polymeric or similar material" to cover the bone structure of the mechanical leg to create a more realistic load dissipation between the force on the brace and that measured on the ligaments. Obviously, a direct connection of the brace to the bone by rigid means would have a stiffening effect, but would not be a realistic representation of brace usage.

Mathematical models

Mathematical models simulate and take into account the geometry of the knee, the ranges of motion, and the different physical characteristics of the parts of the knee joint. In order to do this, they must rely on previously obtained physical data on which to base the mathematical relationships. The bone and condylar surfaces are generally represented as rigid bodies with the soft tissue, ligaments, and tendons treated as linear or non-linear springs or beam type elements (Crowninshield, et al., 1976; Wismans, et al., 1980; Andriacchi, et al., 1982).

The ligaments, although strong in resisting tensile loads, do exhibit mild visco-elastic properties as they will stretch to a limited degree under force. A simple approach to model this behavior was used by Wismans, et al. (1980), and by Andriacchi, et al. (1982). They used various arrangements of strong springs to allow limited elastic movement of the ligament. Wismans, et al. (1980), used the following quadratic force elongation relationship to explain the mechanical behavior of the ligament

$$\underline{f} = \underline{k} * (\underline{l} - \underline{l}_0)^2$$

$\underline{F}$ was in Newtons, $\underline{k}$ being the spring constant, $\underline{l}$ the stretched length of the spring, and $\underline{l}_0$ the initial length of the spring. The constant used for the MCL was divided into the anterior and posterior parts, each being fifteen Newtons per square millimeter. Andriacchi, et al. (1982), used a the folowing force elongation relation without squaring the difference:

$$\underline{f} = \underline{k} * \Delta x$$

(Δx was the displacement, $(\underline{l}-\underline{l}_0)$, of the spring element) They represented the MCL as four springs of varying lengths and stiffnesses. The total spring constant force was 1400 N/cm.

MATERIALS AND METHOD

The objective was to test the effect material stiffness had on the brace's ability to protect the knee. An apparatus was constructed upon which to conduct impacts on the knee either with or without the brace. A single brace design was used and uprights made of three different materials were tested. The data from these impacts was collected by a program onto a personal computer for later analysis.

The Test Apparatus

The apparatus used consisted of a wooden framework used to support a pendulum, an artificial leg, and sensors with which to collect the data (see Figure 7).

The framework

The supporting frame for the artificial leg and the impactor pendulum was constructed of two by six inch lumber. The members were glued with construction grade adhesive and nailed with eight penny nails for maximum strength. The height of the frame was dictated by the length of the "average" leg. This will be discussed further in the section entitled "The Artificial Leg."

A solid half inch diameter steel pin (or axle) held the leg at hip height while the ankle pin was held in a hinged

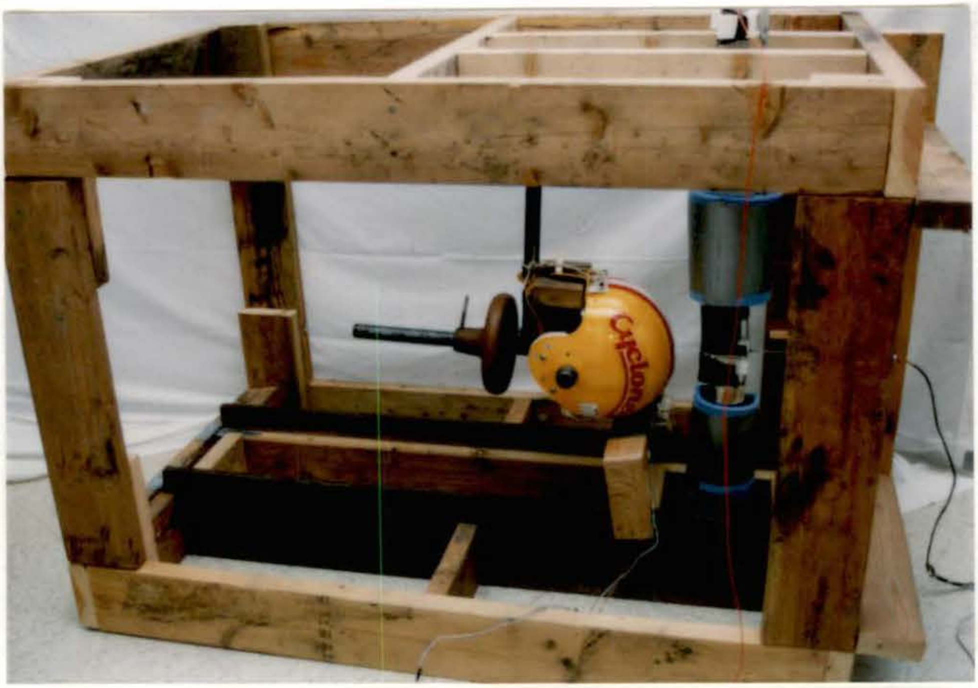

Figure 7. The framework of the test apparatus

frame within the main frame. This hinged frame allowed the ankle pin to move upward slightly as the knee joint was bent during impact. As the knee joint was bent, its vertical height decreased slightly. The hinged frame, however, prevented the artificial leg from moving laterally with the impact and together with the upper fixed hip pin prevented rotation of the leg. The pins at the ankle and hip allowed only lateral motion of the leg (in the frontal plane) of the test apparatus, and were inserted into bearing sleeves that had been firmly pressed into the aluminum leg. The steel pins were isolated from vibrations in the framework by rubber plugs around their ends where they fit into the wooden frame (see Figure 8).

The framework above the pendulum was notched to accommodate the arm of the pendulum. The notch could be fitted with different plugs to position the pendulum so it would be swung from the same height each trial. The pendulum could be released from thirty, sixty or ninety degrees of arc. The pendulum was manually pulled back until it was in firm contact with this notched position and then released (see Figure 9).

<u>The pendulum</u>

A pendulum was used to impact the knee joint of the artificial leg with varying mass and velocity. It pivoted on

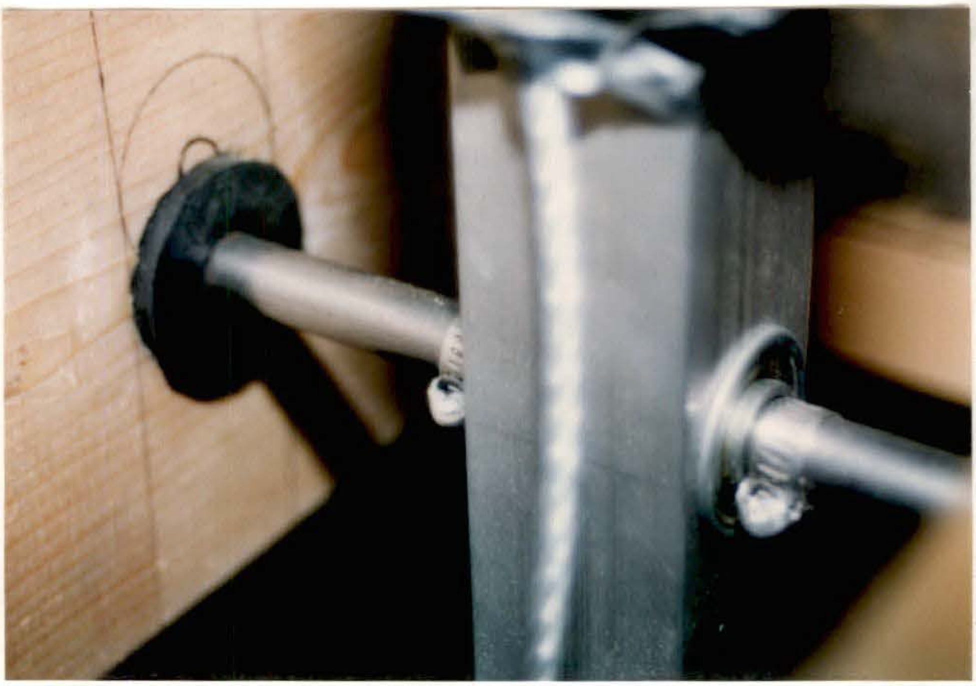

Figure 8. The upper, fixed "hip" position showing the rubber mounts to dampen vibrations and the bearing sleeves on which the shaft pivots

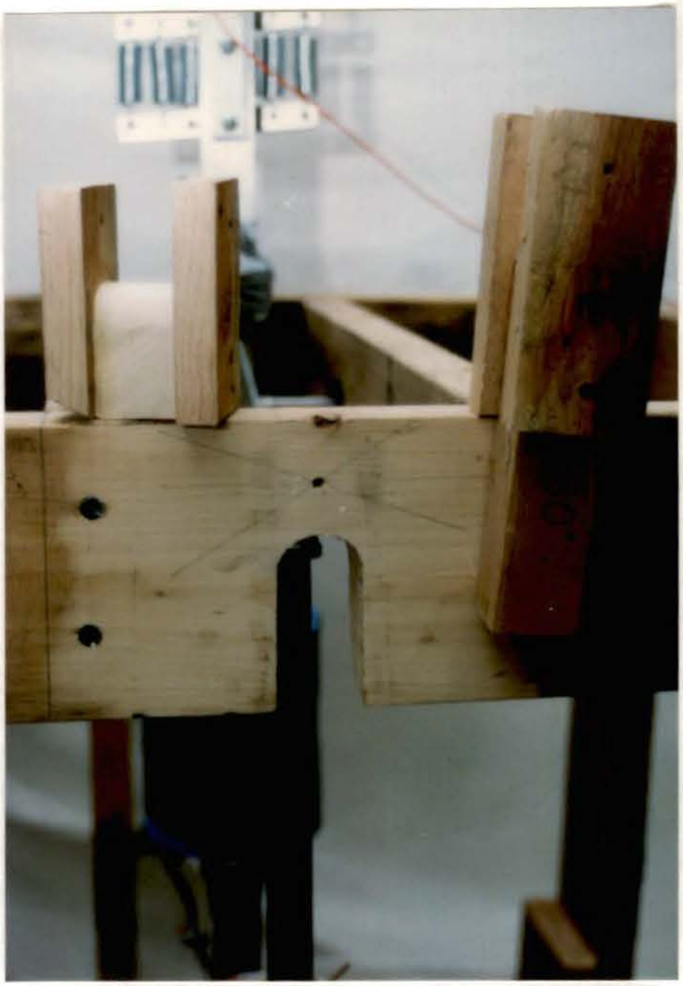

Figure 9. The framework notch for positioning the pendulum

an axle inserted into bearing sleeves at the same height as the hip pin. It was located at a horizontal distance from the hip pin so that it impacted the knee joint at the bottom most point of its swing, at which point it had its maximum velocity. The pendulum was made of three quarter inch "black" pipe with a "T" at the bottom angled away from the point of impact. Additional weights could be placed on this "T" section to change the mass of the pendulum. The impactor surface was an actual football helmet attached to the pendulum. This gave the pendulum a realistic, large, rounded contact point with which to hit the knee joint (see Figure 10).

<u>The artificial leg</u>

The leg was made from square aluminum tubing with one eighth inch wall thickness. Aluminum has been found to have the desirable characteristics of strength, stiffness, and bending that are similar to bone (France, et al., 1987; Mason, et al., 1989). Square tubing was chosen to limit torsional motion on impact. The sides of the square help direct the bending force into the frontal plane even if the impact was slightly off center.

The length of the leg was chosen to match the estimated "average" 1985 male (Anthropometric Source Book. Vol.1, 1978). Based on their estimated values the tibia is 35.1 cm and the

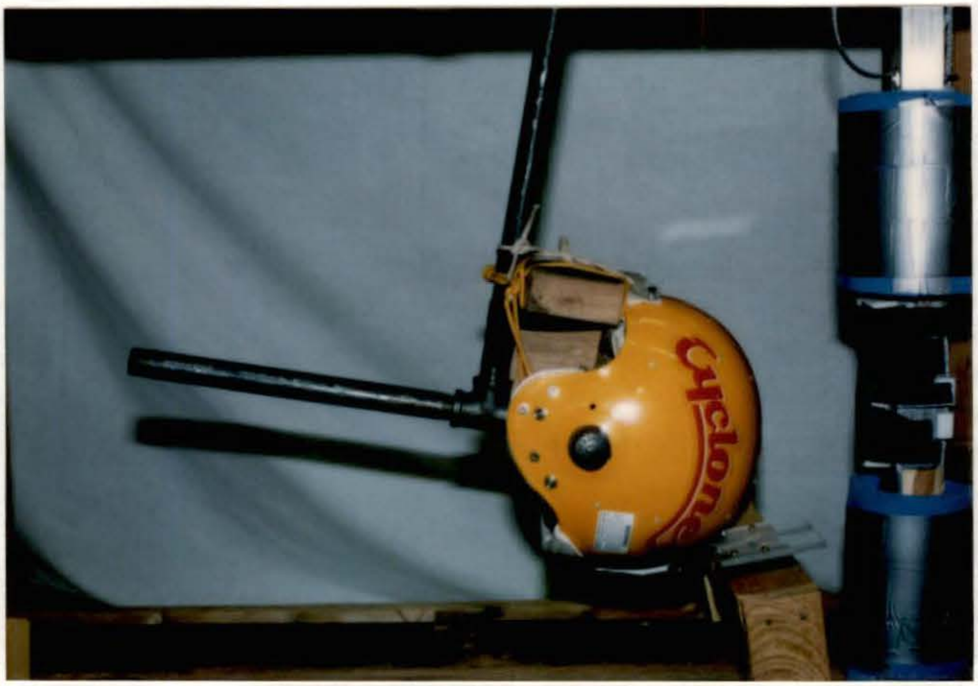

Figure 10. The pendulum for impacting the leg

femur is 45.7 cm giving a total leg length from ankle to hip of 80.8 cm. The thigh and calf circumferences were also constructed to match the "average" 1985 male to provide the proper soft tissue bulk under the cuffs of the brace. They were 59.5 cm and 37.5 cm, respectively.

The tissue bulk was built up on the square tube in three layers. First, half moon shaped strips of wood were applied to give the square tube a round shape. Second, commercially available "Bio-Soft Gel" terry cloth covered wrist weights were slipped onto the tubes from each end. These gave the leg added inertial weight and also the firm sponginess of real muscle tissue. Finally, the Gel layer was wrapped with closed cell Ensolite foam to give the leg the required circumference (see Figure 11).

The knee joint itself was modeled as a simple hinge. A heavy duty strap type hinge was bolted to the distal end of the femur and the proximal end of the tibia. The axis (or pin) of the hinge was on the lateral side of the test leg. Normal sagittal bending of the knee would dictate putting the axis of the hinge on the posterior side of the test leg. Since no motion in any plane but the frontal plane was to be measured, the bending action of the hinge was in this plane (ie. on the lateral side), rotated ninety degrees from normal sagittal bending. The pendulum impacted on the side of the

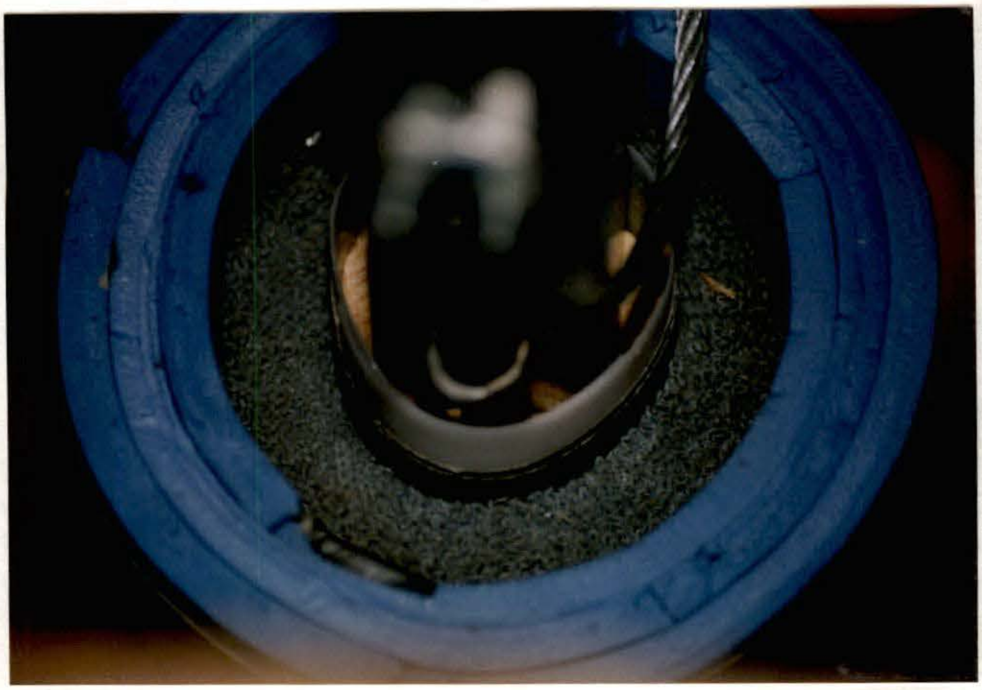

Figure 11. Shows the layers used to make "thigh" tissue around the upper part of the test leg

leg held by the hinge (the lateral side of the model knee) causing it to bend open on the medial side. This bending action was resisted by the artificial medial collateral ligament (a steel cable) (see Figure 12).

The MCL was modeled by a three-eighth inch diameter, seven by nineteen stranded steel cable with a working load rating of nine hundred eighty pounds (Mason, et al., 1989). The distal end was anchored to the lower tibia by two cable clamps. The cable passed over the medial joint opening along the center of the aluminum tube, gliding on two grooved Teflon blocks, one on either side of the joint, minimizing friction on the cable during bending of the leg. The cable was retained and guided along its path by several eyelets. The proximal end of the cable was clamped onto a bar attached to a set of six springs. The springs transmitted pressure via a steel "U" bolt and flat plate to a quartz load cell (see Figures 12,13,14_a&14_b).

The spring set had a constant of approximately 470 Newtons per centimeter (see Figure 15) which is somewhat less than that used by Andriacchi, et al. (1982). The springs allowed stretchability similar to a ligament when tension was put on the cable. The springs themselves were all identical, commercially available, seven centimeter long, Select-A-Spring brand #166 springs.

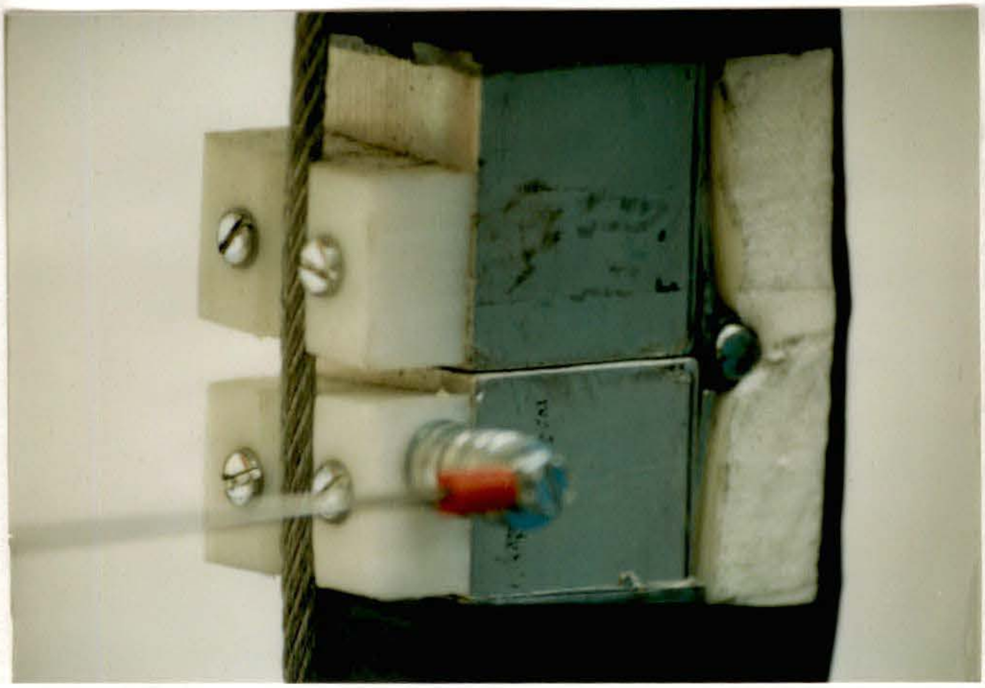

Figure 12. The joint of the test leg showing the aluminum tubing, the hinge (on the right), and the steel cable gliding on the Teflon blocks

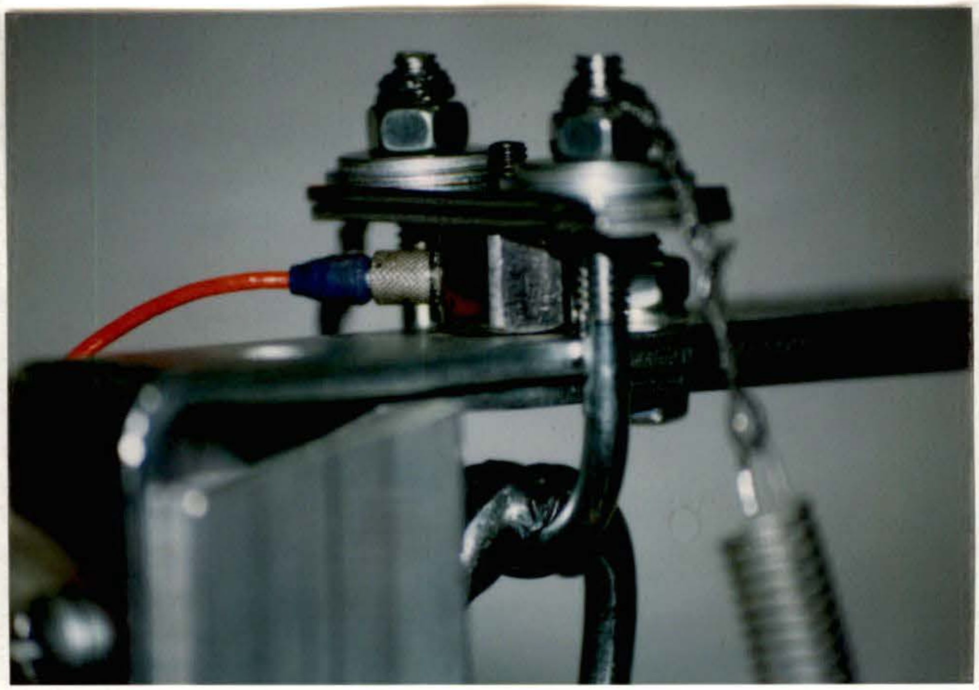

Figure 13. Close-up view of the quartz load cell, "U"-bolt, and pressure plate on the sensor platform

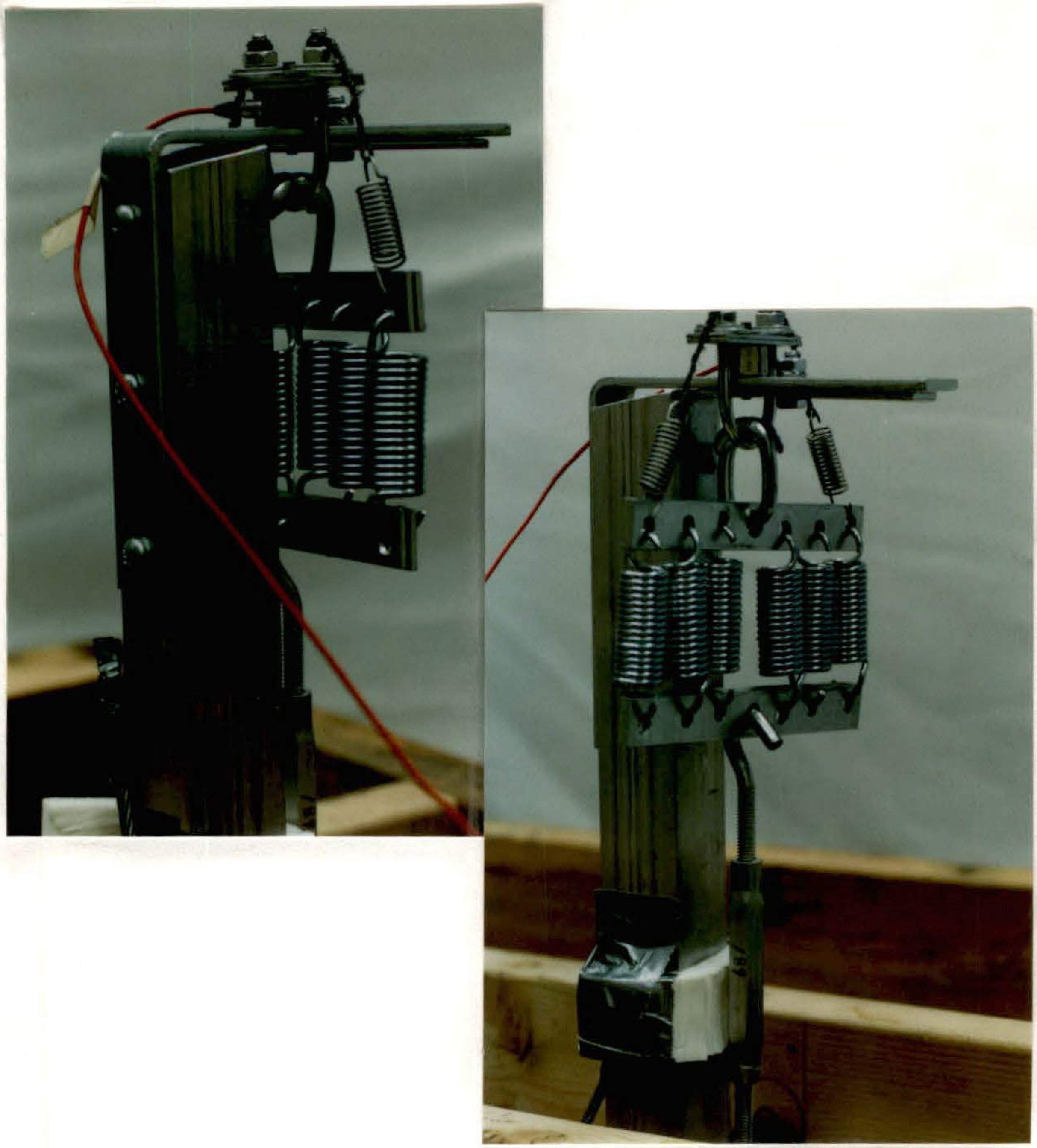

Figure 14_a and 14_b. These photos show two views of the set of springs and sensor platform on the proximal end of the test leg

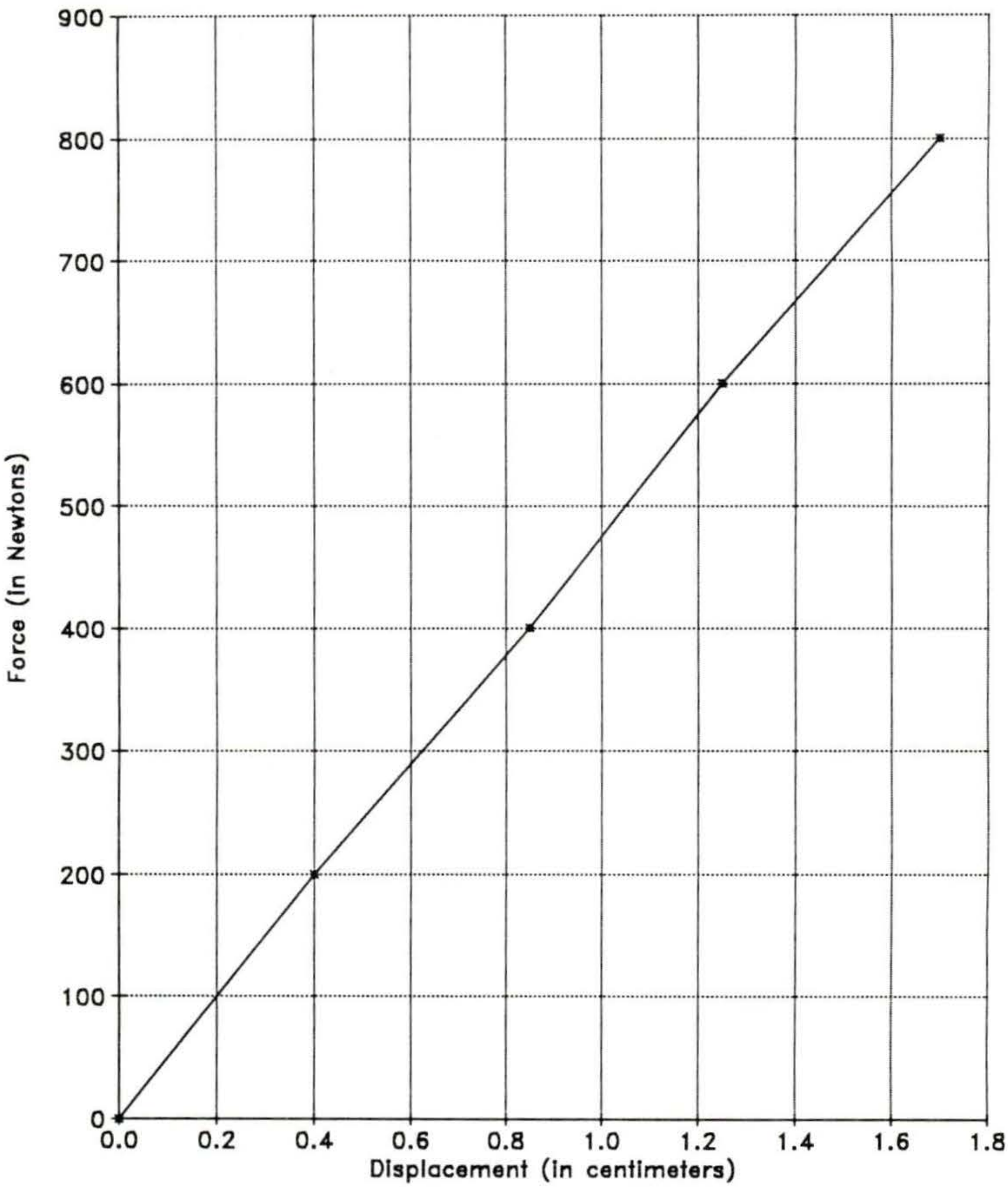

Figure 15. The six spring set had a constant of 470 N/cm

The brace

The brace style used in this research was the Ampro Knee Guard, a bi-lateral prophylactic brace. The uprights were fabricated of nylon and the cuffs from a combination of polypropylene and polyethylene. The exact chemical composition of both the uprights and the cuffs was not reported in Randall, et al., 1984. The cuffs were held onto the leg by bands of neoprene with velcro attachments (see Figure 6).

The Ampro Knee Guard was modified for use in the two other experimental conditions. The original nylon uprights of the brace were removed and duplicated in 6061-T6 aluminum as well as graphite fiber. Figure 16 shows the relative bending stiffness of the aluminum upright and the plastic upright. The aluminum and graphite configurations were much stiffer than the plastic, and should, therefore, according to the hypotheses of this research, result in less knee displacement than either the control or plastic configurations.

The sensors

The load cell used to measure tension in the cable was a Kistler Model 912 Quartz Load Cell capable of sensing forces up to 5,000 (22,000 Newtons). The breaking strength of the medial collateral ligament is much less than this and is well within the range of the load cell (Paulos, et al., 1987).

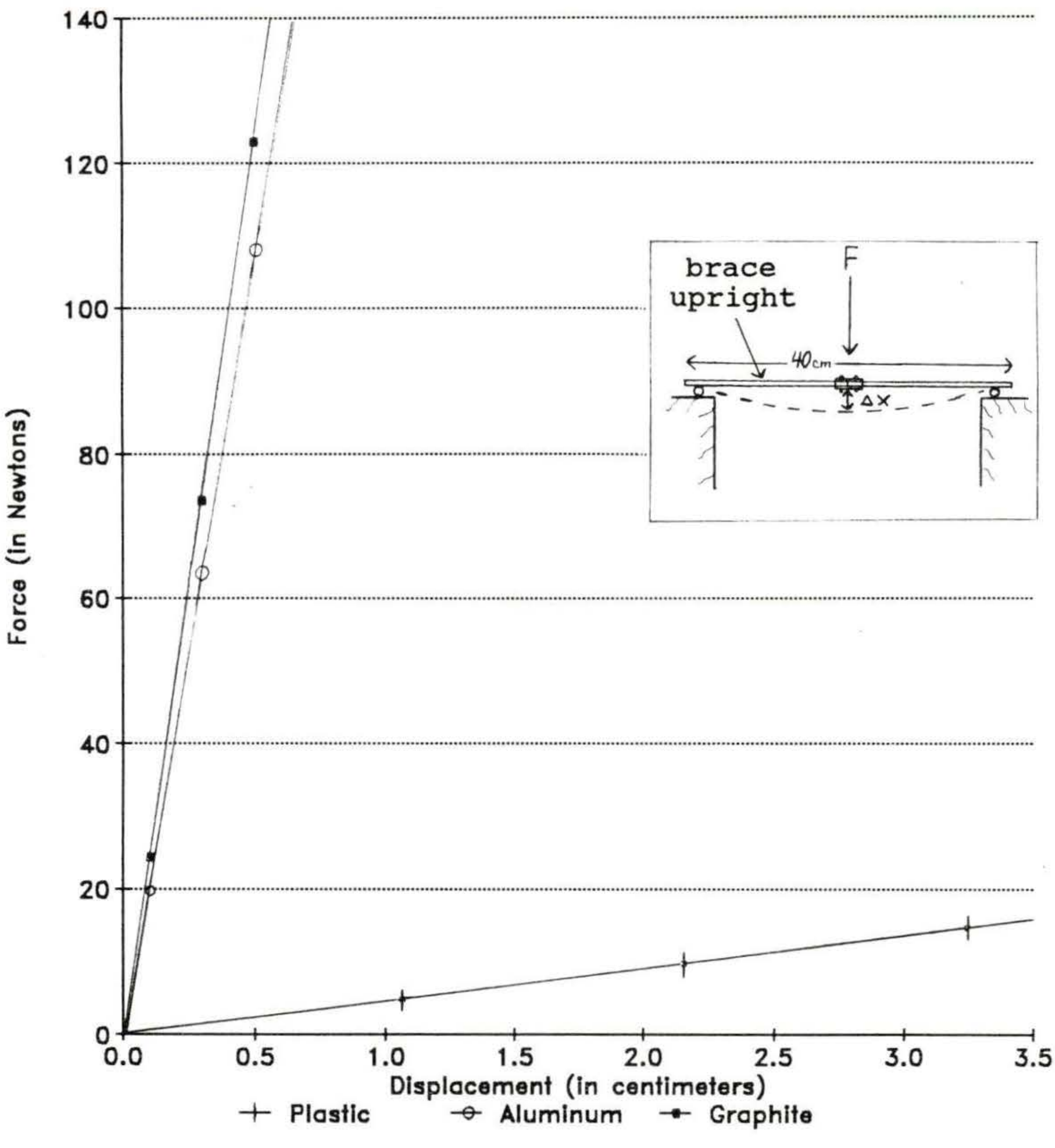

Figure 16. Comparison of material bending stiffness

The load cell was connected to a Kistler Model 568 Universal Electrostatic Charge Amplifier. The amplifier converted the static electrical discharge of the quartz load cell into DC volts per unit of force on the cell. This output was fed into a Keithley analogue to digital converter and then to a personal computer.

A manual calibration of the paired load cell and charge amplifier was done prior to testing to double check the accuracy of the calibration data supplied with said instruments. Once calibrated, the settings on the paired instruments were untouched throughout testing and only the baseline was zeroed (grounded) prior to collecting each set of data. The procedure followed for calibration and zeroing was performed as per the instruction manuals supplied by the manufacturer. The analogue output of the charge amplifier was ten millivolts per pound of force (or per 4.45 Newtons). At this setting it was capable of sensing forces up to 1,000 pounds, the approximate limit of the working load of the cable, and more importantly a force high enough to rupture a real ligament. Actual applied loads during testing only reached about half this limit.

In addition to the load cell, a displacement sensor was also used for the lower weight impacts. This displacement measuring device was an HP Sanborn 7DCDT-1000 Displacement

Transducer. The transducer was used to measure sideways displacement of the knee during the less forceful, lower weight impacts. The higher weight impacts pushed the knee beyond the range of this device.

The voltage output of the transducer was connected directly to the Keithley A/D converter. Output was manually calibrated prior to testing. The transducer was clamped to the frame of the testing apparatus while the moving core rod was attached to the knee joint. This method of connection allowed the rod to freely piston in and out of the transducer (see Figure 17).

Data Acquisition Program

Data were collected (on an IBM personal computer) under the control of a program written in QuickBasic. The program (Appendix A) read the digital signal from the Keithley A/D converter and stored the information on disk. The stored data were analyzed using a Microsoft Works spreadsheet. Averages, standard deviations, and statistical differences were calculated while in this spreadsheet. All graphs and charts using the data were also created by this media.

The final form of the Basic program took approximately 1.2 ms to execute the sampling loop (830 samples per second).

Figure 17. The displacement transducer is attached to the frame by a pivoting clamp. The core rod is attached to the knee joint on the left edge of the photo

During this loop, it read data from the quartz load cell and the displacement transducer.

The duration of the impact event was about 100 to 170 milliseconds depending on the test conditions, therefore, the impact frequency was 6-10 Hz. In order to get a valid sample, at least two or more samples per cycle must be taken (Black, 1953). That means twenty or more samples per second were needed. Since the sample rate of the program was much higher, many other frequencies were being detected. These unwanted frequencies included electrical power noise, high frequency vibrations and other unknown sources. To eliminate this problem, digital filtering loops were written into the program. This method allowed the higher frequencies to be analyzed and then eliminated so as not to bias the fundamental frequency of the impact event being tested. Four different loops, filtering four main higher frequencies were used. The filtered signal was then stored on disk to be analyzed later.

The pendulum initiated the start of the data collection loop of the data acquisition program. A photo cell gate (see Figure 18), fixed to the frame of the apparatus in the path of the pendulum, was hooked to the Keithley A/D converter and constantly monitored by the program. Just prior to impact, the photo cell was tripped by the pendulum to start the data collection. In this manner, timing of the sample points from

Figure 18. The photo cell gate

trial to trial was coordinated for valid comparison. Variations of no more than two or three milliseconds between trials were apparent.

Data Analysis

First, the data were imported into a Microsoft Works spreadsheet for calculations and comparisons. There were four cases of cable (ligament) tension data for each of the two pendulum weights and four cases of displacement data for the lighter pendulum weight. The four cases represented the control data (impacts on the leg with no brace), the plastic Ampro brace (as manufactured), the Ampro modified with aluminum uprights, and the Ampro modified with graphite uprights. There were twelve spreadsheets in all, one for each case. Each case resulted in a spreadsheet of fifty trials (columns) with each trial having two hundred samples (rows).

Each row (sample) represented a point in time. These samples were averaged over the set of trials for each case. This averaged set of points was then compared to the other cases experiencing the same pendulum impact weight.

RESULTS

The data collected in this research supported the original hypothesis. That is, the material chosen in the manufacture of bi-lateral prophylactic brace uprights influenced the level of protection afforded by the brace to the knee. The following results indicated a stiffer material provided a greater degree of protection than did the original brace configuration or no brace at all. However, there was an increase in force transferred brought about by the increase in stiffness of the brace. When hit with impacts of equal momentum, the stiffer brace returned to normal more quickly than the relatively more flexible brace (or no brace). This shorter duration of impact for the stiffer brace caused the force transferred to the leg to be greater even though the momentum was the same.

Low Impact Force

The pendulum in the low impact case was released from sixty degrees of arc with a total mass of 6.68 kg. The velocity at the bottom of the arc (the point of impact) was 2.25 m/s or 5.03 mph. Velocity was calculated from:

$$v=\sqrt{2gh}$$

where $g = 9.81\ m/s^2$ and $h = 25$ cm (Tipler, 1982).

Momentum of the pendulum was defined to be the mass of the pendulum times its velocity (M = m*v), which in this situation is 15.03 kg*m/s. The momentum was held constant from trial to trial and case to case.

From Newton's Second Law, the summation of force imparted to the knee by the pendulum is the change in momentum divided by the length of time necessary to transfer the force. This force is not always the same from case to case as the time of contact between the pendulum and the knee varied with the type of brace. In all trials, however, the cases using the graphite and aluminum braces had shorter durations of impact (by about twenty eight milliseconds) than did the control case. The plastic braced knee had shorter impacts than the control case by about ten milliseconds. This means the total force experienced by the braced knees was up to twice as much as that experienced by the control knee, due strictly to the duration of impact.

As can be seen on Figure 21, the durations of impact for the control, plastic, aluminum, and graphite cases are 55, 46, 27, and 27 milliseconds, respectively. The resultant summations of forces from Newton's Second Law for each is thus 273 N for the control case, 326 N for the plastic case, and 557 N for both the aluminum and graphite cases.

Displacement

Due to the mechanical setup and the limited range of the displacement transducer, high impact forces caused the lateral displacement of the test knee to be greater than the transducer was capable of measuring. For this reason, measurement of displacement was only carried out under the condition of low impact force.

Control The test knee with no brace applied was displaced laterally a maximum of 4.80 cm (1.89 in.) on average. The standard deviation of this maximum was .030 cm (.012 in.), which is only 0.6% of the mean. See Table 1 and Figure 19.

Plastic brace In this case, the test knee wore the Ampro Knee Guard as manufactured and with no modifications. The maximum displacement, as can be seen in Table 1 and Figure 19, was reduced to 4.62 cm (1.82 in.) from that of the control case. The standard deviation of the maximum is .048 cm (.019 in.), a 1.0% variation. This maximum displacement represents a 3.6% reduction from that of the control case. The reduction is significant at $p < 0.01$.

As a further comparison, the areas under the mean curves in Figure 19 were analyzed. A simplified method to calculate the "area" was used. Since each data point represented a very small segment of time (and also a very small segment of the X

Table 1. LOW MASS/MOMENTUM IMPACT RESULTS

STATISTICAL COMPARISON OF DISPLACEMENT MEASUREMENTS

The maximum displacement (the peak of the curves in Figure 19) is the average of 50 trials for each case.
All percent reductions from control are significant at $P < 0.01$

	Maximum Displacement	Standard Deviation	% of Control	% Reduction from Control	Statistical Results: d.f.= 98 t value compared between		
					Control	Plastic	Aluminum
Control	4.80 cm 1.89 in	0.030 cm 0.012 in	100	0			
Plastic	4.62 cm 1.82 in	0.048 cm 0.019 in	96.4	3.6	21.4		
Aluminum	3.86 cm 1.52 in	0.036 cm 0.014 in	80.6	19.4	140.4	89.3	
Graphite	3.81 cm 1.50 in	0.025 cm 0.010 in	79.6	20.4	174.3	104.4	7.81

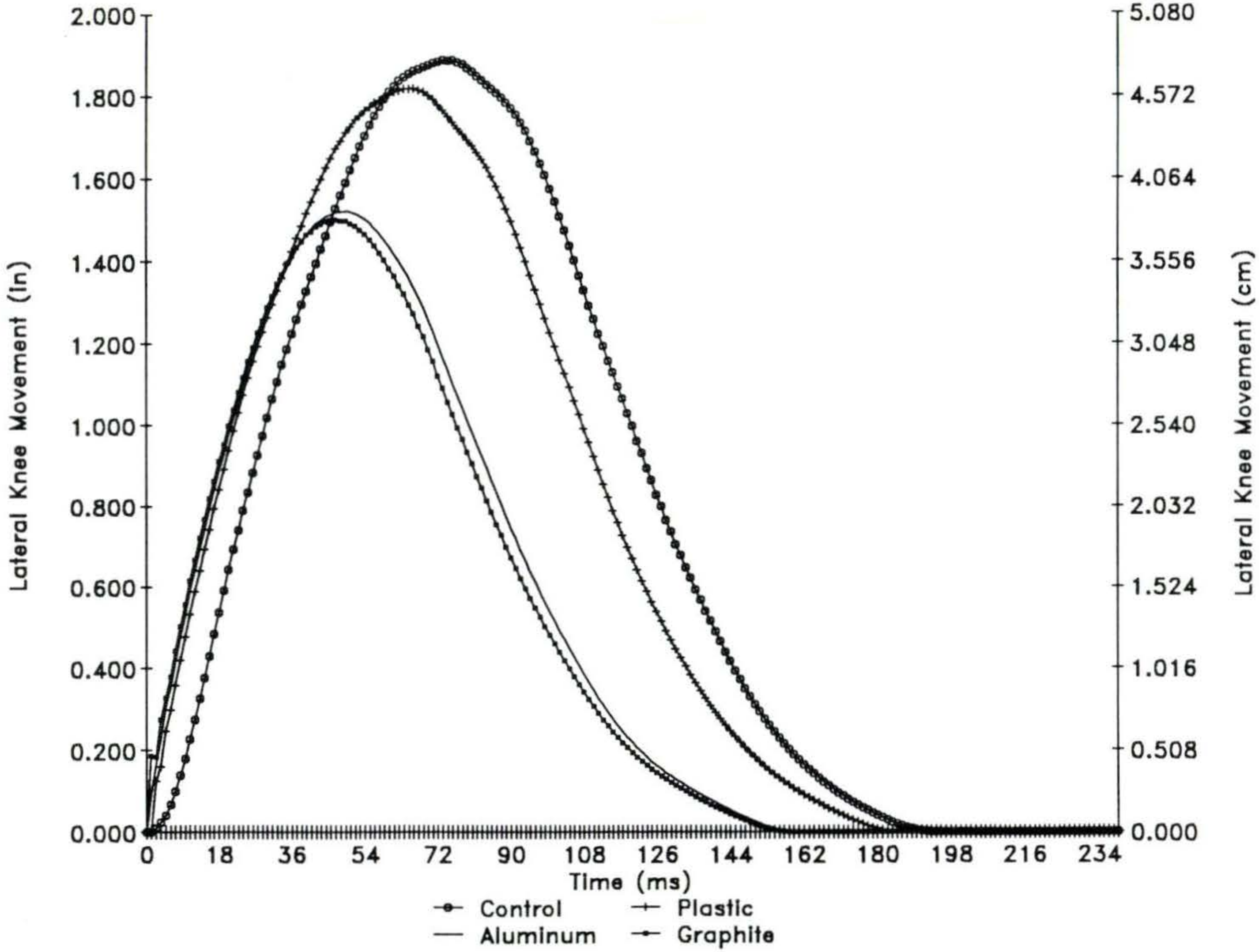

Figure 19. Low impact momentum mean displacement curves

axis), the raw displacement values were added together for each mean curve. The "Maximum Area" shown in Table 2 is not really a two dimensional area, but rather a summation of the individual data point values for each mean curve.

The area values in Table 2 should be thought of as the relative amount of displacement along with the amount of time that the knee was displaced. A large "Maximum Area" number would indicate either a greater displacement, a greater length of time displaced, or both (e.g. the control case as shown in both Table 1 and Figure 19). The 138.88 in. for the plastic case (Ampro Brace), is a 6.4% reduction from the 148.38 in. of the control case.

In most cases, the percent reduction from the control case is greater when looking at Table 2 than when looking at Table 1. As can be seen from Figure 19, the reduction in area is due not only to the lower maximum displacement but also the fact that the curve returns to zero sooner. The added stiffness of the brace on the test leg most likely caused this quicker "spring" back to zero displacement. Figure 20 shows graphically the percent reductions listed in Table 2.

Note should also be taken of the positive slope portion of all three braced case curves on Figure 19. Although there is area under these curves that is not under the control curve, it is not enough to offset their quicker return to the

Table 2. LOW MASS/MOMENTUM IMPACT RESULTS

STATISTICAL COMPARISON OF AREA UNDER DISPLACEMENT CURVES

The maximum area value is a summation of magnitudes (in inches of displacement) of the individual sample points which make up the mean curves of Figure 19. The column "% of Control" is shown graphicgally in Figure 20. All values are significant at $P < 0.01$

	Maximum Area (inches)	Standard Deviation	% of Control	% Reduction from Control	Statistical Results: d.f.= 98 t value compared between		
					Control	Plastic	Aluminum
Control	148.38	0.694	100	0			
Plastic	138.88	1.683	93.6	6.4	36.9		
Aluminum	96.64	1.011	65.1	34.9	298.3	152.1	
Graphite	93.46	0.479	63.0	37.0	460.5	183.5	20.1

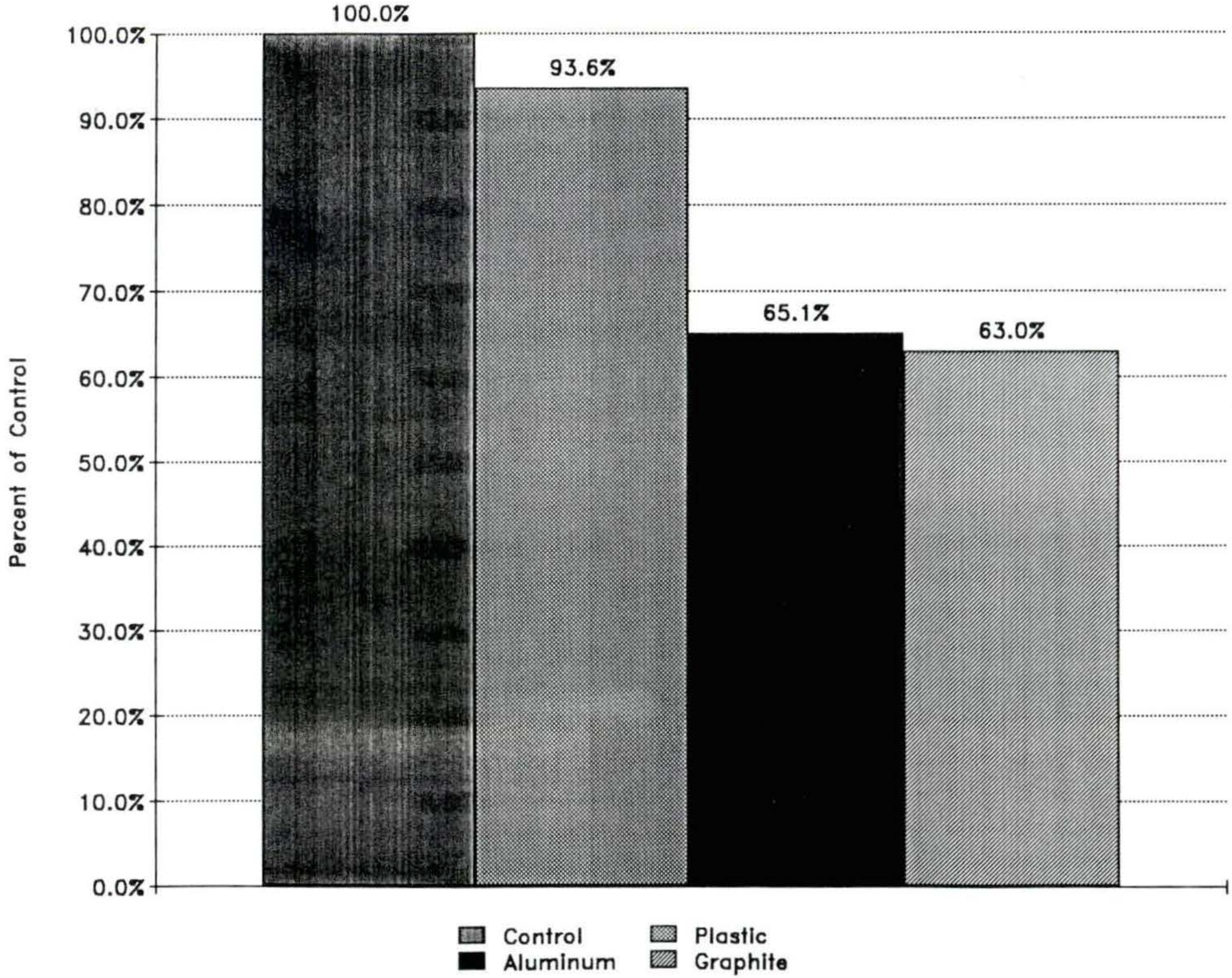

Figure 20. Low impact momentum percent of control displacement

zero baseline. The apparent reason that all three braced case curves have the positive slope portion of their curve sooner than the control case is the added width of the brace. The pendulum came into contact with the braced test leg several milliseconds sooner (after triggering the photo cell) than it came into contact with the unbraced test leg. The hinge of each brace was a half inch thick or 1.27 cm. There was also several millimeters space between the hinge and the joint surface. At the pendulum's maximum velocity of 2.25 m/s, this extra 1.3 to 1.4 cm would equate to the pendulum hitting about 6 ms sooner.

<u>Aluminum brace</u> Maximum displacement with the aluminum brace was 3.86 cm (1.52 in.), a reduction of 19.4% from the control and 16.5% from the plastic brace. The standard deviation of .036 cm was only 0.9% of the maximum. The difference between the aluminum case and both the control and plastic cases was significant at $p < 0.01$.

The area under the aluminum brace curve was 96.64 in., a reduction of 34.9% from the control case and 30.4% from the plastic brace. Both reductions were significant at $p < 0.01$. The standard deviation of the aluminum curve area was 1.01 (or 1.0% of the value).

<u>Graphite brace</u> The original Ampro brace was again modified using a graphite/epoxy matrix as the material used in

the brace. It can be seen in Figure 16 that the graphite upright was somewhat stiffer than the aluminum. The initial aim had been to create a graphite upright that was substantially stiffer than the aluminum, but the resultant uprights were not, probably due to the simple design of their manufacture in the lab. Even though they were not much stiffer than the aluminum, the graphite uprights had a resiliency better able to withstand the higher impacts which will be discussed later.

Graphite uprights, compared to the other materials, had the least lateral displacement under low impact conditions. Their maximum lateral displacement was 3.81 cm (1.5 in.) with a standard deviation of 0.025 cm (0.010 in.). The s.d. was 0.7% of the maximum. The graphite case offered 20.4% reduction from the maximum displacement of the control. It was 17.6% and 1.3% better, resp., than the plastic and aluminum cases. All reductions were significant at $p < 0.01$.

The area in the graphite case was 93.46 in. with a standard deviation of 0.479, a 0.5% variation, a reduction of 37.0% from the control value. Graphite had less total displacement than the plastic case by 32.7% and the aluminum case by 3.3%.

Tension

Measurements of tension in the steel cable simulating the medial collateral ligament were taken under both the low and high impact forces. The range of the quartz load cell was not a limiting factor in the high impact condition as was the range of the displacement transducer.

Control The test leg without a brace developed 438.4 N (98.56 lbs.) of tension with a standard deviation of 3.40 N (0.765 lbs.). This deviation was 0.8% of the maximum force. See Table 3 and Figure 21.

Plastic brace The maximum tension in the test leg with the Ampro Knee Guard applied was 437.8 N (98.43 lbs.) with a standard deviation of 2.28 N (0.513 lbs.), a 0.5% variation. The reduction from the control case was 0.1%, and was not significant at $p < 0.05$. See Table 3 and Figure 21.

Areas under the plastic and control curves did not show the same trend. As can be seen in Table 4 and Figure 22, the plastic case had 0.7% more area than did the control case. The standard deviation of the area value was 1.1%. The difference in area was significant at $p < 0.01$.

Aluminum brace The average maximum tension was 358.6 N (80.63 lbs.) with a standard deviation of 2.44 N (0.548 lbs.). This was a 0.7% variation around the maximum. The reduction from the control case was 18.2% and from the plastic

Table 3. LOW MASS/MOMENTUM IMPACT RESULTS

STATISTICAL COMPARISON OF TENSION MEASUREMENTS

Maximum tension is that in the steel cable of the test leg (the peak of the curves in Figure 21), which represents the MCL. All t values greater than 2.33 indicate $P < 0.01$

* not significant with $P = 0.15$

	Maximum Tension	Standard Deviation	% of Control	% Reduction from Control	Statistical Results: d.f.= 98 t value compared between		
					Control	Plastic	Aluminum
Control	438.4 N 98.56 lbs.	3.40 N 0.765 lbs.	100	0			
Plastic	437.8 N 98.43 lbs.	2.28 N 0.513 lbs.	99.9	0.1	1.044 *		
Aluminum	358.6 N 80.63 lbs.	2.44 N 0.548 lbs.	81.8	18.2	134.8	167.7	
Graphite	363.0 N 81.62 lbs.	1.00 N 0.225 lbs.	82.8	17.2	150.3	212.2	11.9

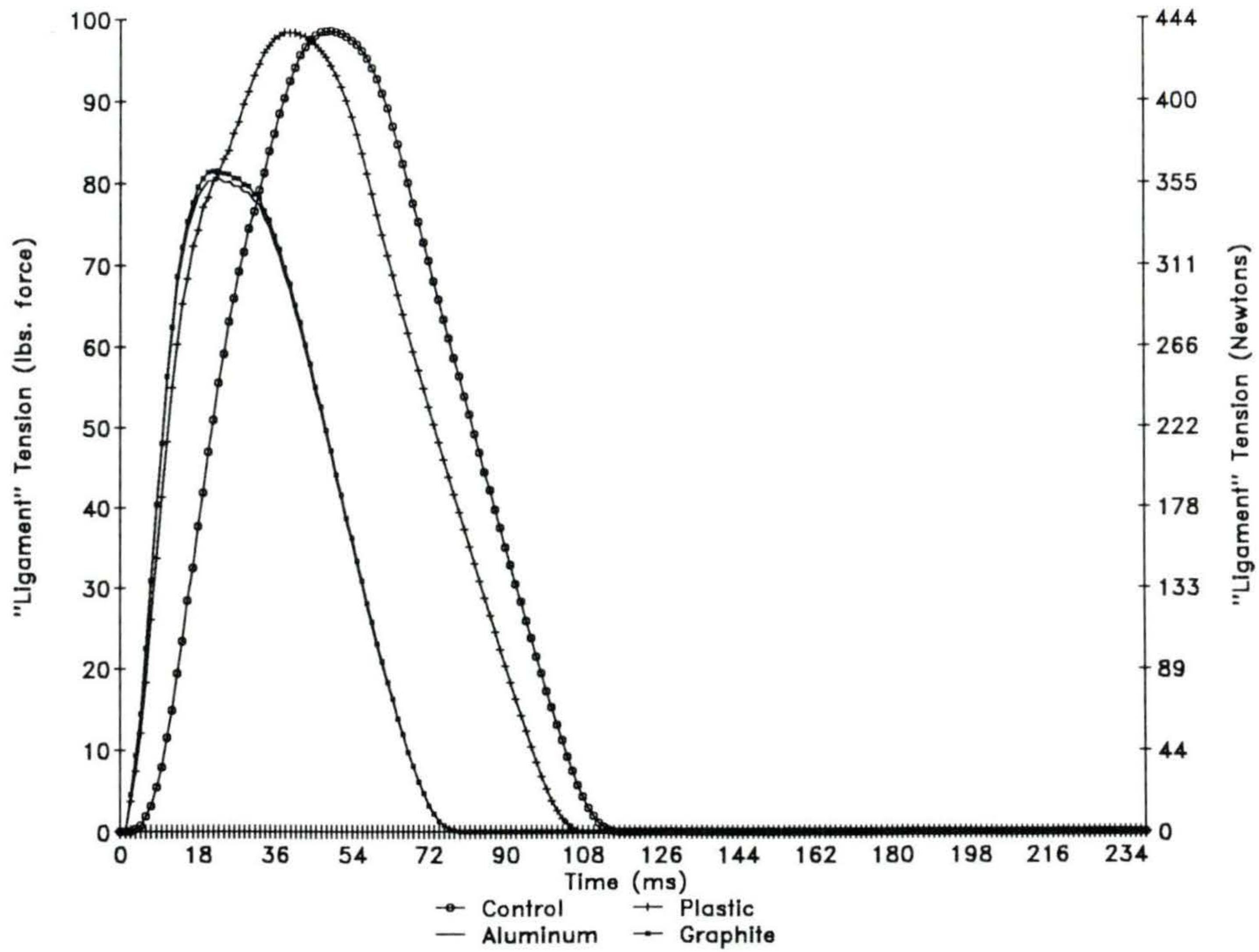

Figure 21. Low impact momentum mean tension curves

case, 18.1%, both of which were significant at $p < 0.01$. See Table 3 and Figure 21.

The area reduction under the aluminum curve vs. the control was 41.6%. The standard deviation of 28.57 lbs. was only 1.0% of the maximum value. Figure 22 graphically shows the percentage of the control case from Table 4. The area reduction from the control and from the plastic case was significant at $p < 0.01$.

<u>Graphite</u> <u>brace</u> The maximum tension in the graphite case was 363.0 N (81.62 lbs.) with a standard deviation of 1.00 N (0.255 lbs.). This was a 0.3% variation around the maximum. The reduction from the control, plastic, and aluminum cases was 17.2%, 17.1%, and -1.2%, respectively. All differences were significant at $p < 0.01$. See Table 3 and Figure 21.

The area under the graphite curve was 40.9% of the area under the control curve. The standard deviation of 7.32 lbs. was 0.2% of the maximum value. The area differences between this and all the other cases were significant at $p < 0.01$. See Table 4 and Figure 22.

Table 4. LOW MASS/MOMENTUM IMPACT RESULTS

STATISTICAL COMPARISON OF AREA UNDER TENSION CURVES

The numbers in the maximum area column represent the magnitude of the area (in lbs. force) under the tension curves in Figure 21, relative to each other. All values are significant at $P < 0.01$

	Maximum Area (lbs.)	Standard Deviation	% of Control	% Reduction from Control	Statistical Results: d.f.= 98 t value compared between		
					Control	Plastic	Aluminum
Control	4967.08	63.80	100	0			
Plastic	5001.02	55.06	100.7	-0.7	2.848		
Aluminum	2899.67	28.57	58.4	41.6	209.1	239.5	
Graphite	2934.12	7.32	59.1	40.9	223.8	263.1	8.26

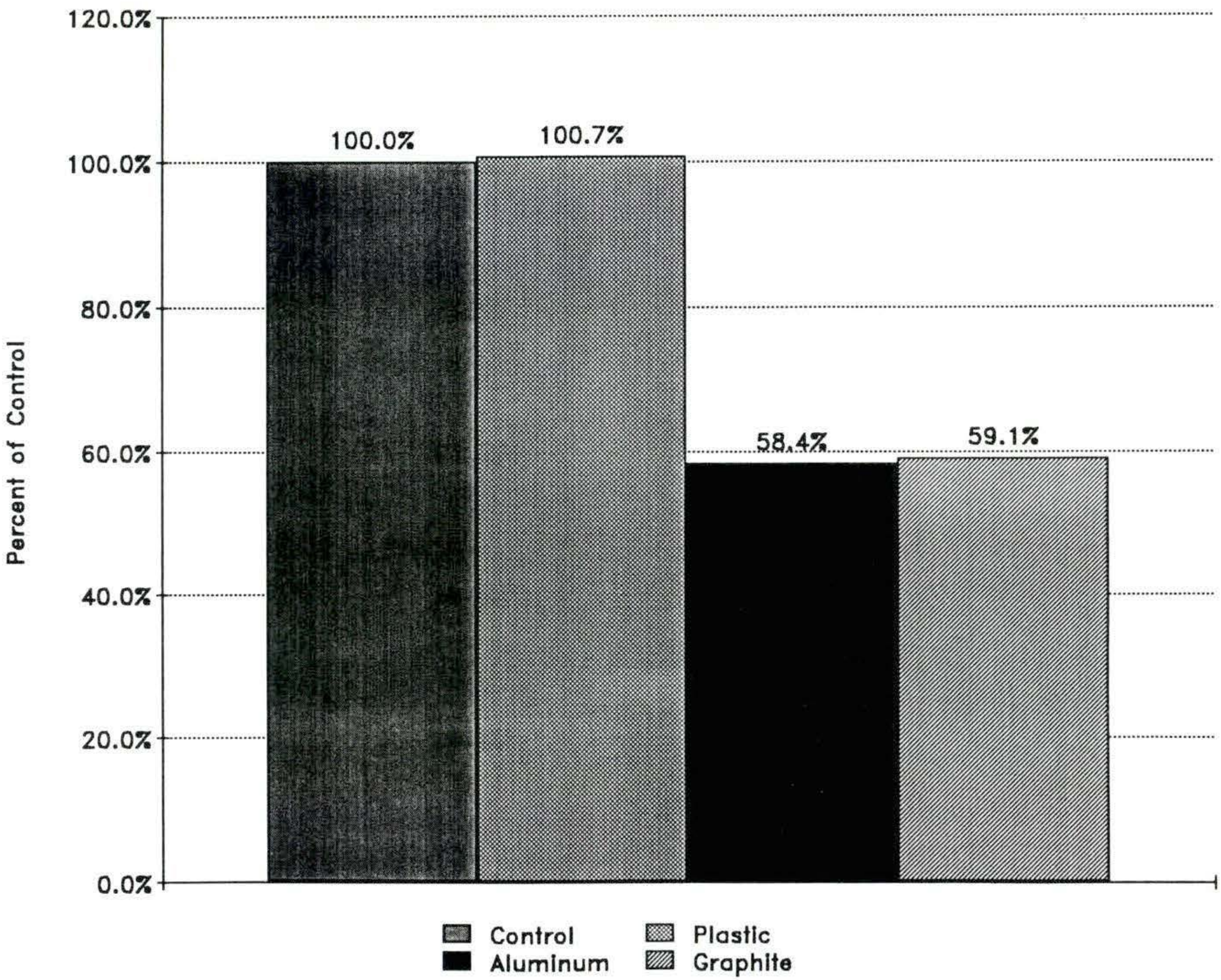

Figure 22. Low impact momentum percent of control tension

High Impact Force

The pendulum was released from sixty degrees of arc, as it was for the low impact force, again providing a velocity at impact of 2.25 m/s. The weight of the pendulum was increased from 6.68 kg (low impact case) to 16.9 kg. Momentum on impact was therefore increased to 38.0 kg*m/s. Once again, the momentum remained the same from trial to trial and case to case. However, the total force varied due to the changing duration of impact.

From Figure 23, the durations of impact for each case are: 85 ms for control, 81 ms for plastic, 62 ms for aluminum, and 54 ms for graphite. The summation of forces for each is thus 447 N for control, 469 N for plastic, 613 N for aluminum, and 704 N for graphite.

Tension

Only tension data was gathered under the high impact force. The greater impact momentum of these trials created tension in the steel cable, which, if it had been on the actual MCL, could have done some damage according to Brown, et al. (1986), and Paulos, et al. (1987), regarding the load limits of the MCL. For this reason, the higher impact force is probably more important than the low force in accessing the safety benefits of the braces.

Table 5. HIGH MASS/MOMENTUM IMPACT RESULTS

STATISTICAL COMPARISON OF TENSION MEASUREMENTS

Maximum tension is that developed in the steel cable under this impact momentum (the peak of the curves in Figure 23). The cable represents the MCL on the test leg. All values were significant at $P < 0.01$

	Maximum Tension	Standard Deviation	% of Control	% Reduction from Control	Statistical Results: d.f. = 98 t value compared between		
					Control	Plastic	Aluminum
Control	1333.7 N 299.84 lbs.	25.6 N 5.76 lbs.	100	0			
Plastic	1452.6 N 326.57 lbs.	14.8 N 3.32 lbs.	108.9	-8.9	28.46		
Aluminum	1110.7 N 249.70 lbs.	35.0 N 7.87 lbs.	83.3	16.7	36.36	63.65	
Graphite	1193.2 N 268.26 lbs.	18.6 N 4.18 lbs.	89.5	10.5	31.40	77.31	14.73

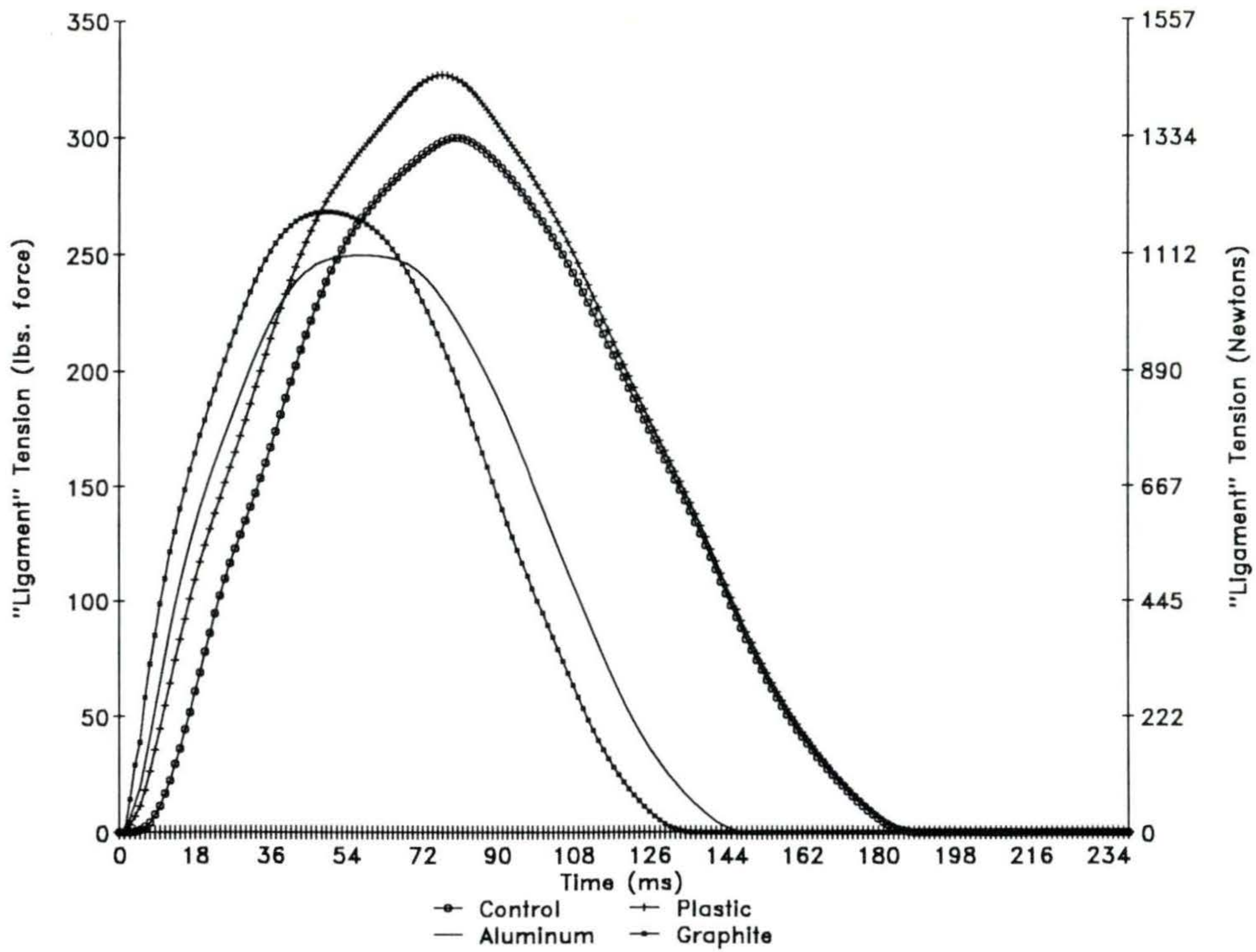

Figure 23. High impact momentum mean tension curves

Control case   The average maximum tension was 1333.7 N (299.84 lbs.) with a standard deviation of 25.6 N (5.76 lbs.). This amounted to 1.9% variation around the maximum. See Table 5 and Figure 23.

Plastic brace   This case peaked at 1452.6 N (326.57 lbs.) and a standard deviation of 14.8 N (3.32 lbs.). The standard deviation was 1.0% of the maximum tension. The change from the control case was -8.9%, significant at $p < 0.01$. Possible reasons for the negative change will be discussed later.

Table 6 lists the differences in area under the curves of Figure 23. Figure 24 shows the differences graphically. The plastic case had 11.8% more area than did the control case. The standard deviation of the plastic area was just 0.8% of the maximum. The increase in area from the control area was significant at $p < 0.01$.

Aluminum brace   The aluminum brace limited the tension to 1110.7 N (249.70 lbs.), 16.7% less than the control case. The standard deviation was the highest of any case at 35.0 N (7.87 lbs.) which was 3.2% of the maximum tension. The reduction from the plastic case was 23.5%. The decrease in area from both the control and from the plastic cases was significant at $p < 0.01$.

Table 6. HIGH MASS/MOMENTUM IMPACT RESULTS

STATISTICAL COMPARISON OF AREA UNDER TENSION CURVES

The numbers in the maximum area column are magnitudes (in lbs. force), relative to each other, of the area under the tension curves in Figure 23. Figure 24 shows the column containing "% of Control" in a graphical manner. All comparisons here are significant at P < 0.01

	Maximum Area (lbs.)	Standard Deviation	% of Control	% Reduction from Control	Statistical Results: d.f.= 98 t value compared between		
					Control	Plastic	Aluminum
Control	24496.7	537.48	100	0			
Plastic	27381.1	214.64	111.8	-11.8	35.24		
Aluminum	17731.9	449.18	72.4	27.6	68.29	137.06	
Graphite	17157.6	248.58	70.0	30.0	87.64	220.10	7.91

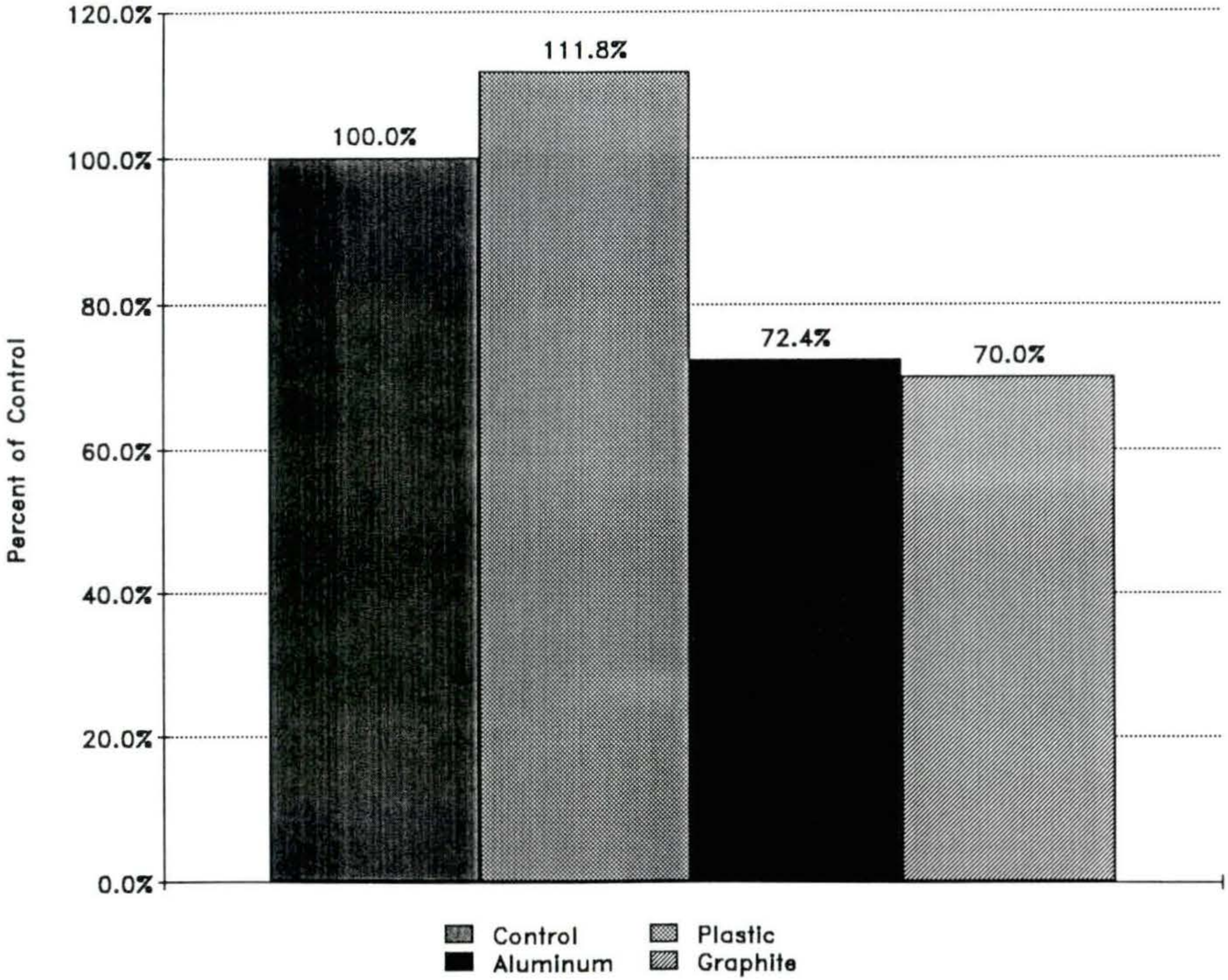

Figure 24. High impact momentum percent of control tension

The area under the aluminum case tension curve had a standard deviation of 449.2 lbs., a variation of 2.5%. This was 27.6% less than the control and 35.2% less than the plastic case. Figure 24 shows the areas as a percentage of the control area.

Graphite brace The graphite brace allowed an average maximum tension of 1193.2 N (268.26 lbs.) with a standard deviation of 18.6 N (4.18 lbs.). This was a 1.6% range around the maximum. A reduction in tension of 10.5% from the control case was observed while allowing 7.4% more force on the cable ligament than did the aluminum brace and 17.9% less force than the plastic braced case. All of the differences were significant at $p < 0.01$.

The area of the graphite tension curve was 30.0% less than the control case with a standard deviation that varied 1.4% around the maximum. The area was 3.2% less than the aluminum and 37.3% less than the plastic cases. All of the differences were significant at $p < 0.01$. It should be noted that while the peak value of the graphite tension measurement was more than that of the aluminum, the area under the graphite curve was less. This indicated less total force was absorbed by the cable ligament even though the maximum momentary force was greater.

DISCUSSION

The initial hypothesis of this research was to determine if the material properties of prophylactic brace uprights influenced the level of protection provided by the brace. Common intuition suggests that a stronger and stiffer material would, in fact, protect the knee to a greater degree. An extensive search of the literature, however, failed to find previous research that would support this hypothesis. The results of the present research showed a marked reduction in force transference to the knee due to the increased strength of the brace materials. These results suggest a similar reduction might be expected on an actual living subject. However, further study with human subjects would be necessary to confirm these results in vivo.

Past research has indicated the importance of brace mechanical and material properties, particularly relating to the factors of force distribution, absorption, and transmission (Paulos, et al., 1987). The present research investigated the singular importance of the material properties of brace uprights. Although rigidity was considered an important material property and brace/joint line clearance an important design feature, neither has been proven as a safety factor in ligament injury. The present research investigated rigidity as one primary factor in preventing ligament injury. Unfortunately, due to the design of the Ampro Knee Guard, joint line clearance became a contributing factor in some of the results.

It was found that the brace came into contact with the joint in most every case reported here, with the exception of the aluminum and graphite cases under the low impact force. The brace design allows only a small space between the joint and the hinge, such that only the most rigid of materials can resist joint contact - even under low impact conditions. The tissue alone, under the cuffs of the brace, compresses and deforms the necessary amount bringing the hinge into contact with the bony joint surface. This contact could account for the fact that the plastic brace case exhibited higher maximum tension and area under the curve than the control case in the high impact force trials (Tables 5 and 6). Both Paulos, et al. (1987), and France, et al. (1987), pointed out that joint line contact may concentrate forces on the knee that are normally distributed along the brace, actually increasing the damage due to a three point bending effect created by the brace. Another possible cause of the plastic case exhibiting higher tension than the control was the increase in force due to the decrease in duration of impact.

As stated previously, the momentum of the pendulum was the same from case to case, but the force impulse applied was not due to the changing duration of impact. France, et al. (1987), based their comparisons of braces on an Impact Safety Factor (ISF) which dealt only with impact momentum (M = mass x velocity), and not the force impulse ($\Sigma F = \Delta M / \Delta \underline{t}$; F-force, M-momentum, $\underline{t}$-time). Depending on the length of impact, the force imparted to the brace/leg combination can vary widely. In the present research, impact forces were seen to more than double under the low impact conditions between the control case and both aluminum and graphite. The high impact condition varied from 447 N in the control case to 704 N in the graphite case, not quite doubling but still a significant increase.

In Tables 1-6, the values for "% of control" were calculated on the basis of equal momentum, dividing the maximum displacement or tension value for each case by the maximum of the control case. This result is then a reciprocal of France's ISF (they divided the control by each case). If the "% of control" values in the six tables are recalculated using the variable force on each (as was done with the variable momentum for France's ISF), the resulting values show even greater apparent protection than first reported in Tables 1-6 (see Table 7). The percentages of the control case, as

Table 7. RESULTS USING VARIABLE FORCE OF IMPACT

ADJUSTED VALUES FOR '% of Control' FROM TABLES 1-6

The upper, shaded rows of each case are the values from tables 1-6. They were calculated on the basis of equal momentum and did not take into account the variable force of impact. The shaded values are equivalent to the reciprocal of France's ISF. The unshaded values are a more accurate picture of each different braces effect on the test leg under the given conditions. The following was used to calculate the new percentages:

$$\% = \frac{\text{case value from table \#x / force of impact for that case}}{\text{control value from table \#x / force of impact for control}}$$

	Low Mass/Momentum Impact				High Mass/Momentum Impact	
	Displacement, % of control from:		Tension, % of control from:		Tension, % of control from:	
	table #1	table #2	table #3	table #4	table #5	table #6
Control (shaded)	100	100	100	100	100	100
Control	100	100	100	100	100	100
Plastic (shaded)	96.4	93.6	99.9	100.7	108.9	111.8
Plastic	80.6	78.3	83.6	84.3	103.8	106.5
Aluminum (shaded)	80.6	65.1	81.8	58.4	83.3	72.4
Aluminum	39.4	31.9	40.1	28.6	60.7	52.8
Graphite (shaded)	79.6	63.0	82.8	59.1	89.5	70.0
Graphite	38.9	30.9	40.6	29.0	56.8	44.5

reported in Table 7, are more representative than those of Tables 1-6 because Table 7 takes into account the variable force, which France, et al. (1987), did not report in their research.

A notable feature of this project was the relatively low percentages of data variation within each case. The high precision of the data enabled differences of as little as 0.7% to be statistically significant. The case with the greatest percent variation, aluminum in the high impact force case, can be explained due to the material itself. Aluminum is a malleable metal that is strong and lightweight, but was easily bent by the force of the impacts. A commercially available brace would most likely not use aluminum due to this bendable nature.

Although the particular structure of the graphite uprights used was quite simple, they performed as well if not better than the aluminum. The graphite exhibited less data variation and no material deformation due to its resilient properties. Graphite's ability to bend and "remember" its original shape lends itself to impact type situations. If more construction detail were given to the lay-up pattern of the layers of graphite, certainly a much stronger and lighter upright could be made than was used in this research. Probably other, more exotic, high tech fibers or compounds could be used to improve on the performance of graphite. More research into manufacture is needed to establish which material is best concerning properties of resiliency, rigidity, and lightweight.

Future Research

Several, possibly important, design changes and features were hypothesized during the course of this research project. These changes all seem to have the capability for improving the function of the Ampro brace and could be beneficial to most prophylactic braces. Further research is needed to test the following:

(a) Braces seem to have developed into a streamlined shape, one that hugs the leg and knee as closely as possible. Why? Granted, minimum brace width is necessary on the medial side to prevent the braces from touching each other, but is it needed on the lateral side? It appears that more joint line clearance is important to prevent the three point bending effect as described by Paulos, et al. (1987) and France, et al. (1987). Instead of one centimeter between the knee and the hinge on the lateral side, future braces should have more joint line clearance, maybe up to five or more centimeters.

(b) The brace uprights are typically narrow and thin and are designed to resist tensile and compressive forces.

This is a necessary feature, but they must also resist bending forces to prevent joint line contact. The design of the lateral side upright could be wider and thicker. A truss type structure or an arch and bowstring structure would bend less and transmit load (to the distal end of the tibia and proximal end of the femur) better than a flat bar.

(c) Soft tissue containment is important to how well a brace performs. Baker, et al. (1987), found rigid soft tissue containment enhanced the safety of the MCL. The cuff connecting the two uprights acts like the center layers in plywood or the center segment of an "I" beam. The strength to resist bending is greatly improved by these "filler" materials. If the glue between the sheets of plywood or the weld of the I-beam is weak, the whole structure is weakened. The same can be said about the brace. If the cuff is too flexible in connecting the uprights, strength benefits of the double upright brace over a single upright brace are lost. The cuff could be combined with the thigh pad already worn as protection by football players. This pad is more rigid than the co-polymer cuff of the Ampro brace and it is already a necessary piece of the football player's equipment.

(d) The fact that impact duration affects the total force on the knee can be exploited by extending the length of the impact. Short duration impacts result in greater forces on the knee than do long duration impacts of the same momentum. The effect is similar to the stunt person jumping from a building and landing on an air bag. If the individual is able to decelerate through the ten foot thickness of the bag and not have an instantaneous impact with the concrete, no harm occurs. If the brace/leg could be padded the duration of impact would be increased, decreasing the force imparted. Shoulder pads and helmets were mandated to protect the user and are padded on the inside. Pad them on the outside, also, to protect the other players who are to be tackled. The ability to tackle, to block, and to generally be aggressive (as most players would like to appear to their opponents) would not be impaired. The ability, however, to inflict physical injury would be less as the brace and also the player would be padded, both of which would lengthen the duration of impact and thereby lesson the force imparted to the knee.

CONCLUSIONS

The present study determined if prophylactic knee braces could better protect the MCL if the brace uprights were constructed of a stiffer material. Paulos, et al. (1987), noted the importance of brace rigidity in distributing the impact force away from the knee and also that most current braces were less than half as rigid as the knee itself.

The data collected support this hypothesis. Stiffer materials, those which can better resist bending, provided more protection to the MCL in this test situation. Further evaluation should be done to see if this conclusion can also be reached for a human subject outside the lab situation. Controlling the duration of the impact by manipulating the materials and design of the next generation of prophylactic brace was also shown to be of critical importance. Both increased bending strength and increased impact duration will protect the knees of tomorrow's professional and amateur athletes.

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APPENDIX A

This program controlled the collection of data from the Keithley A/D converter.

```
                  'A/D Data Acquisition Program

                        '(in QuickBasic)

CLS
DEFINT T,D,F,G        'Restricts array variables to integers
DEF SEG = &HCFF8      'Specifies hex # CFF8 as current segment
                         'of memory for Keithley System 570

INPUT "Number of trials you wish to run, up to 100?"; U
INPUT "Number of samples per trial, at 830 samples/sec.?"; v

DIM T(U+5,v), D(U+5,v), F(U+5,v), G(U+5,v)

           'Defines two, two dimensional arrays for cable
           'tension, T, and knee displacement, D; each having
           'U+5 columns for up to U+5 trials and v rows for
           'up to v individual samples.

POKE 1,6       'Memory address CFF81; 6 selects the A/D
                  'converter
POKE 26,0      'Memory address CFF98 (CFF80 + 26); 0 selects
                  'x1 voltage gain
POKE 10,0      'Memory address CFF8A; 0 selects channel zero

FOR R = 1 TO u        'Loop for T number of trials, r represents
                      'the column in each array

PRINT "Ready for trial #"; R
READY: POKE 24,0 : IF (PEEK(2) + 256) * (PEEK(3) - 240) > 1025
       THEN GOTO AtoD ELSE GOTO READY

    'POKE command signals an A to D conversion and then it
    'checks photo cell gate (PEEKs) to start A/D conversion
    'loop

  AtoD:
    FOR C = 1 TO v        'Loop for number of samples per trial
```

```
      POKE 10,2           'Selects channel 2
      POKE 24,0           'Starts new A/D conversion

      T(r,c) = (PEEK(3) - 240) * (PEEK(2) + 256)

                          'Assigns value in channel 2 to T(r,c)

      POKE 10,3           'Selects channel 3
      POKE 24,0           'Starts new A/D conversion

      D(r,c) = (PEEK(3) - 240) * (PEEK(2) + 256)

                          'Assigns value in channel 3 to T(r,c)

      POKE 10,1           'Selects channel 1
      POKE 24,0           'Starts new A/D conversion
    NEXT
  POKE 10,0             'Selects channel 0
  GOSUB DELAY
NEXT R

   'Next we will filter the arrays of data

GOSUB 20Hzfilter
GOSUB 59Hzfilter
GOSUB 118Hzfilter
GOSUB 138Hzfilter

   'Next we will store the collected data in an output file

OPEN "a:Tensin.prn" FOR OUTPUT AS #2      'Creates a file on
                                          'drive A for tension
                                          'data
FOR R = 1 TO U
  PRINT #2, "Trial ";R          'Flags beginning of each trial
  FOR C = 1 TO v                   'with trial number
    PRINT #2, ((T(R,C)/410)*100)        'Loads each data point
                                          'in file as it
                                          'converts voltage into
                                          'force
  NEXT C
NEXT R
CLOSE #2                'Closes tension file

OPEN "a:Displa.prn" FOR OUTPUT AS #3      'Creates a file on
                                            'drive A for
                                            'displacement data
FOR R = 1 TO U
```

```
  PRINT #3, "Trial ";R          'Flags beginning of each trial
  FOR C = 1 TO v                   'with trial number
    PRINT #3, ((D(R,C)/410)*.263)   'Loads each data point in
                                      'file as it converts
                                      'voltage to displacement
  NEXT C
NEXT R
CLOSE #3               'Closes displacement file
END

DELAY:                                          'Delay loop
  TIME$ = "0"
  WHILE VAL(RIGHT$(TIME$,2)) < 5   'The 5 causes a 5 sec.
  WEND                             'delay which allows the
  RETURN                           'pendulum to be pulled
                                   'back for the next trial

20Hzfilter:
  FOR R = 1 TO U
    FOR C = 15 TO v - 14
      F(R,C) = (D(R,C-14) + D(R,C) + D(R,C+14)) / 3
      G(R,C) = (T(R,C-14) + T(R,C) + T(R,C+14)) / 3
    NEXT C
  NEXT R
  FOR R = 1 TO U
    FOR C = 15 TO v - 14
      D(R,C) = F(R,C)
      T(R,C) = G(R,C)
    NEXT C
  NEXT R
RETURN

59Hzfilter:
  FOR R = 1 TO U
    FOR C = 7 TO v - 6
      F(R,C) = (D(R,C-6) + D(R,C-4) + D(R,C-2) + D(R,C) +
                     D(R,C+2) + D(R,C+4) +D(R,C+6)) / 7
      G(R,C) = (T(R,C-6) + T(R,C-4) + T(R,C-2) + T(R,C) +
                     T(R,C+2) + T(R,C+4) +T(R,C+6)) / 7
    NEXT C
  NEXT R
  FOR R = 1 TO U
    FOR C = 7 TO v - 6
      D(R,C) = F(R,C)
      T(R,C) = G(R,C)
    NEXT C
  NEXT R
RETURN
```

```
118Hzfilter:
  FOR R = 1 TO U
    FOR C = 4 TO v - 3
      F(R,C) = (D(R,C-3) + D(R,C-2) + D(R,C-1) + D(R,C) +
                      D(R,C+1) + D(R,C+2) +D(R,C+3)) / 7
      G(R,C) = (T(R,C-3) + T(R,C-2) + T(R,C-1) + T(R,C) +
                      T(R,C+1) + T(R,C+2) +T(R,C+3)) / 7
    NEXT C
  NEXT R
  FOR R = 1 TO U
    FOR C = 4 TO v - 3
      D(R,C) = F(R,C)
      T(R,C) = G(R,C)
    NEXT C
  NEXT R
RETURN

138Hzfilter:
  FOR R = 1 TO U
    FOR C = 3 TO v - 2
      F(R,C) = (D(R,C-2) + D(R,C) + D(R,C+2)) / 3
      G(R,C) = (T(R,C-2) + T(R,C) + T(R,C+2)) / 3
    NEXT C
  NEXT R
  FOR R = 1 TO U
    FOR C = 3 TO v - 2
      D(R,C) = F(R,C)
      T(R,C) = G(R,C)
    NEXT C
  NEXT R
RETURN
```