

FATIGUE CHARACTERISTICS OF
KNEE LIGAMENTS

By

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CHAPTER I

INTRODUCTION

The knee joint, unlike the hip, is inherently unstable. The geometry of the knee alone cannot provide sufficient constraint of motion and the joint instead depends on a complex ligamentous structure for stability. The extra-articular ligaments, the medial and lateral collaterals, restrict varus-valgus motions and external tibial rotations (47). The intra-articular ligaments, the anterior and posterior cruciates, prohibit anterior-posterior tibial displacements. Restriction of medial and lateral tibial displacements and internal rotations are coupled functions of the collaterals and cruciates (13). If any of the ligaments are damaged, the result is instability. The degree of instability depends on the severity of damage and the number of structures involved.

Gross instability of the knee joint can lead to a osteo-arthritic condition. It is advantageous under many circumstances to surgically repair damaged or ruptured ligaments, thereby restoring stability. Acute surgical repairs, provided damage is not too severe, are often successful. However, with severe damage or chronic conditions, surgical repair is frequently unsuccessful. In such

cases, ligament replacement or augmentation is an attractive alternative.

Fresh autografts for ligament replacement have been used with some success for many years. Unfortunately, obtaining a useful autograft is not always possible or desirable. Frequently, the autograft is unable to perform mechanically and the structure replaced subsequently fails. Fresh allografts have also been used, but with some difficulty. Major problems include logistics, mechanics, storage and immunological reactions. Chemically preserved allografts have been used to repair or replace ligaments with only moderate success. Again logistic, mechanical and tissue reaction problems plague this technique. As such, prosthetic replacement of ligaments continues to have great appeal. If successful, a prosthetic replacement eliminates the problems associated with most graft-materials.

Two avenues of research have dominated prosthetic replacement. The first utilizes a scaffold replacement approach that allows the ingrowth of new collagenous tissue. This technique provides only temporary mechanical integrity until new tissue can assume the mechanical function. Working along these lines, Jenkins (28, 29, 30) demonstrated that ligaments can be replaced by filamentous carbon implants. New tissue grows and aligns as the carbon-fibre scaffold gradually fractures and degrades mechanically.

The second utilizes the concept of permanent replacement of the damaged ligament with a suitable ligament

implant. Such a replacement must be compatible. It must have sufficient mechanical strength with some promise of surviving the millions of fatigue cycles associated with normal ligament use. Considerable interests now being shown in the development and use of prostheses for ligaments, specially the cruciate ligaments of the knee. Several ligament-implant systems are being marketed, but no adequate data exist on their strength characteristics.

The present investigation is directed towards the study of one such ligament prosthesis. However, only the fatigue behavior of the ligament implant will be studied here. The design of the experiment and the procedural steps followed in performing the fatigue tests are detailed in Chapter VI. The results as obtained in the tests are discussed in Chapter VII, and finally, Chapter VIII gives conclusions and recommendations based on the test results.

Chapter II describes the anatomy of various ligamentous structures of the knee joint. A brief literary survey on the motion of the normal knee joint and its ligamentous function is given in Chapter III. Mechanical properties of the anterior and posterior cruciate ligaments are reviewed in Chapter IV. Also, a short discussion of various ligamentous implants now available is presented in Chapter V.

CHAPTER II

ANATOMICAL TERMINOLOGY OF THE KNEE JOINT

The knee is the largest of human joints in articular cartilage area and of synovial membrane. It is the most complicated in terms of internal components, and in terms of mechanics; flexion is successively a combination of rotation, rocking and gliding movements far removed from the concept of a simple hinge. It occurs between the large rounded condyles of the femur and the flattened condyles of the tibia. Each of the tibial condyles is deepened by a meniscus. In addition, there is an articulation between the patella, a sesamoid bone within the tendon of quadriceps femoris, and the femur.

A strong capsule surrounds the joint and blends with the ligaments and tendons. Superiorly, it is attached to the anterior surface of the femur on either side of the quadriceps femoris tendon, to the femoral condyles, and to the intercondylar fossa of the femur. Inferiorly, the capsule attaches to the tibia but not to the fibula, and it extends down to the tibial tuberosity. Anteriorly, the capsule is formed by the tendon of quadriceps femoris which is attached to the patella, and by the ligamentum patellae.

Posteriorly, there is an opening in the capsule for the tendon of the popliteus, which actually penetrates the capsule.

The supporting structures of the knee are divided into two major groups: (1) the static stabilizers (ligaments), which are further divided into two layers, capsular and non-capsular ligaments, and (2) the dynamic stabilizers (musculo-tendinous units and their aponeuroses). The supporting structures are further divided into those of the medial compartment (Figure 1), considered to extend medially from the patellar tendon to the posterior cruciate ligament, and those of the lateral compartment, considered to extend laterally to the posterior cruciate ligament. In each compartment there is a capsular ligament intimately attached to the meniscus and divided into meniscomfemoral and meniscotibial portions (Figure 2). The anterior and middle thirds of the capsular ligaments in the two compartments are similar. In the anterior one-third the ligaments are thin, loose, and covered superficially by the extensor retinaculum of the quadriceps mechanism which functions as a dynamic stabilizing aponeurosis. In the middle one-third the capsular ligaments are strong and supported superficially by the tibial collateral ligament medially and the iliotibial band laterally. The middle one-third of the lateral capsular ligament is continuous with the fat pad and it is, for this reason, that it has been misinterpreted

Figure 1. Right Tibia Showing the Capsular Ligament and Its Various Thickenings or Strong Points. (1) Medial Capsular Ligament, Middle Third; (2) Lateral Capsular Ligament, Middle Third; (3) Posterior Oblique Ligament; (4) Arcuate Ligament; (5) Tibial Collateral Ligament; and (6) Iliotibial Band.

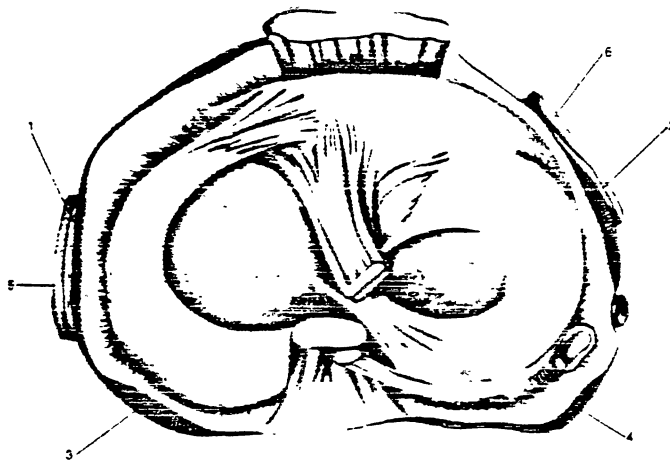
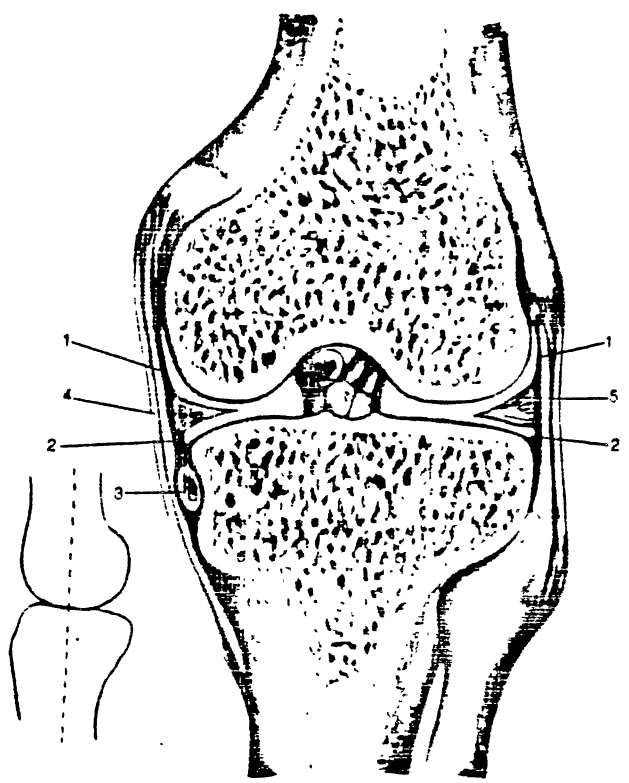


Figure 2. Medial and Lateral Meniscomfemoral and Meniscomtibial Ligaments. (1) Meniscomfemoral Ligament; (2) Meniscomtibial Ligament, Medial and Lateral; (3) Anterior Tendon of Semi-membranosus Muscle; (4) Tibial Collateral Ligament; and (5) Fibular Collateral Ligament.



as being an insignificant structure anatomically and functionally.

In the posterior thirds of the two compartments, the capsular ligaments differ. The thickened posterior third of the medial capsular ligament is termed the posterior oblique ligament. Its supporting function is augmented by the dynamic stabilizing effect of the capsular arm of the semimembranosus tendon and its aponeurosis, the oblique popliteal ligament (Figure 3).

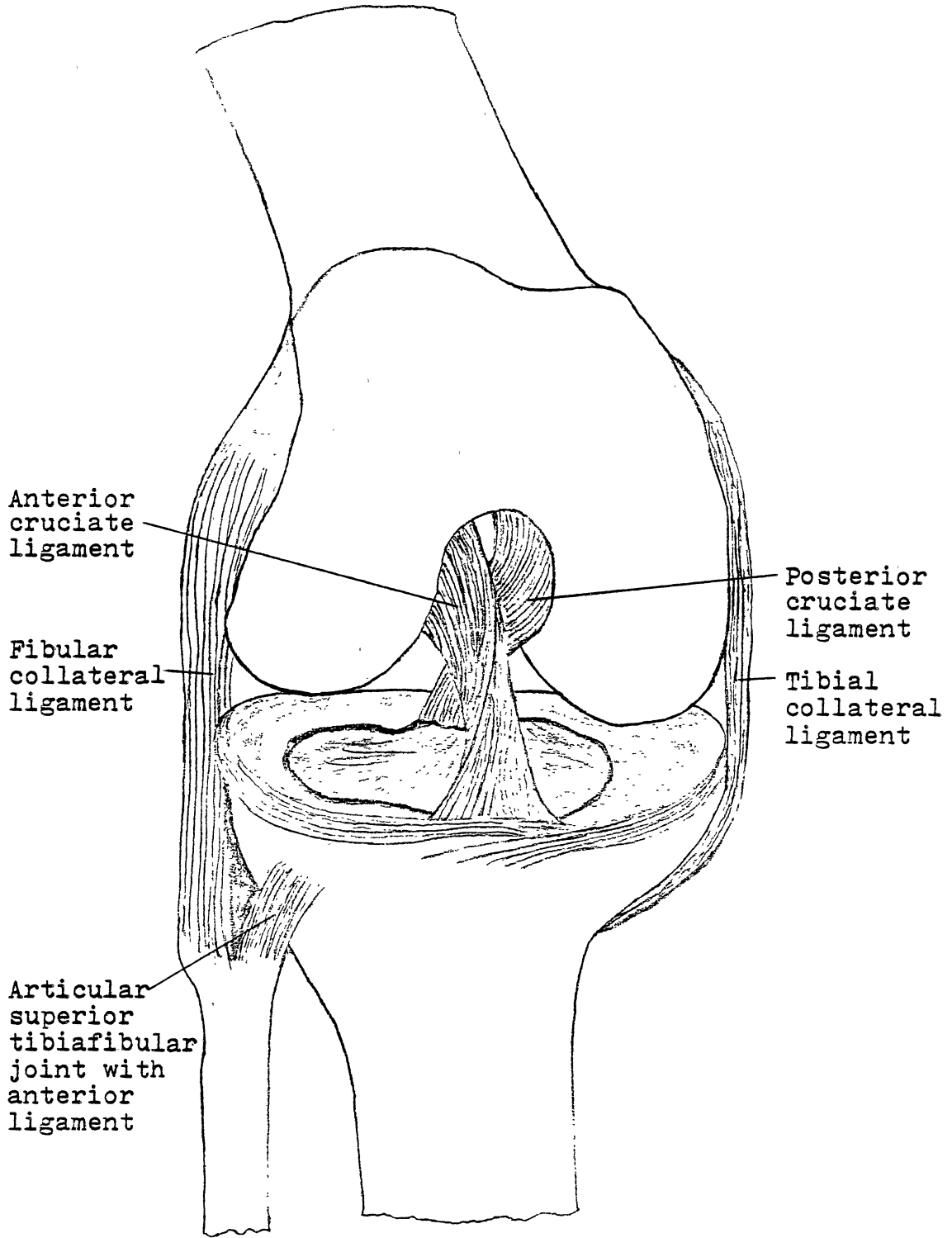
In a somewhat different arrangement, the posterior third of the lateral capsular ligament is composed of the fibular collateral ligament, the arcuate ligament, and the aponeurosis of the popliteus muscle. The supporting function of this arcuate complex is augmented by the dynamic effects of the biceps femoris and popliteus muscles. The medial and lateral heads of the gastrocnemius also support their respective compartments dynamically.

The central ligaments lying within the capsule are the anterior and posterior cruciates (Figure 4). The anterior cruciate ligament extends from the lateral femoral condyle to the tibial surface in front of the medial tibial tubercle. When the knee is flexed 90 degrees this ligament is oriented almost parallel to the tibial plateau. The posterior cruciate ligament, attached posteriorly on the tibia, has a fan-shaped line of attachment on the medial femoral condyle which resembles the line of the plotted instant-centers of rotation. When the knee is in full extension

Figure 3. Medial-posterior Aspect of the Right Knee. The Oblique Popliteal Ligament (1), Originates From the Capsular Arm of the Semimembranosus Muscle (2), Which Sends Supporting Fibers to the Posterior Oblique Ligament (3). The Tibial Arm of the Semimembranosus Tendon Has Several Branches, Including the Fibrous Covering (5) Which Spreads out Over the Popliteus Muscle (6). The Popliteal Recess in the Posterolateral Capsule (7) is Bound Laterally by the Tendon of the Plantaris (8) and Medially by the Posterior Capsule. The Medial (9) and Lateral (10) Heads of the Gastrocnemius Give Further Dynamic and Fibrous-tissue Support to the Posterior Aspect of the Knee.



Figure 4. Central Cruciate Ligaments and Collateral Ligaments of the Human Knee Joint.



(in the standing position), the posterior cruciate ligament forms an angle of about 30 degrees with the horizontal.

The anatomical features as described above are well accepted and point out the complexity of the ligamentous structure of the knee.

Closely attached to the joint surfaces are the articular cartilage which provides a wear resistant, low friction lubricated surface. The cartilage is slightly compressible and elastic and ideally constructed for ease of movements over a similar surface, but able to accommodate the relatively enormous compressive and shear forces generated during the muscle action. Young and healthy cartilage looks white, smooth, and glistening to the naked eye. Aging cartilage is thinner, less cellular, firmer, more brittle, has a less regular surface, and appears yellowish in tint.

A pale yellow, viscous, glaring fluid called synovial fluid is found in the cavities of the knee joint. The physical properties of synovial fluid show viscous, elastic and plastic components. The function of the synovial fluid is to provide a nutritive source for the articular cartilage, discs, and menisci, and to provide the necessary joint lubrication.

The menisci (medial meniscus and the lateral meniscus-semilunar cartilage) are two crescentric lamellae which deepen the surfaces of the upper end of the tibia in articulation with the femoral condyles. The peripheral attached

border of each meniscus is thick and convex, the free border is thin and concave. The upper surface of the menisci is smooth, concave, and in contact with the condyle of the femur. The lower surface is smooth, flat and rests upon the tibia.

CHAPTER III

MOTION OF THE NORMAL KNEE JOINT AND IT'S LIGAMENOUS FUNCTION

Starting from full extension to full flexion, the knee joint is believed to have in general, three kinds of motion which are:

1. Rolling motion, where equidistant points of the femur contact equidistant points of the tibia or vice versa.
2. Sliding motion, where the contact of the tibia with the femur is a localized area or a point which sweeps over the whole contour of the other part.
3. Terminal rotational motion, where the femur rotates internally on the tibia during the terminal phases of extension.

While analyzing the motion of the knee joint, we need to consider three perpendicular planes, viz; the transverse, the frontal, and the sagittal plane. For reference, these planes are illustrated in Appendix B.

A. Axis of Movement of the Knee Joint

Knee flexion in the sagittal plane does not represent a continuously centered hinge motion as, e.g., the elbow joint. In fact, the various centers of curvature of the

femoral condyle can be easily determined. They proceed along a line that shows an increase in length of the radii of curvature in the direction of extension. The centers of the curvature of the femoral condyle, however, do not correspond to the axis of motion of the knee joint. The axis of motion can be easily traced if the knee joint is compared to a mechanical system, a so-called closed planar, kinematic, four-jointed chain or four-bar linkage (22). Such a classification carries two prerequisites:

1. The motion must be carried out in only one plane; this applies to the knee with the exception of terminal rotation.

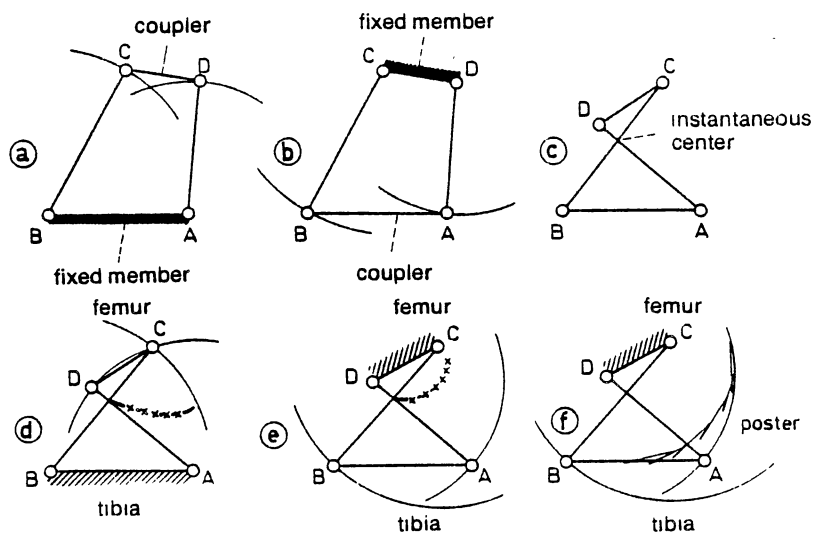
2. The joints of the four-jointed chain must be rigidly connected. This applies to the system; cruciate ligaments, tibia, and femur inasmuch as in all positions some of the fibers in both cruciate ligaments are under tension.

Four-Jointed Chain System

The closed four-jointed chain has, as suggested, four joints connected by four rigid segments. By definition, the planar, closed four-joint chain moves in one plane, and these movements can easily be demonstrated on drawing paper (Figure 5).

They are defined by a change of one segment, whereby one of them (the supporting link) maintains its position while the other three segments change position. The end point of the supporting link represents the centers of

Figure 5. Closed Four-Jointed Chain System. (a) ABCD is a Closed Four-bar Kinematic Chain, Which Moves in the Plane of the Paper. If AB is the Fixed Member, All Points on AD and BC Move on Circular Arcs About A and B Respectively. Points on CD, the Coupler, Other Than C and D, Move on Coupler Curves Which are Not Circular Arcs; (b) Each Member of the Four-bar Chain Can be the Fixed Member; (c) This Crossed, Closed, Four-bar Chain Corresponds to the Anatomic Structure of the Cruciate Ligaments (AB the Intercondylar Eminence of the Tibia, BC the Anterior Cruciate Ligament, AD the Posterior Cruciate Ligament, DC the Intercondylar Fossa of the Femur). The Axis of Motion of the Knee Passes Through the Instantaneous Center, the Crossing-Point of the Cruciate Ligaments; (d,e) Depending on Whether the Tibia or the Femur is Regarded as the Fixed Member, Different Loci of the Instantaneous Center Result; (f) When the Tibia is Moved Relative to the Fixed Femur, the Successive Positions of the Coupler Yield a Coupler Envelope Curve as Shown.



motion for all points of the adjoining members, while the points of the link opposite to the supporting link (the so called coupler) performs non-circular movements, the so-called coupler curves (Figure 5a and b).

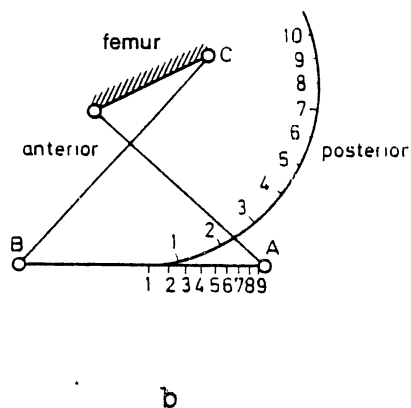
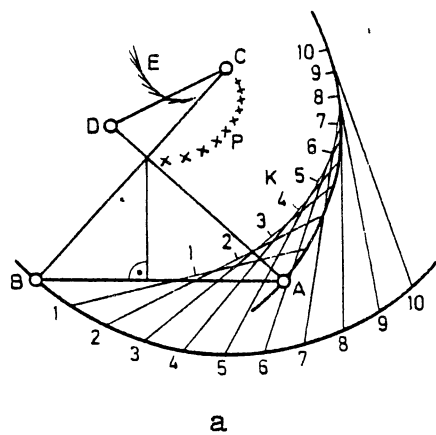
The cruciate ligament system, therefore, resembles a crossed, closed, planar four-jointed chain (Figure 5c). CD corresponds to the intercondylar fossa of the femur, AB to the tibial eminence, and BC and AD to the anterior and posterior cruciate ligaments respectively. The center of the axis of the knee joint conforms to the crossing point of link BC and AD and is called the "pole" or instantaneous center. During knee flexion, the instantaneous center moves dorsally on its pole curve. Depending on whether the femur or tibia represents the supporting segment, two different pole curves can be traced with ruler and compass (Figures 5d and e).

Another curve can be derived from the motion of the four-jointed chain. Insertion of the anterior and posterior cruciate ligaments at the tibia represent approximately the altitude curve of the joint surface of the tibial condyles. With the femur fixed, the tibial joint surface becomes the coupler. During knee flexion, the locus of the midpoint of the coupler is a curve corresponding to the contour of the femoral condyles (Figures 5f and 6a).

B. The Rolling and Sliding Motion

All points of the coupler move with an instantaneous

Figure 6. Locus of the Midpoint of the Coupler Corresponding to the Contour of the Femoral Condyles. (a) 10 Different Positions of the Coupler AB Were Drawn Between the Circular Arcs With Radii BC and AD. The Resulting Coupler Envelope Curve K Corresponds to the Contour of the Femoral Condyle. The 10 Corresponding Instantaneous Centers Produce the Pole Curve P. The Contact Points of the Coupler are Determined by the Normals Through the Appropriate Instantaneous Centers. The Continuations of These Normals Yield the Evolute (Locus of the Center of Curvature of the Femoral Condyle). (b) If the Distance Along the Coupler to the Contact Point in Each of the Different Positions of Flexion is Marked on AB, the Corresponding Points on the Femur and Tibia Result. The Fact That as Flexion Increases These Points are More Closely Grouped Posteriorly Shows the Predominance of Sliding Motion in Squating.



velocity directed perpendicularly to the radius from the instantaneous center. Since the velocity at the femoro-tibial contact point corresponds to the direction of the tibial joint surface, the point of contact must always lie perpendicularly below the instantaneous center (i.e., below the crossing of the cruciate ligaments). Therefore, by constructing a pole curve, it is possible to arrive at a geometric determination of the contact points between femur and tibia during various positions of flexion. This is done by drawing the normal from the pole to the coupler. By marking the contact points on the tibia in succession (Figure 6), the combined sliding and rolling movements typical of the knee joint can be demonstrated. A purely rolling motion never occurs. In pure rolling, the distances between two points of contact on the tibia and femur would become equal. This occurs only during the primary phases of flexion; with increasing flexion, the contact points at the tibial condyle are drawn increasingly together. If there were a purely sliding motion, only one point of contact would exist on the tibial segment.

The rolling and sliding motion can also be explained from the length of the femoral joint curve which is estimated to be 10 cm and that of the tibia, estimated to be 8 cm. Hence, the radii of the contacting surfaces are different (i.e., of the tibia more than the femur). As a result, there is neither a constant sliding, as in the case of equal radii surfaces contacting each other, nor a

constant rolling motion, as in the case of curved surface contacting more plain surface. As a result, a combination of both the rolling and sliding motion occurs.

C. Terminal Rotational Motion

From full flexion to full extension, the femoral condyles and tibial plateaus move in the sagittal plane during active motion. In the terminal stage of extension, there is also a rotational movement in the transverse plane. This rotation reaches about 10° in full extension. Firmly fixed on the ground, the foot represents internal rotation of the femur on the tibia. When free, it represents external rotation of the tibia. When the active knee stabilizers are completely relaxed, terminal rotation leads to maximum tightening of all passive stabilizers. This rotation movement is partially controlled by the ligament system and partially by the joint surfaces (22). Although the joint surfaces and capsular ligament systems function optimally together during terminal rotation, the dissected specimen shows that this phase can be maintained with light axial compression without any ligament support, even after division of the cruciate ligaments.

Terminal rotation corresponds to the direction in which the lower leg turns spontaneously after division of the capsular ligaments. When divided, the cruciate ligaments unwind from each other, lie parallel, and their full length is exposed (22). Loosening of the cruciate ligaments

during terminal rotation is necessary because of the noticeable rise of the anterior tibial plateau contact surfaces on extension, specially on the medial side (22), and the anterior elongation of the femoral radii of curvature.

D. The Function of the Ligaments

One of the principal functions of knee ligaments is to limit the motion between the tibia and femur. The ligaments do this by developing a restraining force whenever they are stretched. The total restraining force is simply the sum of the contributions from the individual ligaments, excluding such considerations as weight-bearing forces and geometric constraints.

There is a vast amount of literature which discusses the movements of the knee joint and the function of ligaments controlling these movements. These literatures reaveal no unanimity of opinion concerning the function of the knee-joint ligaments and often equivocal statements are made. As reported by Brantigon and Voshell (5), all controversial subjects are discussed at length in Fick's publication (12). Some of the fundamental concepts of the mechanics of the knee-ligaments which are generally accepted without controversy, are given below.

General Mechanical Concepts of Knee-ligaments

Both collateral ligaments are taut in complete exten-

sion. The cruciate ligaments, by twisting on themselves, prevent abnormal medial rotation of the tibia on the femur. The posterior aspect of the capsule and the oblique popliteal ligament aid in preventing hyperextension. The fibular collateral ligament is relaxed in flexion.

Backward gliding of the tibia relative to the femur is controlled by the posterior cruciate ligament, and forward gliding of the tibia is controlled by the anterior cruciate ligament. However, in hyperextension, both collateral ligaments are tight and, by forcing the tibia and femur tightly together, will reduce such motion to a minimum. Most authors (16, 18) state that adduction of the joint is prevented by the lateral collateral ligament and the cruciate ligaments; abduction by the medial collateral ligament and the cruciate ligaments. Steinder (49), however, maintains that this movement is checked entirely by the collateral ligaments. The medial collateral ligament and the cruciates, acting together, limit rotation of the femur on the tibia (i.e., about the long axis of the joint) in all positions of the joint. That some portion of the medial collateral ligament is taut in all phases of extension and flexion is evident from the fact that abnormal rotation of the joint is prominent when this ligament is cut. The lateral collateral ligament resists lateral rotation when the joint is in extension. Medial or lateral movement of the femur on the tibia is prevented by interaction between

the tibial intercondyler eminence and the femoral condyles, and the restraint of the ligaments.

As mentioned earlier, the total restraining force to limit the motion between the tibia and femur is the sum of the contributions from the individual ligaments. Thus, the relative importance of a single ligament can be assessed in terms of the percent of total restraining force it provides. Butler, Noyes, and Grood (6) reported on the ranked order of importance of each ligament and capsular structure in resisting the clinical anterior and posterior drawer tests. Their ranked importance is based on the force provided by each ligament in resisting the drawer tests. The contributions of the major ligaments and capsular structures to restraining anterior-posterior drawer were determined as follows. A baseline test to five millimeters of peak anterior and posterior tibial displacement was first conducted on the intact knee. The applied joint displacement and restraining force were continuously monitored and recorded. The contribution of the ligament to restraining drawer was taken to be the drop in restraining force that occurred between the two tests. The process was repeated until all structures had been sectioned. The ligaments and capsular structures investigated were the anterior and posterior cruciate ligaments, the medial and lateral collateral ligaments, the popliteus musculo-tendinous unit, the medial and lateral capsules, the iliotibial band, and the iliotibial tract. The average restraining forces for eleven

knees at 90 degrees of flexion and three knees at 30 degrees are shown in Table I. The effect of cutting the anterior cruciate ligament is shown by the broken line in a typical force-displacement curve (Figure 7) for anterior-posterior drawer in an intact knee joint. The restraining force is reduced greatly for anterior drawer, but is unaffected for posterior drawer. The restraining force of the cruciate ligament alone is obtained by subtracting the result of the test done after the cut from the result of the test done before the cut. Such "difference curves" were determined for all ligament structures studied. Typical force-displacement curves for the anterior and posterior cruciate ligaments are shown in Figure 8.

The concept of primary and secondary restraints for each kind of motion of the knee has also been described in the report by Butler et al. The anterior cruciate ligament was found to be the primary restraint to anterior drawer (85 percent). In contrast, other ligament structures provided only a small secondary restraint, and the posterior cruciate ligament provided no restraining action at all. In posterior drawer, the posterior cruciate ligament was found to be the overwhelming primary restraint (94 to 96 percent). Medial collateral ligament and posterior capsule were found to be important secondary posterior restraints. The secondary restraints to anterior drawer in the absence of anterior cruciate ligament are shown in Table II. Six specimens were tested with an average joint displacement

TABLE I
RESTRAINING FORCES AT FIVE MILLIMETERS OF DRAWER

Specimen	Anterior Drawer			Posterior Drawer		
	Intact Force (N)	Anterior Cruciate Force (N)	Force Carried with Anterior Cruciate Cut (Per cent)	Intact Knee (N)	Posterior Cruciate Force (N)	Force Carried with Posterior Cruciate Cut (Per cent)
90 degrees of flexion						
1	440	349	79.3	405	385	95.0
2	363	252	69.3	274	238	87.0
3	467	414	88.7	515	*	*
4	414	359	86.8	400	*	*
5	368	340	92.4	*	*	*
6	351	298	85.0	245	245	100.0
7	392	338	86.3	426	415	97.5
8	453	403	89.0	662	659	99.5
9	366	321	87.6	310	284	91.7
10	610	500	81.9	520	437	84.0
11	612	550	89.9	456	456	100.0
Mean ± S E M	439.6 ± 28.1	374.9 ± 26.3	85.1 ± 1.9	421.3 ± 39.9	389.9 ± 49.1	94.3 ± 2.2
30 degrees of flexion						
12	238	200	84.2	305	294	96.4
13	352	309	87.8	413	399	96.5
14	410	368	89.7	274	261	95.2
Mean ± S E M	333.3 ± 50.5	292.3 ± 49.2	87.2 ± 1.6	330.7 ± 42.1	318.0 ± 41.6	96.0 ± 0.4

* Posterior drawer test not performed

Figure 7. Anterior-Posterior Drawer Test. A Typical Force-Displacement Curve in an Intact Joint (Solid Line) and After Cutting the Anterior Cruciate Ligament (Broken Line). The Arrows Indicate the Direction of Motion.

Figure 8. Force-Displacement Curves for the Anterior and Posterior Cruciate Ligaments. Restraining Forces of the Anterior (Solid Line) and Posterior (Broken Line) Cruciate Ligaments During a Five-millimeter Anterior-Posterior Drawer Test on a Typical Knee Preparation. The Curves are Constructed by Taking Differences Between the Curves of the Joint Before and After Cutting the Ligament. The Anterior Cruciate Resisted Nearly all the Force Anteriorly With no Contribution Posteriorly. The Posterior Cruciate Restrained Posterior Joint Displacement But Had Minimum Effect Anteriorly.

DRAWER TEST

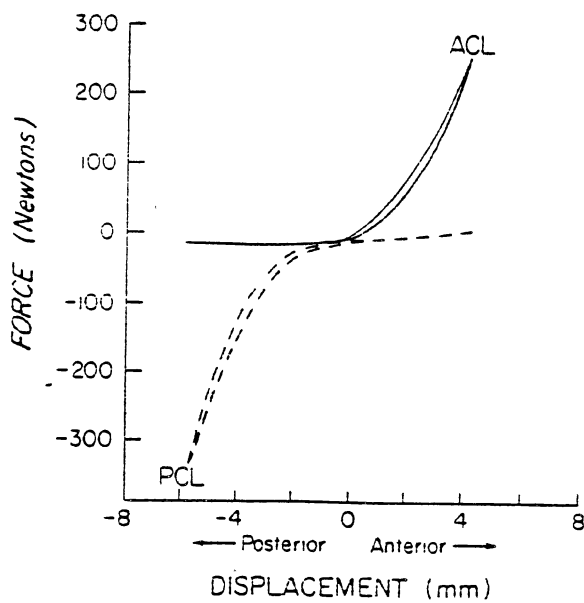
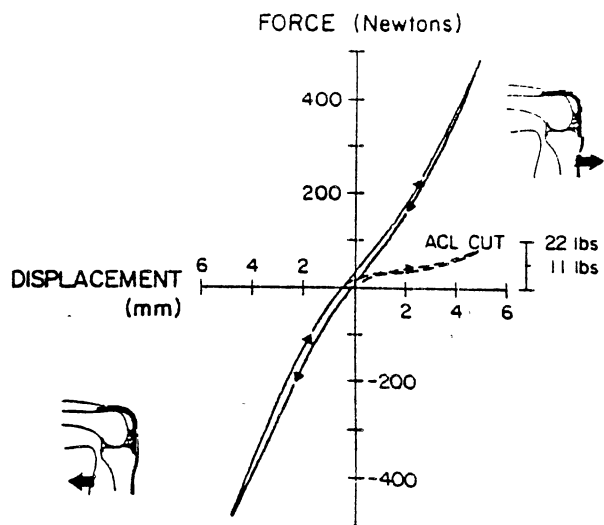


TABLE II
COMPARISON OF SECONDARY STRUCTURES AT
INCREASED ANTERIOR DISPLACEMENT

	Iliotibial Tract and Band	Mid-Medial Capsule	Mid-Lateral Capsule	Medial Collateral Ligament	Lateral Collateral Ligament
Mean \pm S.E.M. (<i>Per cent</i>)	24.8 \pm 4.7	22.3 \pm 6.9	20.8 \pm 5.4	16.3 \pm 2.9	12.4 \pm 3.3
Minimum (<i>Per cent</i>)	9.8	3.4	1.6	8.1	7.0
Maximum (<i>Per cent</i>)	44.4	43.6	36.9	29.4	25.2

* n = 6, anterior drawer of 12.2 to 16.3 millimeters at 90 degrees of knee flexion. By Duncan's multiple range test, no statistical difference was found among the percentages for any of the structures shown ($p > 0.05$).

TABLE III
COMPARISON OF SECONDARY STRUCTURES AT
INCREASED POSTERIOR DISPLACEMENT

	Posterior Lateral Capsule and Popliteus Tendon†	Medial Collateral Ligament	Posterior Medial Capsule	Lateral Collateral Ligament	Mid-Medial Capsule
Mean \pm S.E.M. (<i>Per cent</i>)	58.2 \pm 10.1	15.7 \pm 5.2	6.9 \pm 3.2	6.3 \pm 4.7	6.2 \pm 2.4
Minimum (<i>Per cent</i>)	36.7	5.6	0.0	0.0	0.0
Maximum (<i>Per cent</i>)	82.1	29.7	14.5	20.4	11.5

* n = 4, posterior drawer of 19.0 to 25.0 millimeters at 90 degrees of knee flexion.

† By Duncan's multiple range test, only the posterior lateral capsule and popliteus tendon were statistically different from all other structures ($p < 0.05$).

of fourteen millimeters (range, 12.2 to 16.3 millimeters). Table III shows the secondary restraints to posterior drawer in the absence of the posterior cruciate ligament. Four knees were tested with an average joint displacement of twenty-one millimeters (range, 19.0 to 25.0 millimeters). Both the tests were carried at 90 degrees of flexion.

The ligamentous function, as studied by Butler et al. (6) is independent of the cutting order of various ligaments because the joint displacement controls the amount of ligament stretch, and hence its force. Reproducing the displacement reproduces the force in each ligament. This means that even after a single ligament is sectioned, the remaining ligaments are unaffected; i.e., there is independence of cutting order.

Many prior studies have examined ligaments function by measuring increases in joint laxity with selective cutting of ligaments. Typically, a test is conducted by applying a force on an intact knee, cutting a ligament, repeating the test, and measuring the increase in laxity. One problem with such studies is that this increased laxity is dependent on the order in which the ligaments are cut. If the cutting order is changed, the measured laxity changes. Hence, it is not possible to define the function of a single ligament precisely.

Piziali, Seering, Nagel, and Schurman (40) adopted the similar technique as that of Butler et al. and investigated the functions of human knee ligaments during anterior,

TABLE IV

CONTRIBUTION OF THE LIGAMENTS TO FORCE ON THE FEMUR CAUSED
BY MEDIAL AND LATERAL TIBIAL DISPLACEMENTS

Flexion angle	Specimen and peak applied force	Tibial displacement	Percent force resisted by.							
			Anterior cruciate ligament	Posterior cruciate ligament	Medial collateral ligaments	Lateral structures*	Posterior structures†	Bone contact	Menisci	Other
0°	A (690 N)	4.5 mm Medial	52	1	- 5	30	2	15	5	0
	B (660 N)	4.2 mm	25	19	- 3	28	7	24	0	0
	A (650 N)	4.9 mm Lateral	7	36	22	9	21	3	1	1
	B (510 N)	4.5 mm	18	32	42	2	1	2	0	3
30°	A (690 N)	4.5 mm Medial	24	11	14	18	6	25	2	0
	B (510 N)	4.4 mm	33	14	16	11	0	23	0	3
	A (630 N)	5.1 mm Lateral	14	44	30	6	3	3	0	0
	B (470 N)	4.5 mm	13	25	50	5	0	7	0	0

*Lateral collateral ligament and popliteal tendon.

†Posterior oblique ligament and posterior capsule.

posterior, medial, and lateral tibial displacements. They went further to explore the three-dimensional load transmission characteristics of knee ligaments. Contributions of the ligaments to force on the femur, during medial and lateral tibial displacements as given in the report of Piziali et al., is shown in Table IV. The anterior cruciate ligament is seen to be the most important element in resisting medial tibial displacement while the medial collateral ligament and the posterior cruciate ligament are of primary importance in resisting lateral tibial displacement.

CHAPTER IV

MECHANICAL PROPERTIES OF THE LIGAMENTS

To replace the ligaments with an artificial ligament system requires a knowledge of ligament strength and mechanical properties. When information is available to describe the mechanical properties of the cartilage, meniscus, and ligament aspects of knee function, components can be designed so that knee reconstruction, rather than knee replacement, can be undertaken.

In recent years, many studies have been performed to establish the mechanical properties and behavior of tendons and other collagenous structures. Prior studies on ligaments have provided important information; however, as yet our knowledge of their biomechanical properties is incomplete. The unique anatomical characteristics of individual ligaments, such as the twisting of the cruciates or the intricacy of the medial ligaments of the knee, impose specific structural properties which cannot be generalized to other ligaments. In addition, there appears to be an important relation between macroscopic mechanical behavior and tissue microstructure (8, 11, 43). The mechanical behavior of ligaments depends not only on the material properties of the fibers themselves, but also on the geometrical arrange-

ment of collagen fibers and fiber bundles, the proportioning of different types of fibrous constituents, and the relatively unknown effect of the surrounding ground substance (8, 16, 26, 33). The ligament insertion site and the underlying bone are additional parts of the ligament unit which must be considered in evaluating strength properties. In the analysis of ligament properties, it is often difficult to separate the effect of these different factors on macroscopic behavior. For these reasons, a great deal of caution is required in generalizing the mechanical properties of one ligament to other ligaments.

An alternative approach to investigate each component of a ligament unit separately is afforded by using a bone-ligament-bone preparation, such as the femur-anterior cruciate-tibia preparation. The mechanical properties of all components of the unit are then studied together in the same relationship that they have in the body, so that the combined interaction determines the properties of the structure, as in-vivo.

Mechanical properties of human soft tissue have been studied previously in various ways by a number of researchers. Seering (46) has given an extensive survey of the literature. In 1931, Gratz (17) reported that "As tension is applied to the tissue an elongation takes place, at first quite pronounced. The material then stiffens and and finally relaxes again." He also noted that the tissue,

when tension was released, did not quite return to its original length.

In 1936, Cronkite (9) conducted tests on almost every tendon in the human body. He reported no significant difference between tendons in their tensile strengths which ranged from 2×10^8 to 2×10^9 dyn/cm². Gratz (17) in 1931, conducted similar tests on three tendons and obtained tensile strength within this range and Young's modulus of 3×10^8 to 4×10^8 dyn/cm².

Hardy (20), in 1951, noted the difference in modulus between ligaments containing large amounts of collagenous material and those containing elastic fibers. In 1954, Smith (48) examined the elastic properties of the anterior cruciate ligaments of rabbits. His data included evidence that these ligaments act as viscous bodies under certain conditions. When elongated up to 20 percent for short durations, the ligaments responded elastically. For ten repeated loads and for applied loads equal to the body weight of the respective rabbit, the ligaments, upon unloading, returned to their original length. But for loads near failure, the ligaments elongated and did not fully recover.

Rigby (43, 44), in 1959 and 1964, studied the mechanical properties of rat tail tendons. He found that the mechanical properties were reproducible within a strain range of 0-4 percent. Variation of temperature from 0 to 37⁰C produced no effects while an increase in strain rate produced a slight shift in the stress-strain curve toward

the stress axis. The average maximum modulus for a strain rate of 10 percent/min was from 6 to 10×10^9 dyn/cm².

During the last twenty-five years, mechanical properties of passive soft tissues have been studied by many investigators. Though these authors were pursuing different interests, they agree that ligamentous tissue is essentially elastic, but exhibits some viscoelastic characteristics. As compared with Cronkite's (9) values for ultimate tensile strength (8,700 to 18,000 psi), Wright and Rennels (57) reported a range of 3,000 to 4,000 psi and Walker, Edward, and James (53) measured strength from 10,700 to 21,000 psi.

Abrahams (1) attempted to correlate the shape of the stress-strain curve to the behavior of the collagen fibers. He found that the stress-strain curves of horse and human tendons described three distinct regions. The primary region was that of a 0-1.5 percent strain in which there was a considerable increase in length, with only a slight increase in stress. In this region, it is thought that the collagen fibers begin to straighten their wavy pattern. The secondary region was that of a 1.5-3.0 percent strain where the collagen fibers are thought to become fully oriented and begin to assume most of the load. In the final region, that of a 3.0-5.0 percent strain, it is thought that the entire response is due to the collagen fibers in pure tension. The stress-strain curve is a straight line in this region. Visible ruptures of the collagen fibers begin at a 5.0 percent strain level.

Kenedy, Hawkins, Willis, and Danylchuk (31), tested the cruciate ligaments and the tibial collateral ligaments of human knee joints in-vitro to determine yield point, ultimate failure stress, and disrruption caused by failure loading. For these tests, elements were excised from the in-vitro knee and loaded individually. Yield points and maximum loads were reported for individual ligaments. At strain rates of 12.5 cm/min, the posterior cruciate ligament exhibited a yield point of 81.3 ± 5.1 kg and a maximum load of 94.0 ± 4.8 kg. At 50 cm/min, these loads increased to 88.5 ± 6.8 kg and 107.3 ± 8.1 kg. For the anterior cruciate ligament at 12.5 cm/min, yielding occured at 40.2 ± 2.8 kg and the maximum load was 47.7 ± 3.4 kg. At 50 cm/min, the values were 51.2 ± 2.9 kg and 63.8 ± 2.3 kg. Values for the tibial collateral ligament were similar to those for the anterior cruciate ligament. At 12.5 cm/min, the yield point was 46.3 ± 3.2 kg and the maximum load was 47.7 ± 3.4 kg. At 50 cm/min, the values were 54.5 ± 3.1 kg and 67.8 ± 7.6 kg. For all ligaments, the strain at maximum load was between 20 to 30 percent.

Investigation on anterior cruciate ligament has been carried out by a number of researchers. Noyes, James, and Peter (38), recorded energy absorption in the anterior cruciate ligament of rhesus monkeys as a function of strain rate. Average energy to failure was 31 percent higher at 8.5 mm/sec (42 cm-kgf) than at 0.085 mm/sec (31 cm-kgf). Both peak force and strain at failure were greater at

higher strain rates. Noyes and Grood (39) reported that elastic modulus, failure stress, and strain energy to failure for human anterior cruciate ligaments were all age dependent. Failure stresses for anterior cruciate ligament from donors 20 years of age were three times those for specimens from donors 50 years of age (43 MPa vs. 14 MPa). The change in specimens from donors 50 years of age to those 80 years of age was much less, 14 MPa to 12 MPa. Kenedy et al. (31) reported no significant change in failing strength with age. Mean age for donors in his study was 62 years; so it is likely that many of the donors were in the age range of 50 to 80 years, during which Noyes and Grood (39) claim little change is taking place.

Seering (46) has cited a number of authors who have reported that the moisture content of ligaments affected their mechanical properties. Among these authors, Hirsh and Galante (15, 25) stated that tissue stored at 100 percent relative humidity and at a temperature of 25°C or lower did not change in moisture content. Tkaczuk (50) determined that water is neither taken-up, nor given off, by specimens from anterior and posterior longitudinal ligaments of lumber vertebral columns when they are kept in a chamber at 100 percent relative humidity and approximately 25°C. He also concluded that rapid freezing and subsequent thawing of the lumber spine does not influence their tensile properties.

A. Structure of Ligaments

Ligamentous tissue has the distinguishing characteristic of large amounts of intercellular material being present. This intercellular material may consist of collagenous, elastic or reticular fibers, and varying amounts of an amorphous ground substance. Since the elastic fibers contain the low modulus material elastin (7), the high modulus effect of ligaments is due to the collagenous material present.

The properties of ligaments vary depending upon their location in the body. Hence, it is necessary to study each of the ligaments separately. One such ligament, the anterior cruciate ligament in the knee joint, is composed almost entirely of collagenous fibers which appear as long wavy ribbons which are not branched. The fiber production is apparently in the form of a secretion process in which an aggregate of extracellular material is secreted by star-shaped cells called fibroblasts. The primary building blocks of these fibers are the collagen molecules, three chains of amino acids in certain sequences which coil into left-hand helices and interwine to form a right-handed superhelix (42). Bundles of these fibers lie in a somewhat irregular arrangement parallel to the central axis of the ligament. These collagen bundles are presumably as long as the tendons or ligaments (21). They are surrounded by a woven mesh of loose connective tissue, the peritendineum

internum, containing elastic fibers that tend to draw the bundle into a wavy formation in the relaxed condition. Ligament fibers are continuous with the fibers of the bones and at this point are called the fibers of Sharpey.

B. Viscoelastic Properties of Ligaments

Viscoelastic nature of ligamentous tissue has been verified by many researchers and a number have proposed analytical models for predicting viscoelastic behavior (10, 13, 14, 23, 24, 27, 52). A specimen of ligament, when stretched and maintained to a constant predetermined length, shows an immediate fall-off in the load required to maintain the same degree of extension. This effect is known as "Stress relaxation". Conversely, when the specimen is subjected to a constant load, it undergoes a change in length, a phenomenon known as "Creep". Stress relaxation and creep occur in all materials to some extent because all are viscoelastic in that they can act as viscous fluids or elastic solids under various conditions. The fact that stress relaxation and creep occurs so readily in the ligament indicates that for the loading conditions (rate and temperature), it behaves in a viscous fashion.

Seering (46) has performed a number of stress relaxation tests on human knee specimens. A sudden tibial displacement was applied to each specimen and held for up to five minutes. The force required to maintain this displacement was measured at regular intervals. As would be

expected for a material which exhibits some viscoelastic characteristics, the required force decreased with time. The largest rate of decrease always occurred during the first few seconds of loading; however, the force continued to decrease throughout loading in each case. For relatively small applied displacements, about 3 mm, the knee ligaments softened for a few seconds, but further changes with time were very small (not measurable on an oscilloscope). For larger initial displacements, about 7 mm, the change was much more noticeable. The applied load value did not seem to approach an asymptote over time; rather, it appeared to drop at a uniform rate.

Seering (46) also performed follow-up tests on several of the specimens which had exhibited stress relaxation. In each case, some of the stiffness that was lost was recovered with time. Figures 9 and 10 are given as the results from one such series of tests. Medial and lateral tibial displacements of about 7 mm were each applied and held for 75 seconds. For both displacements, the resulting load decreased about 20 percent during the test period. Figure 9 shows the changes in applied load as a function of time. The largest decrease occurred in the few seconds after displacement was applied. The change then appears to become a linear function of time. Figure 10 contains curves fitted to data collected during single sinusoidal displacement cycles applied to the same specimen at various time intervals after the stress relaxation tests were performed.

Figure 9. Forces Required to Maintain Large Medial and Lateral Tibial Displacements for 75 Seconds.

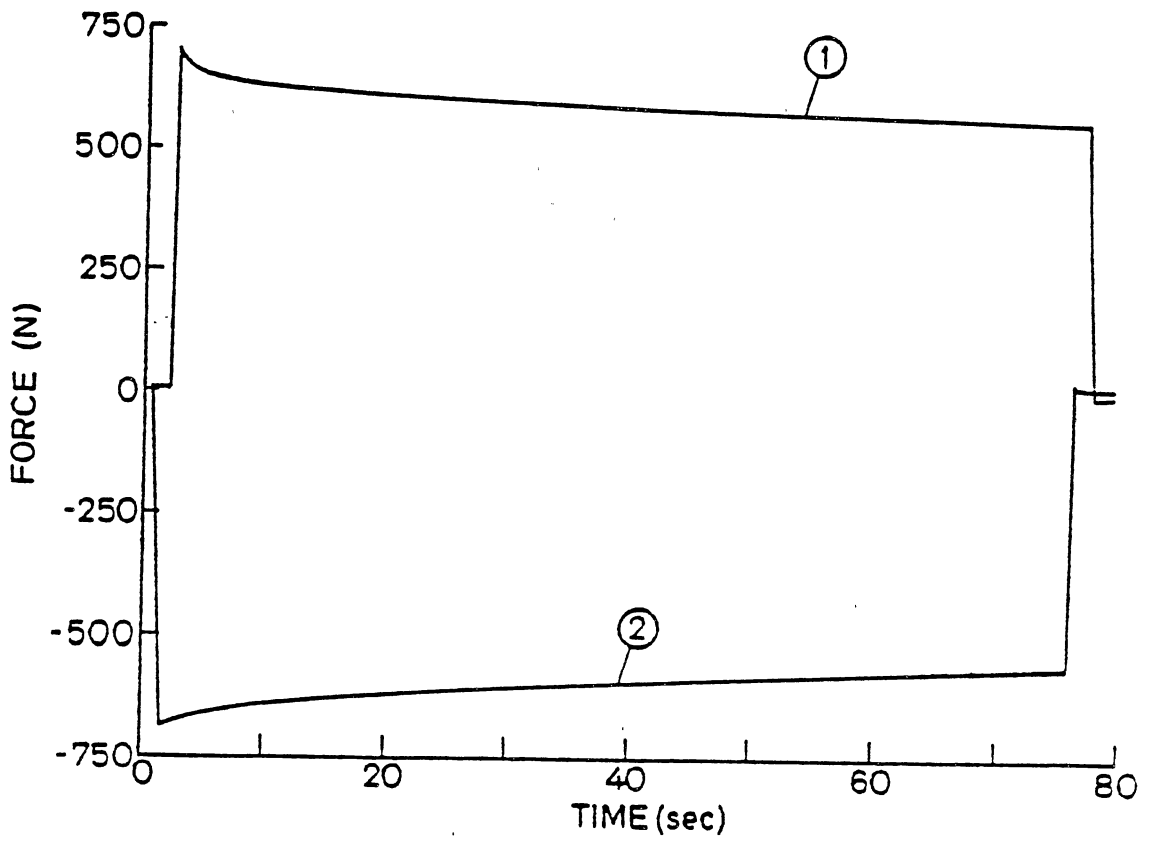
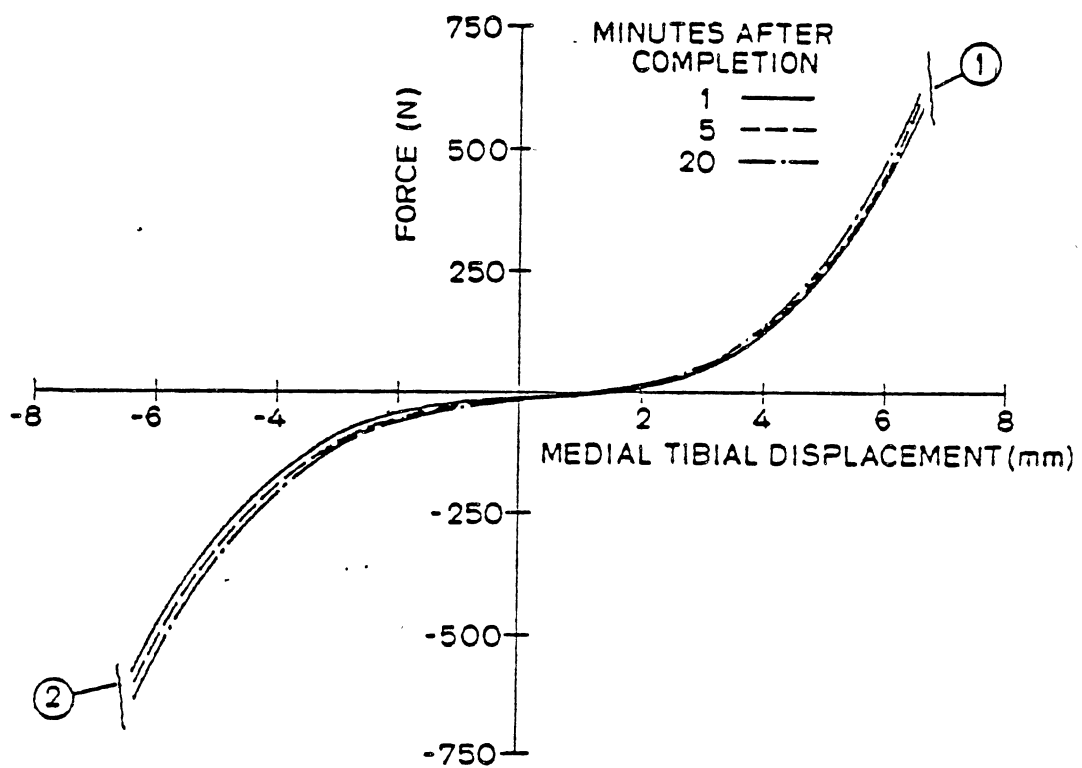


Figure 10. Plots of Force vs. Displacement for the Medial (1) and Lateral (2) Tibial Displacements Presented in Figure 8. The Curves Fit to Data Collected During Sinusoidal Tibial Displacement Cycles Performed at Various Times After Completion of the Stress-Relaxation Tests.



These curves show that the stiffness lost during the stress relaxation tests is recovered with time.

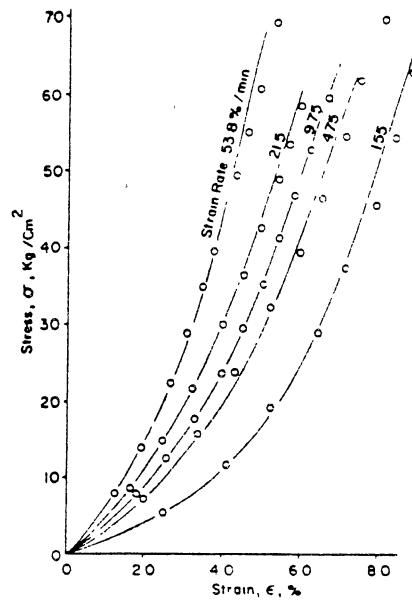
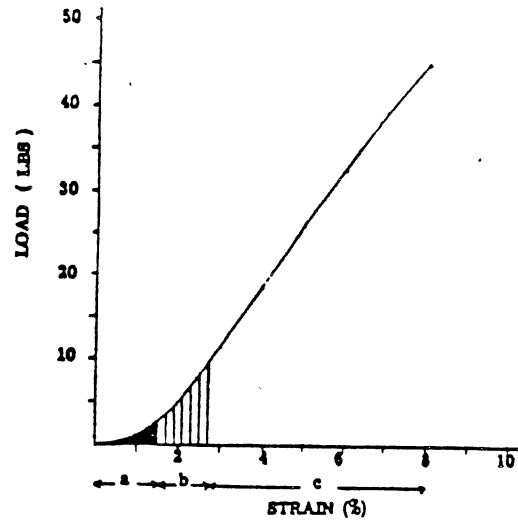
C. Stress-Strain Curves

Stress is an expression of the load per unit area to which a material under loading is subjected. Strain is the increase in length which it undergoes, expressed as a percentage of original length. From a load-displacement tracing, it is possible to construct a load-strain curve or stress-strain pattern for the material concerned. A typical load-strain curve is shown in Figure 11 and has the characteristic shape found in other biological tissue (56). Abrahams (1) described such a curve as having three distinct regions before rupture of the specimen occurred. The primary region (a) is that of 0-1.5 percent strain where there is considerable increase in extension with a smaller increase in tension; the secondary region (b), 1.5-3.0 percent strain, may be regarded as a transitional zone where the stiffness is increasing steadily until, in the final phase (c), it assumes a linear relationship between stress and strain. Frisen and Viidik (13) have likened tendon to a coiled spring which can be stretched until it becomes a straight steel wire. The stiffness changes gradually from a low value for the undeformed spring, region (a), to the much higher value for the straight wire, region (c).

Haut and Little (23), while studying the mechanical properties of the anterior cruciate ligament of canines,

Figure 11. Load-Strain Curve for Specimen of Rabbit Tendon (F.D.L.) Loaded at 10 Inches per Minute. Zone (a) 0-1.5 Per Cent Strain; Minimal Increase in Tension; (b) 1.5-3.0 Per Cent Strain; Stiffness Gradually Increases; (c) Greater Than 3.0 Per Cent; Linear Relationship Between Load and Strain.

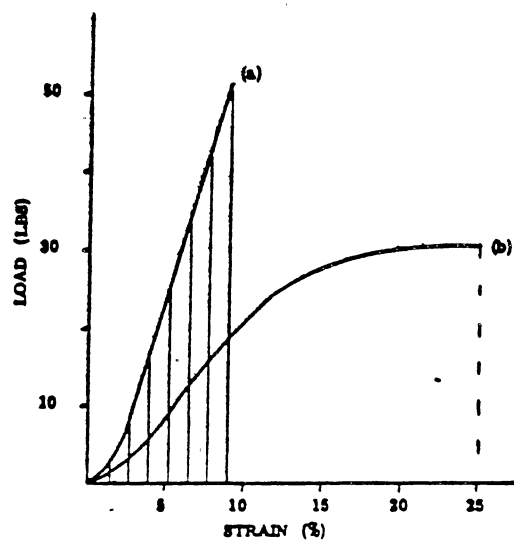
Figure 12. Stress-Strain Plot for Specimen of Canine Anterior Cruciate Ligament. The Specimen was Immersed in a Saline Bath at 72°F.



plotted the stress-strain curves at various strain rates. Different testing environments were used. One such plot for specimens immersed in a saline bath at room temperature is shown in Figure 12. The shape of the stress-strain curves is not altered by various strain rates, except for high strain rates where the toe region appears at an early strain level. Unlike the range of 0-1.5 percent strain for the toe regions of tendons given by Abrahams (1), the anterior cruciate ligament of dogs displays a toe region which varies with strains up to six percent. Also, as it is seen in this figure, the increase in the apparent Young's modulus with increasing strain rate is quite pronounced. For slow strain rates, the modulus changes rapidly with strain rate, but for higher values of strain rate, only a gradual increase can be seen.

While low rates of loading are less efficient in terms of the transmission of force by a ligament, they do allow for a greater absorption of energy by the same ligament. From Figure 13, as given by Welsh, Macnab, and Rilley (56), in their studies of rabbit tendons, it is evident that the energy which can be absorbed at a low rate of loading (represented by the area under the stress-strain curve), is far greater than that at a high rate of loading. Thus, an injury such as a rupture of the medial collateral ligament of the knee is sustained by application of a sudden sharp valgus strain to the limb, whereas an equivalent force applied slowly would probably have little effect.

Figure 13. Load-Strain Pattern for F.D.L. Tendon. (a) Loading Rate - 10 Inches per Minute; (b) Loading Rate - 0.2 Inches per Minute. The Strength May be Greater for a Tendon Loaded Rapidly, But it Requires More Energy to Produce Failure if the Tendon is Loaded Slowly. Energy Absorbed by Tendon Represented by the Area Under Load-Strain Curve.



Ligament Stiffness as a Function
of Strain Rate

As a general rule, the behavior of all materials under tensile loading is dependent on strain rate to a greater or lesser degree. For example, a common material like steel, when subjected to very rapid loading, behaves as a brittle solid. However, when loaded steadily over a long period of time, the same material will exhibit viscous properties and creep quite markedly. A similar response of ligaments to different rates of loading is important in efficiently transmitting forces to the bone. At high rates of loading, a high stiffness prevails and this results in less energy being absorbed by the ligament. Thus, a sudden exertion is more effective in lifting or moving a heavy load than if the same amount of load is expended over a longer period of time.

Seering (46), in his study of the mechanical properties of the knee ligaments, measured the changes in ligament stiffness as a function of strain rate. He performed a series of tests with a constant displacement, but at different strain rates. A single knee specimen was chosen and a 30-minute interval was given between tests for the specimen to recover the stiffness lost as a result of previous testing. Seering carried out several other tests to show that within 20-minutes or less, a specimen would completely recover the stiffness lost as a result of a single large

loading cycle, if the specimen is stored in a high humidity chamber. However, the test started with a five second sinusoidal 7 mm amplitude posterior tibial displacement applied to the specimen and the same test was repeated every hour. Again, every hour on the half hour, the specimen was cycled to the same amplitude sinusoidally, but at a time ranging from 10 seconds per cycle to 0.1 second per cycle. According to Seering (46), 30-minutes between tests was more than enough time to insure full recovery of the stiffness lost during the previous test cycle. For the five seconds per cycle tests, the force at 7 mm displacement varied from the average by only ± 2 percent for all the runs (Table V). The forces at 7 mm displacement for tests performed at other rates varied through a range equal to 20 percent of the average value for the 5 seconds per cycle tests (Table VI and Figure 14). From this work, it was concluded that ligament stiffness is a function of strain rate. However, the change in stiffness over a wide range of strain rates was small enough that, for purposes of modeling, a single curve can be chosen to represent the stiffness of a given element or set of elements for all strain rates within that range.

D. Ligaments Under Fatigue (or Cyclic) Loading

Not many reports on the properties of ligaments under fatigue (or cyclic) loading could be found in the literature. Weisman, Malcom, and Robert (55) reported on the

TABLE V

PEAK FORCE AND PEAK STIFFNESS RESULTING FROM
FIVE SECOND PER CYCLE SINUSOIDAL POSTERIOR
TIBIAL DISPLACEMENTS APPLIED ONCE EACH
HOUR FOR SEVEN HOURS

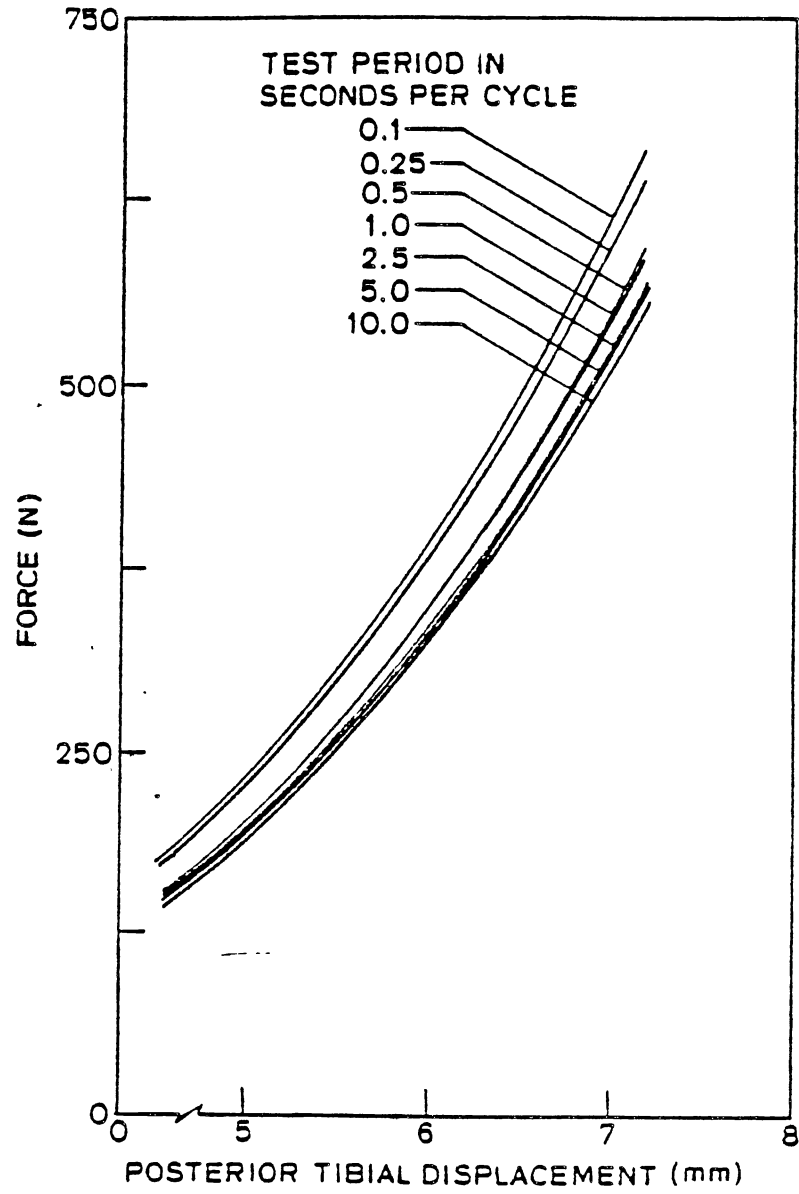
Test No.	Peak Force (N)	Percent of Average Force	Peak Stiffness (kN/m)	Percent of Average Stiffness
1	598	101	241	102
2	598	101	238	100
3	585	98	235	99
4	586	99	236	100
5	591	99	240	101
6	599	101	235	99
7	598	101	236	100

TABLE VI

PEAK FORCE AND PEAK STIFFNESS RESULTING FROM
SINGLE SINUSOIDAL POSTERIOR TIBIAL DIS-
PLACEMENTS OF FIXED AMPLITUDE APPLIED
AT VARIOUS CYCLE PERIODS

Test No.	Period of Applied Displacement	Peak Force (N)	Percent of 10sec/cycle Force	Peak Stiffness (kN/m)	Percent of 10sec/cycle Stiffness
1	10 sec	586	100	235	100
2	5	594	101	237	101
3	2.5	598	102	241	103
4	1	622	106	249	106
5	0.5	627	107	256	109
6	0.25	685	117	269	114
7	0.1	708	121	282	120

Figure 14. Curves Fit to Data Collected From an In-Vitro Human Knee During Posterior Sinusoidal Tibial Displacement Cycles of Various Periods.



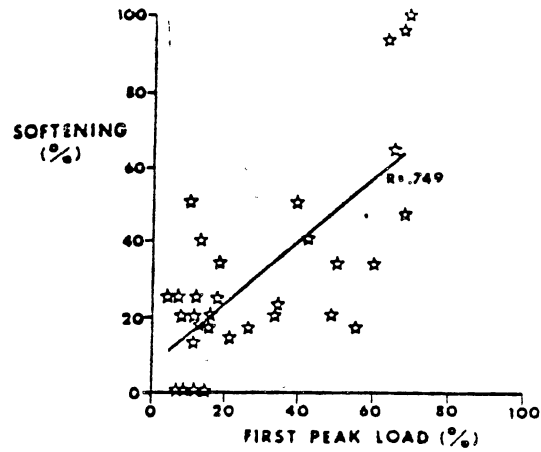
effect of cyclic loading on rat medial collateral ligaments in-vitro. The ligaments were cycled from 0 to 70 percent of their ultimate strength for approximately 960 cycles at a frequency of 16 cycles per minute. Two phenomena were observed. One of these phenomena is the softening of the ligament, defined as the ratio of the last cycle peak to the first cycle peak load, presented as a percent. The second phenomenon is a reduction in strength of the cycled ligament, presented as a ratio of the ultimate strength of the cycled ligament, expressed as a percent in order to reduce animal variability. The authors found a good relationship between the amounts of softening and the value of the first peak load (as a percentage of strength) ($R = .749$) (Figure 15). A logarithmic correlation exists between the softening of the ligament and its reduction in strength ($R = .956$) (Figure 16). Under visual examination, the surface of the uncycled ligaments have a glossy appearance while the cycled ligaments appear duller. The relative waviness of the collagen fibers seen in the cycled ligament is less than that of the uncycled ligament.

Weisman et al. (55) also imparted some known amounts of cyclic loading to ten people in the laboratory to assess the changes in compliance of the medial collateral ligament and they found the mean change in compliance to be 16.83 percent (S.D. = 30.50). All subjects returned to their initial compliance in at least one hour after cycling. Hence, the "stretching out" of the ligaments due to cyclic

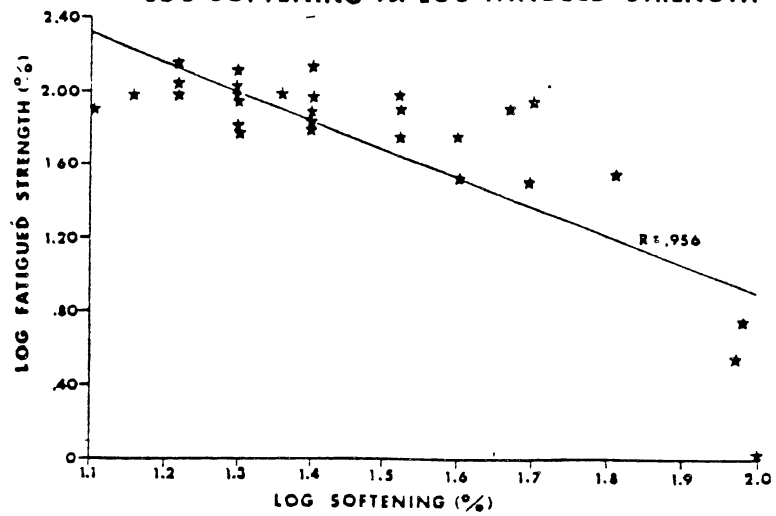
Figure 15. Relationship Between the Amounts of Softening and the Value of the First Peak Load Under Cyclic Loading.

Figure 16. Logarithmic Correlation Between the Softening of the Ligament and its Reduction in Strength Under Cyclic Loading.

FIRST PEAK LOAD vs. SOFTENING



LOG SOFTENING vs. LOG FATIGUED STRENGTH



loading is recoverable in a short period of time. However, it is believed by the authors that a second phenomenon occurred, that of fatigue, which may initiate a healing process ultimately strengthening the ligament. It is also noted that biological materials cannot actually fatigue in the same manner as an inorganic material because of their inherent ability to repair themselves. However, the repairing process is sometimes deficient or too slow and fatigue occurs.

The effect of fatigue or cyclic loading on ligaments and tendons has been studied by few other authors. However, they have often reported with contradiction on their findings. Viidik (52) discussed a softening which occurs during the first few test cycles performed on rabbit ligaments. Seering (46) cited some of the authors who have reported that collagenous structures become stiffer and residual deformation and hysteresis decrease as a result of cycling. Among these authors, Rigby (44) states that tendon collagen kept in 0.9 percent saline at 20°C and cycled at 1.75 minutes per cycle will soften through the first 100 cycles and then, with further cycling, stiffen increasingly throughout 1000 cycles. In contrast, to check these results and to study the effects of repeated cycling on ligament stiffness, Seering (46) tested several human knee specimens, each through hundreds of cycles. For all tests, sinusoidal tibial displacements of fixed amplitude were applied. Specimens were kept in a chamber with air at 21°C and

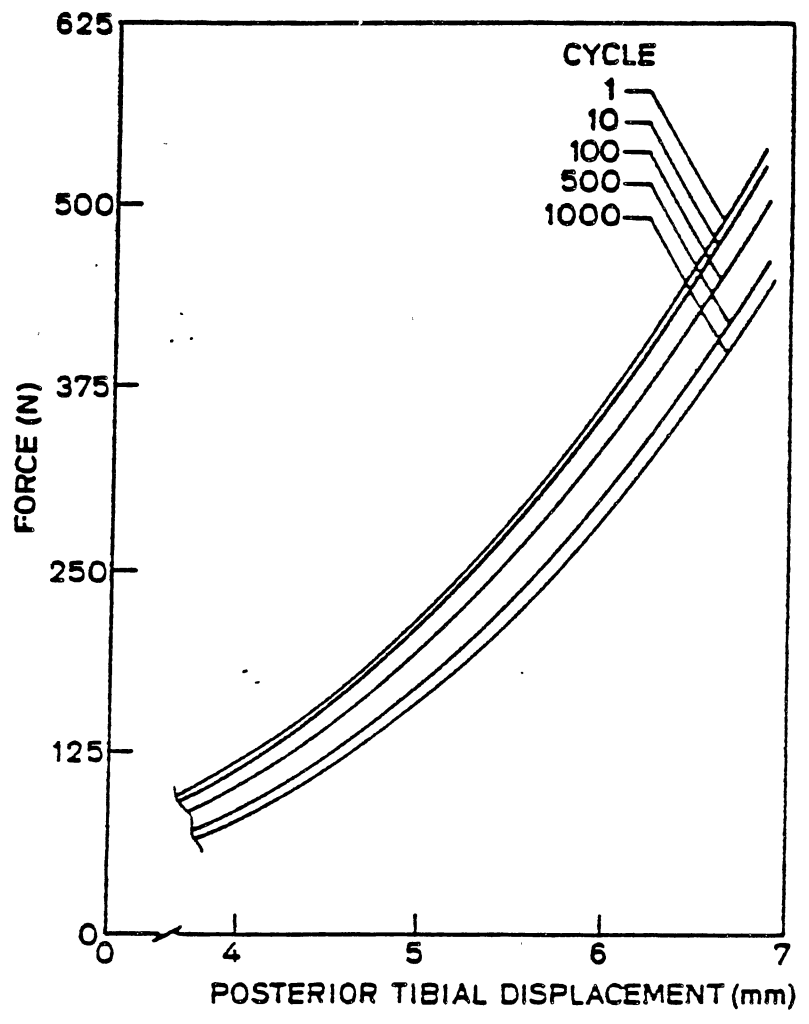
100 percent humidity throughout. After the tests, it was found that, in every case, knee stiffness decreased with cycling. The changes during each of the first 10 to 15 cycles were always the largest. However, stiffness continued to decrease throughout the test period. No increase in stiffness with cycling was ever observed. Table VII gives the information on one such cycling test. A specimen was loaded with a sinusoidal tibial displacement of ± 7 mm amplitude at 1 Hz for 1000 cycles. The table gives the force values and stiffness values corresponding to +7 and -7 mm displacements as determined from the curves fitted to the data collected during each of several cycles. Test results show that the force required to displace the tibia anteriorly 7 mm was 15 percent less on the 1000th cycle than on the first; the force required to displace the tibia 7 mm posteriorly was 19 percent less on the 1000th cycle than on the first. Stiffness for the two cases decreased 13 percent and 9 percent respectively. Figure 17 presents the curves fitted to data collected during various of the 1000 cycles. This plot illustrates the softening which occurred as cycling continued. Seering went further to measure the energy required to displace the tibia during each of the recorded cycles and calculated the energy lost during each cycle. Energy required per cycle dropped throughout the testing with almost half the decrease occurring during the first 100 cycles. Seering attributed this phenomenon largely to the fact that, as the cycling

TABLE VII

PEAK FORCE AND PEAK STIFFNESS ATTAINED FROM
VARIOUS CYCLES OF ANTERIOR AND POSTERIOR
TIBIAL DISPLACEMENTS DURING 1000
SEQUENTIAL CYCLES AT 1 Hz

Cycle No.	Peak Force (N)	Percent Initial Peak Force	Peak Stiffness (kN/m)	Percent of Initial Peak Stiffness
FOR ANTERIOR TIBIAL DISPLACEMENT				
1	552	100	247	100
10	542	98	245	99
50	529	96	242	98
100	519	94	239	97
200	507	92	235	95
300	499	90	232	94
400	493	89	230	93
500	488	88	229	92
600	484	88	228	92
700	480	87	226	92
800	478	87	226	91
900	475	86	225	91
1000	472	85	223	90
FOR POSTERIOR TIBIAL DISPLACEMENT				
1	566	100	237	100
10	554	98	234	98
50	538	95	229	97
100	524	93	224	95
200	506	89	218	92
300	494	87	216	91
400	486	86	213	90
500	480	85	213	90
600	475	84	211	89
700	468	83	207	87
800	464	82	206	87
900	460	81	204	86
1000	459	81	206	87

Figure 17. Curves Fit to Data Recorded During Several of
1000 Consecutive Applied Sinusoidal Tibial
Displacements.



continued, smaller loads were required to create the displacements. Energy lost decreased throughout the first 50 cycles and then gradually increased for each cycle during the remainder of the test. These results, as stated by Seering (46), contradicted those of Galante (15), Gratz (17), Rigby (44), and Viidik (51). However, Seering defended his results saying that the data of his tests were analyzed numerically while the others evaluated their data visually.

In another report, Rigby (43) stated to have found that if the strain does not exceed approximately four percent, a tendon can be cycled indefinitely without damage. Bartel, Marshall, Schieck, and Wang (2), demonstrated that the change in length of the medial collateral ligament during knee flexion is four percent. Damage will accumulate if the ligament is repeatedly cycled beyond this defined physiological limit within a short period of time. This damage should result in "fatigue softening" of the ligament. The possibility of stretching the knee ligaments beyond the physiological range during normal sporting activities such as soccer, hockey, skiing, and basketball has not previously been demonstrated. The combination of rotation and varus-valgus forces imposed on the knee joint during these activities could account for this stretching. If this stretching does in fact occur, it is most probable that some kind of fatigue damage is occurring to the ligament.

CHAPTER V

PERMANENT PROSTHETIC LIGAMENT IMPLANTS

The subject of study in this thesis is to investigate on the mechanical properties of one of the ligament prosthesis, specifically the fatigue behavior of one Dacron artificial ligament. Recently, a considerable interest has been shown in the development and use of prosthesis for ligaments, specially the cruciate ligaments of the knee. As such, a report on some of the ligament implants being used for permanent prosthetic replacement is briefly presented in this chapter.

Any permanent prosthetic replacement must be compatible. It must have sufficient mechanical strength and survive through millions of fatigue cycles associated with normal ligament use. There are several ligament implant systems now being marketed. In view of most of the authors, none of these implants adequately fulfill these requirements. A review of the literature reveals long-term problems with all permanent prosthetic replacements (19, 32, 45, 54).

Polyethylene and Dacron are the two kinds of prosthetic materials being used today. Grood and Noyes (19) studied one such prosthesis now being marketed by Richards Manufac-

turing Company. They have compared fatigue endurance limits and strength of ultra-high-molecular-weight polyethylene implants with those of human specimens. The mechanical properties tested were; tensile strength, residual elongation, creep, and bending fatigue.

As reported by Grood and Noyes (19) and shown in the tensile stress-strain curve (Figure 18), the implant yields and deforms plastically at elongation from 10 to 40 percent. Above 40 percent elongation, the polyethylene implant undergoes strain hardening and stress rises steadily until implant failure. The ultimate failure strength is 41 megapascals (6000 psi) and occurs at elongation ranging from 350 to 450 percent. The load elongation curve is shown in Figure 19. The elastic region is non-linear with the stiffness decreasing as force increases. The higher strain rate causes an approximate 15 percent increase in the yield point and an earlier onset of strain hardening.

Residual elongation tests were performed and Figure 20 shows residual elongation after a single load-unload cycle. A 203 newton force produced a peak elongation of 2.3 percent. After removal of the force, a residual elongation of 0.15 percent remained. It took approximately one hour for the implant to return to its original length. At a low peak force of 117 newtons, residual elongation is not very apparent and recovers fully at 10 or 100 seconds. At the highest applied force, 273 newtons, full recovery does not occur even after 90 minutes. In the cyclic creep tests,

Figure 18. Stress-Strain Curve of Ultra-High-Molecular-Weight Polyethylene Ligament Implant. The Test was Carried at a Strain Rate of 1 Per Cent Per Second at Room Temperature. The Solid Line Indicates a Previously Unstressed Implant. The Interrupted Line Indicates an Implant After it Has Been Subjected to Eighty-one Million Stress Reversals in the Rotating-Beam Fatigue Tester.

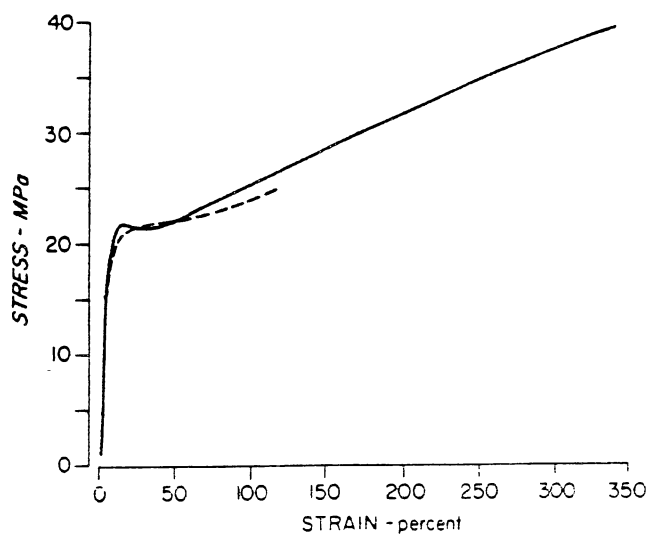
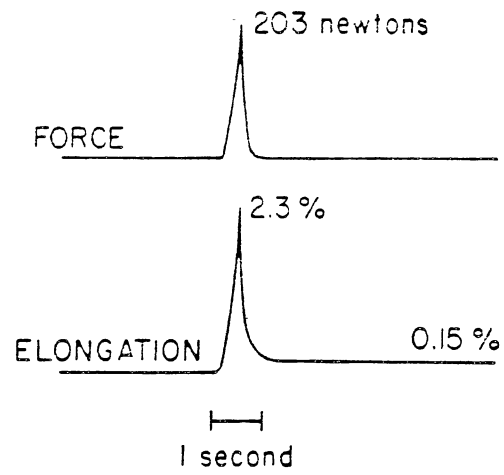
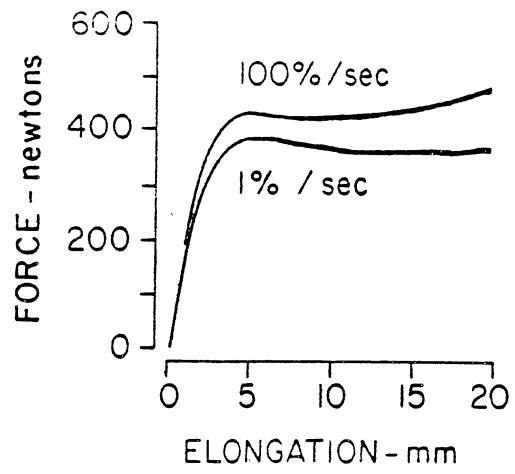


Figure 19. Load-Elongation Curve for the Polyethylene Implant at Low (1 percent per second) and High (100 percent per second) Elongation Rates.

Figure 20. Residual Elongation After a Single Load-Unload Cycle.



the prosthesis was cycled 250,000 times at 250 newtons at 1 Hz. The average load of 130 newtons was subjected to the prosthesis during the cyclic tests which resulted in a 22 percent (7.6 mm) creep elongation of its length. At the end of cyclic testing, the implant began to shorten progressively and recovery was found to be 40 percent of the 7.6 mm maximum elongation after 1000 seconds.

In fatigue tests, the implant was subjected to 81 million stress reversals at a rate of four per second. During the first 20 million cycles, the implant was bent in the narrow center section to an angle of 30 degrees and the remaining portion of the tests were conducted with an imposed bend of 45 degrees. The bending moments for those bends on an untested implant were 261 and 396 newton-meters. After 81 million cycles, the bending moments required were 191 and 364 newton-meters respectively. No sign of failure was visually apparent at the end of the tests. The implant was then tested as to failure in tension and showed an alteration in its curve of load-elongation when compared with normal implants (Figure 18), namely a 10 percent reduction in stiffness and a marked reduction in ductility. Failure occurred through the section with reduced diameter at a stress of 25 magapascals and an elongation of 110 percent. In comparison, the normal implant failed at a stress of 40 magapascals at nearly 350 percent elongation. The fatigued implant did not exhibit a well defined yield point or a subsequent drop in stress prior to the onset of strain

hardening. Instead, the force rose steadily as the implant was elongated.

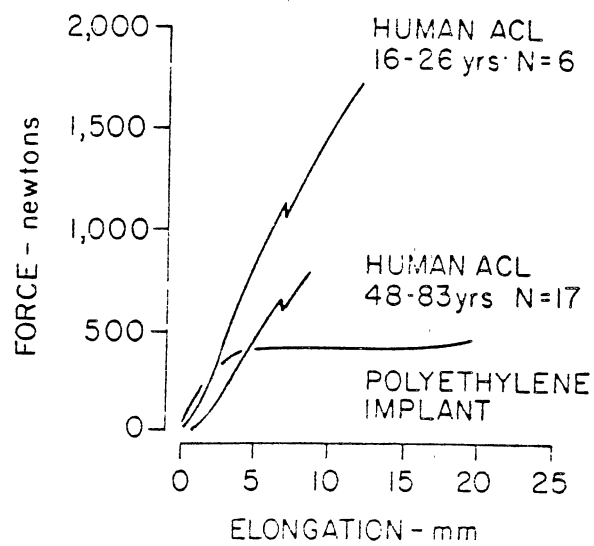
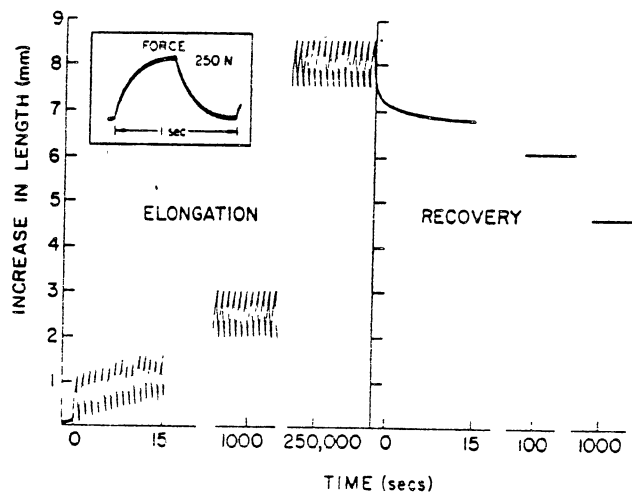
Grood and Noyes (19) compared the mechanical properties of the polyethylene implant with that of human anterior cruciate bone-ligament-bone units. Figure 21 shows the comparison of load-elongation behavior of the implant and the human tibia-anterior cruciate ligament-femur preparations from younger and older adults. The implant deforms plastically and elongates permanently at a force approximately one-half and one-fourth the maximum failure force of the specimens obtained from older and younger humans respectively.

In the cyclic creep tests, no adverse effects on mechanical strength of the implant were found. However, the creep behavior shown in Figure 22, suggests that under in-vivo loading conditions, the prosthesis will undergo significant amounts of creep elongation and recovery concomitant with periods of activity and rest.

The exact forces to which the anterior cruciate ligament is subjected are unknown. Grood and Noyes (19), however, referred to Morrison (35, 36, 37) who calculated the forces acting on each cruciate ligament during level walking (anterior cruciate ligament, 169 newtons; posterior cruciate ligament, 352 newtons); during ascending (67 and 641 newtons) and descending stairs (445 and 262 newtons); and during ascending (27 and 1215 newtons) and descending a 9.45 degree ramp (449 and 93 newtons). For the anterior cruciate ligament, there is one condition in

Figure 21. Creep Elongation During Cyclic Loading. The Creep Elongation and Cyclic Change in Length Which Result From the Load-Cycling for Various Times During the Test are Shown on the Left. A Single Load Cycle is Shown in the Inset. The Shortening of the Implant Which Occured on Cessation of Load Cycling is Indicated on the Right.

Figure 22. Comparison of Load-Elongation Behavior of the Implant and the Human Tibia-Anterior Cruciate Ligament-Femur Preparations.



which the forces presented exceed the yield point force of the implant (420 newtons). If the prosthetic ligament is, in fact, subjected to forces of 449 newtons or larger, such as may be expected during strenuous activity, the prosthesis will elongate and its ability to control anterior-posterior knee stability will be diminished. However, according to Grood and Noyes, with all the assumptions made by Morrison, the average maximum force values calculated by him are within a factor of two of the actual maximum forces during normal walking.

Assessing the adequacy of the strength of the polyethylene implant, Grood and Noyes concluded saying that the margin of safety of the Richards implant is low in terms of its ability to resist expected in-vivo forces without sustaining permanent deformation. The data suggested that the implant may function adequately in an older patient in whom the level of activity is restricted to avoid large muscle and joint forces which subject the implant to high in-vivo loads. No definitive conclusions could however, be made on the safe acceptability of the implant system in general.

Dacron is another kind of prosthetic material being used for artificial ligament. Not much of a publication could be found in the literature on the behavior of dacron ligamentous implant. As reported by Bejui, Tabutin, and Perot (3), many surgeons have attempted to use both dacron (41) and polyethylene (4) ligament implants, but breakage and reactive synovitis are common phenomena. One aspect of

the mechanical strength i.e. fatigue behavior of dacron implant in bending with certain pre-load in tension and torsion, being investigated in this study is discussed in detail in Chapter VII.

Nevertheless, researchers are still investigating on the causes of in-vivo implant failures and on the problems of safe use of both polyethylene and dacron implants as a permanent prosthetic replacement of the knee ligaments. However, much depends on the knowledge of the magnitudes of forces to which the cruciates and other ligaments or their replacements would be subjected to during various activities. With the knowledge of these forces, definitive statements could be made as to the mechanical suitability of the implants.

CHAPTER VI

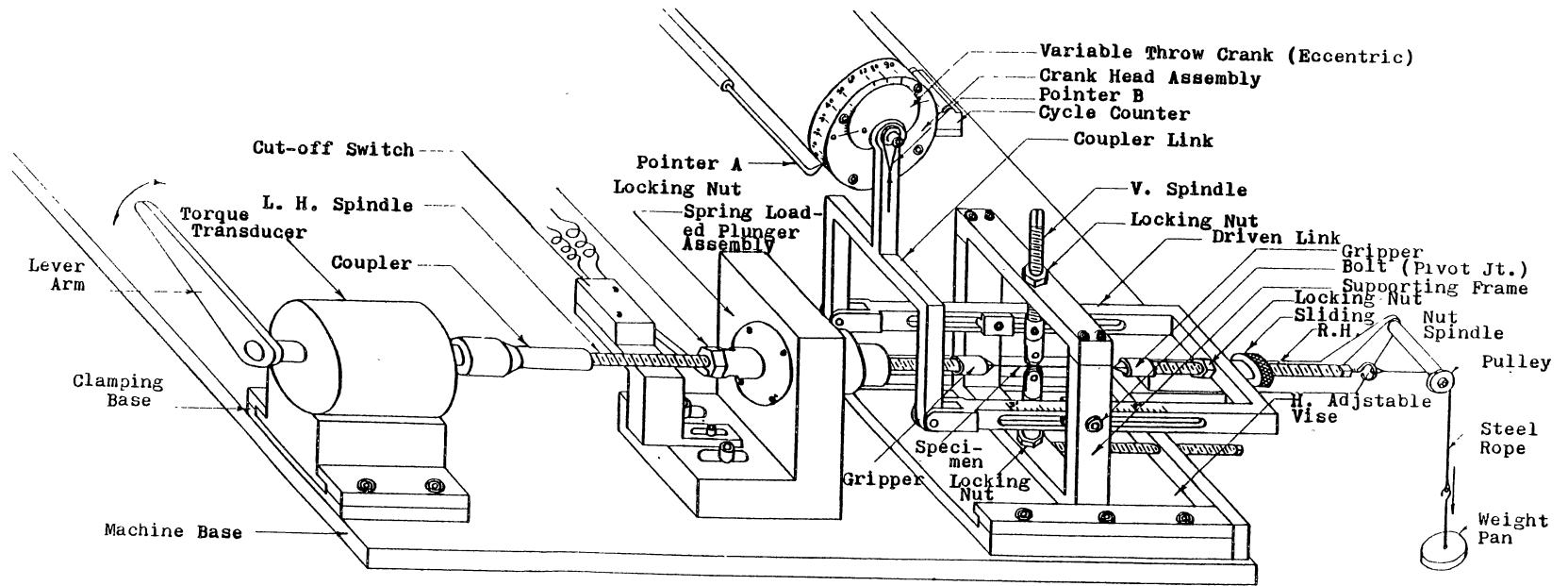
DESIGN OF THE EXPERIMENT

The main objective of this study was to carry out the fatigue analysis of an artificial dacron ligament which could be recommended for use as cruciate ligament prosthesis in human knee joint. The fatigue testing was to be carried out in bending with the specimen pre-loaded in combined tension and torsion. In order to apply this kind of loading, it was necessitated to build a specially designed Fatigue Testing machine alongwith suitable loading fixture.

A. Design of the Fatigue Testing Machine

The machine was designed primarily for the testing of any flexible material having a small cross-section for fatigue testing under a certain range of bending stresses. The specimen could be pre-loaded in tension and torsion. The machine with its combined tensile, torsional, and bending stress attachment as shown in Figure 23, is a compact bench mounted unit, consisting of a rigid bolted frame, horizontally adjustable vise, variable throw crank, cycle counter, cut-off switch, variable speed drive, and provision for applying pre-tensile and pre-torsional load to the

Figure 23. Combined Stress Attachment for Fatigue Testing Machine.



specimen. The unit occupies a space of 30 inches long, 24 inches wide, and $16\frac{1}{4}$ inches high.

Loading Mechanism and Components

The leading force of the machine is transmitted by a variable speed motor through a pulley and belt to the main shaft. Force from the main shaft is then transmitted through a four-link mechanism to the specimen. The variable throw crank itself, at the end of the main drive shaft is acting as the input (driving) link of the four-link mechanism. The variable throw crank adjusts the amount of deflection (i.e., the crank or input link length) and transmits the force through a rigid coupler link to the output (driven) link. The output link is extended beyond the pivot joint in a cantilever fashion. The oscillating motion of this end of the output link is then used to apply cyclic bending load to the specimen. The effective length of the output link, from the coupler link to the pivot joint, can be adjusted by sliding the pivot joints along the longitudinal slots provided in both arms of the output link.

Pre-tensile load is applied by putting dead weights on a weight pan, suspended on a steel rope through a pulley fixed to a bracket. The rope in turn is attached to the right hand spindle. The spindle which is fixed with a specimen holding chuck (gripper) is adjustable to accommodate shorter or longer specimen sizes. Pre-torsional load to the specimen is applied from the other end by a torque

transducer through another adjustable spindle at the left which is also fixed with a specimen holding chuck (gripper).

Frame. The structural frame consists of four vertical supports rigidly bolted to the base plate. The loading fixture contains mainly the coupler and output links of a four-bar linkage mechanism connected together by pin joints. Both the coupler link and output link has two arms. The supporting frame holding the output link has also two arms joined together at the top by a seperate link.

Horizontal Vise. The vise is horizontally adjustable and essentially it adjusts the position of the supporting frame holding the output link. Vertical adjustment to accomodate various sizes of specimen is provided in the supporting frame itself through a vertical spindle at the top and the bottom, independent of each other.

Crank. The variable throw crank has a range of 0 to $1\frac{1}{2}$ inches and is adjusted by loosening the four socket head screws around the face of the crank and adjusting the socket head screw near the eccentric. The coupler link (connecting rod), which rides on a precision bearing, transmits the eccentric action of the crank to the output link, which is also acting as the bending arm for applying bending load to the specimen.

Flywheel. The flywheel is mounted on the main drive shaft. The transmitted force as absorbed by the specimen

and the loading fixture is never uniform during the cyclic up and down movement of the specimen. The oscillating movement of the output link has a zero velocity and acceleration at its two extreme positions. During each cycle, the force absorbed by the output link varies twice from zero to its maximum value. The flywheel serves as a means to store energy during those moments when not much of force is needed to drive the output (driven) link and supply this energy during those moments when some extra force is required to drive the output link.

Spring Loaded Plunger Assembly. The spring loaded plunger holding the left hand spindle is housed on a rigid block at the left. The plunger itself has a free rotating motion and only a limited translatory motion along its horizontal axis. Two cylindrical type oil impregnated bronze bearings and two precision thrust bearings, one at each end of the plunger, facilitates the free movement of the plunger without any significant amount of friction loss. The thrust bearing at the left takes only the spring thrust due to its compression. The thrust bearing at the right takes up all the load provided by the tension of the specimen. The spring is a compression type with ends closed and square grounded. It has a force of 32 lbs/in with 70 percent deflection. After the specimen is pre-loaded in tension, the spring should be in full compression. When the specimen fails, this compressive force of the

spring slides the plunger backward to a dead end. This backward movement of the plunger then strikes the cut-off switch in order to stop the machine.

Cut-off Switch. The cut-off switch is provided to shut off the entire machine when the specimen fails. It is an extended lever type, spring reset, 115 volt, 15 amps micro-switch. The cut-off switch is mounted and fixed in such a position that the lever arm of the micro-switch normally remains pushed by the locking nut at the end of the spring-loaded plunger. At this position, the circuit is open and the machine is not running; the spring loaded plunger is seated on its backward dead end position and the specimen is not yet pre-loaded in tension. When the specimen is pre-loaded in tension, the plunger moves to the right and releases the lever arm of the micro-switch. At this, the circuit gets closed and the machine will run when the power supply from the main is switched on.

Weight Pan. The weight pan is suspended freely by a steel rope through a pulley mounted on a bracket. The rope is fastened to the end of the right hand spindle. The pan itself together with the hanging portion of the rope has negligible weight as compared to the amount of tensile load to be applied to the specimen by putting dead weights on the weight pan.

Cycle Counter. The counter located on the right side

of the machine, is a six digit, resettable unit, which counts every three cycles. The counter is driven by a flexible wire attached to the main drive shaft of the machine.

Variable Speed Drive. The drive system consists of a $\frac{1}{2}$ H.P., 1725 rpm, 115 volt variable speed motor and a link V-belt which drives the main shaft. The speed ranges are adjusted by the speed control knob on the STATOTROL JR controller, which is specially designed for applications requiring only the basic start, stop, and speed control functions.

Torque Transducer. Pre-torsional load is applied with the help of a torque transducer which is mounted on a clamping base at the extreme left of the machine. A lever arm, fixed to one end of the rotor shaft is used for applying the required amount of torque. The torsional load is then transmitted to the specimen from the transducer through a coupler joint and the horizontally mounted left-hand spindle, to which the left-hand specimen holding chuck (gripper) is fixed. The left hand spindle is free to move longitudinally along the axis of the coupler which has a key slot all through the movement of the spindle inside the coupler. However, the key provided at the left end of the spindle, restricts the free rotatory movement between the coupler and the spindle.

B. The Motion Analysis of Four-Link Mechanism

The four-link mechanism through which the motion is being transmitted from the main shaft to the specimen, is essentially a crank-rocker mechanism, in which the input link (the variable throw crank) rotates through 360° and the output link oscillates through an angle ϕ . The mechanism as has been designed is shown in Figure 24. Both the input link, OA and the output link, BM could be varied in length as per requirement. The respective link lengths are as follows:

Input link, OA	;	0 to $1\frac{1}{2}$ "	(variable)
Coupler link, AB	;	8"	(constant)
Output link, BM	;	3 to $5\frac{1}{2}$ "	(variable)
Fixed link, OM	;	8.41 to 9.71"	(varies when input or output link length is varied.)

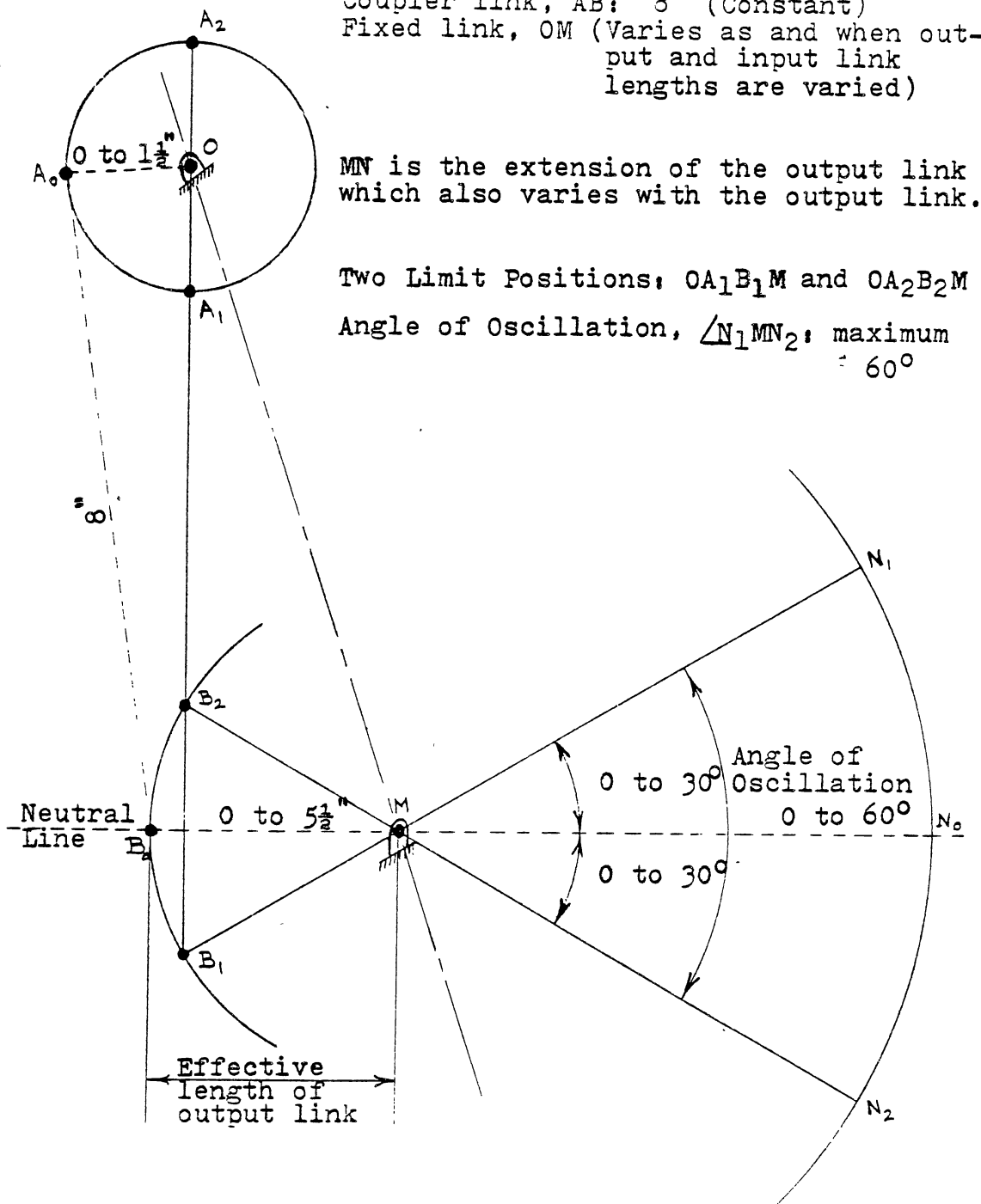
Grashoff Criteria

The type of motion displayed by a four-link mechanism can be predicted by applying Grashoff criteria to this mechanism. The Grashoff criteria are stated as follows:

1. A four-link mechanism belongs to Class I mechanism provided the sum of the lengths of the shortest and longest links is less than or equal to the sum of the lengths of its other two links. For Class I mechanisms, the type of mechanism can be predicted using the following rules:

Figure 24. Four-Link Mechanism of Fatigue Testing Machine.

Input link, OA: 0 to $1\frac{1}{2}$ "
 Output link, BM: 3 to $5\frac{1}{2}$ "
 Coupler link, AB: 8" (Constant)
 Fixed link, OM (Varies as and when out-
 put and input link
 lengths are varied)



a. If the shortest link is the input link, then the four-link mechanism is a crank-rocker mechanism.

b. If the shortest link is the fixed link, then the four-link mechanism is a crank-crank or a drag-link mechanism.

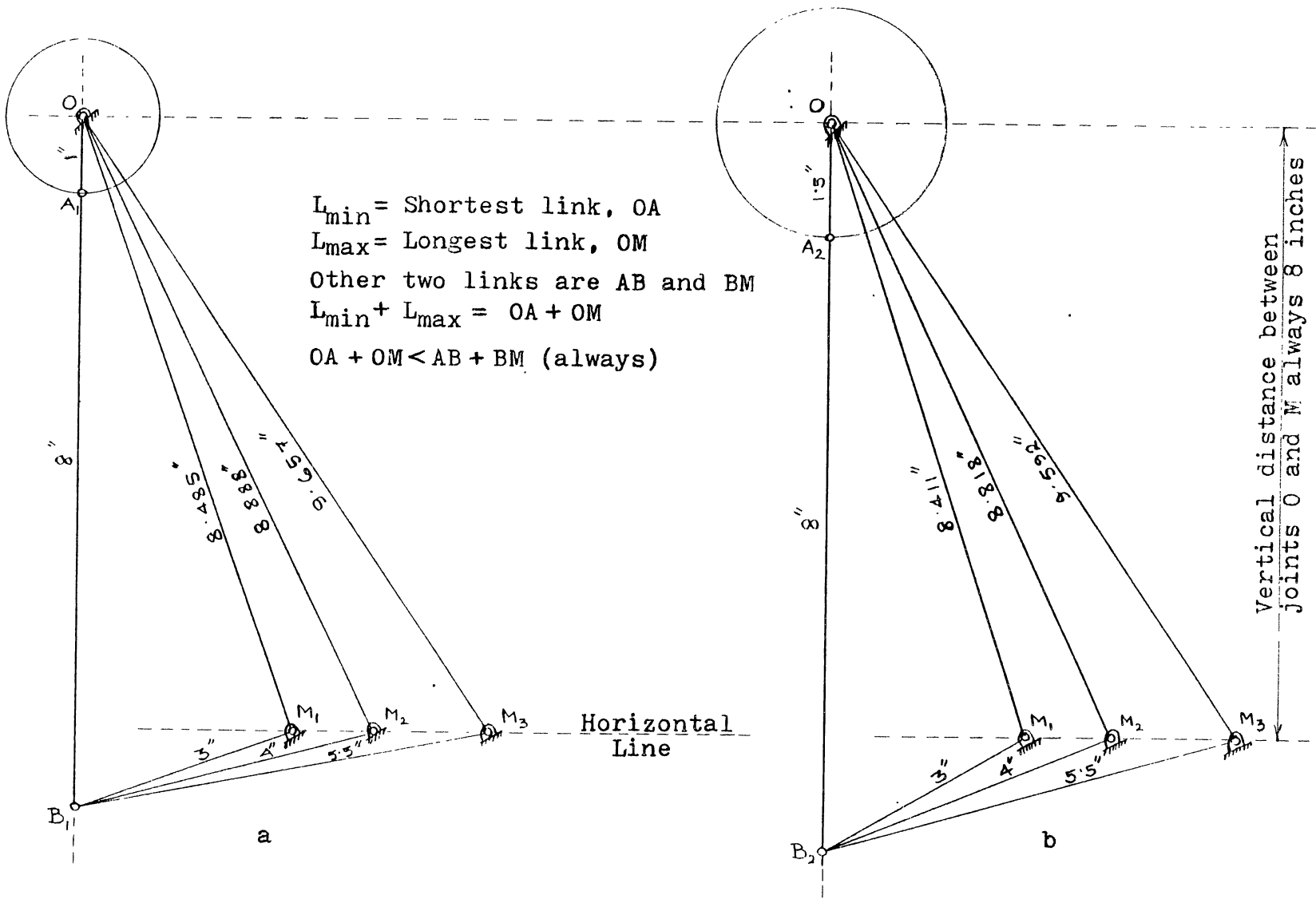
c. If the conditions other than those those described in a or b are satisfied, then the four-link mechanism is a rocker-rocker mechanism.

2. A four-link mechanism belongs to Class II mechanisms provided the sum of the lengths of the shortest and longest links is greater than the sum of the lengths of its other two links. A Class II mechanism is always a rocker-rocker mechanism.

Thus, in the mechanism shown in Figure 24, the input link, OA is always the shortest link of all the links even if we consider its maximum value of $1\frac{1}{2}$ ". Also, as shown in Figures 25a and b, the fixed link, OM is always the longest link of all the links. If L_{\min} and L_{\max} are the lengths of the shortest and longest links respectively, then, it is shown that the summation of the shortest and longest links is always less than the summation of the other two links, AB and BM.

Hence, the mechanism as designed and built is a Class I mechanism as per Grashoff's criteria and since the input link, OA is always the shortest link, we would call this mechanism a crank-rocker mechanism.

Figure 25. Crank-Rocker Mechanism. Respective Link Lengths are Shown for (a) One Inch and (b) One and Half Inch of Input Link Lengths. Fixed Link OM Varies as and When the Input and Output Link Lengths are Varied. The Mechanism as Shown is a Class I Mechanism.



L_{\min} = Shortest link, OA
 L_{\max} = Longest link, OM
 Other two links are AB and BM
 $L_{\min} + L_{\max} = OA + OM$
 $OA + OM < AB + BM$ (always)

Vertical distance between joints O and M always 8 inches

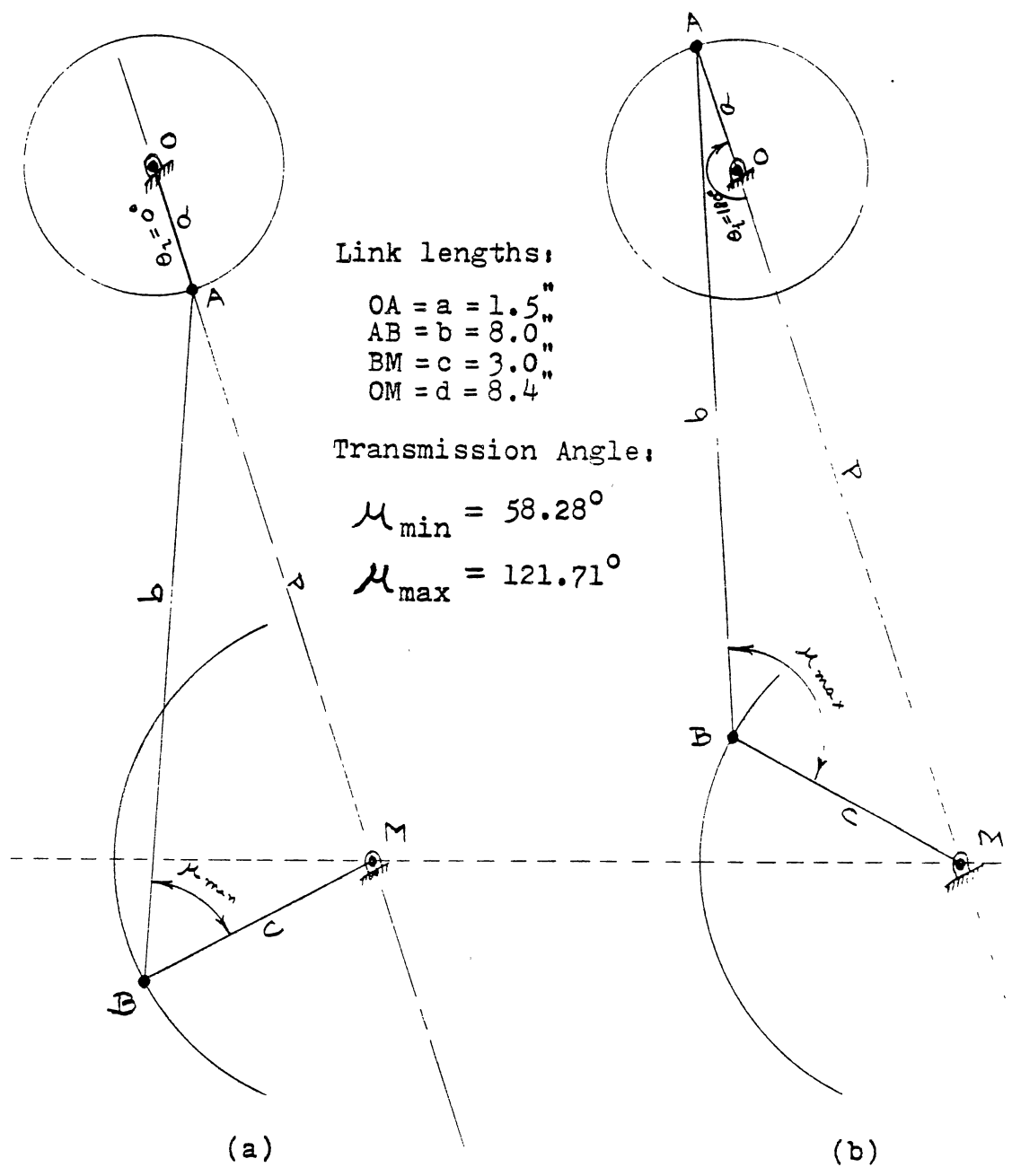
Horizontal Line

Minimum Transmission Angle

The included angle between coupler link, AB and output link, MB is known as transmission angle. As shown in Figure 26, when $\theta_i = 0^\circ$ or 180° , the transmission angle is either maximum or minimum. The quality of motion of a four-link mechanism is dependent upon the minimum value of transmission angle. The force transmission from the coupler link to the output link is most effective when the transmission angle is 90° . With the rotation of the input link, the value of the transmission angle also changes. For better transmission of motion, deviation of the transmission angle from its ideal value of 90° should be kept as minimum as possible. Transmission angle less than 15° create unusually higher acceleration, objectionable noise, and jerk at high speed.

For the mechanism being discussed here, the minimum transmission angle has been calculated by the application of cosine law and as shown in Figure 26, μ_{\min} is 58.28° for the maximum and minimum possible values of input and output link lengths respectively. This is the lowest minimum transmission angle in any combination of input and output link lengths, provided the machine is set up in such a way that in the two limiting positions, as will be discussed below and shown in Figure 24, the input link and coupler link both should remain on a straight vertical line.

Figure 26. Transmission Angle of Four-Link Mechanism. (a) Minimum and (b) Maximum Values of Transmission Angle When Input and Output Link Lengths are Set to Their Maximum and Minimum Values of 1.5 Inches and 3.0 Inches Respectively.



Limit Positions and Angle of Oscillation

In a four-link mechanism, the limit position for a output link is defined as a position in which the interior angle between its coupler link and input link becomes either 360° or 180° . The mechanism as shown in Figure 24, has two limit positions OA_1B_1M and OA_2B_2M and the angle $\angle N_1MN_2$ is known as the angle of oscillation made by the output link in moving from one limit position to the other. The maximum angle of oscillation which one could get with the mechanism shown is 60° . This angle is obtained by setting the input and output link lengths to their maximum and minimum values respectively. The maximum input link length possible with the Fatigue Testing machine is $1\frac{1}{2}$ inch and the minimum output link length possible is 3 inches. In the limit positions, the pivot points O, A, and B lie on a straight line. In order to divide the total angle of oscillation into two equal halves of bending angle from the horizontal neutral line, it is necessary that in the limit positions, the straight line on which the points O, A, and B lie should also be a vertical line. The Fatigue Testing machine has been so designed as to have necessary fixtures for attaining this kind of limit positions when all the points O, A₁, B₁ and O, A₂, B₂ lie on the same vertical line and in the two limit positions the points B₁ and B₂ have the same amplitude from the horizontal neutral line. The total angle of oscillation $\angle N_1MN_2$ is thus divided into two equal

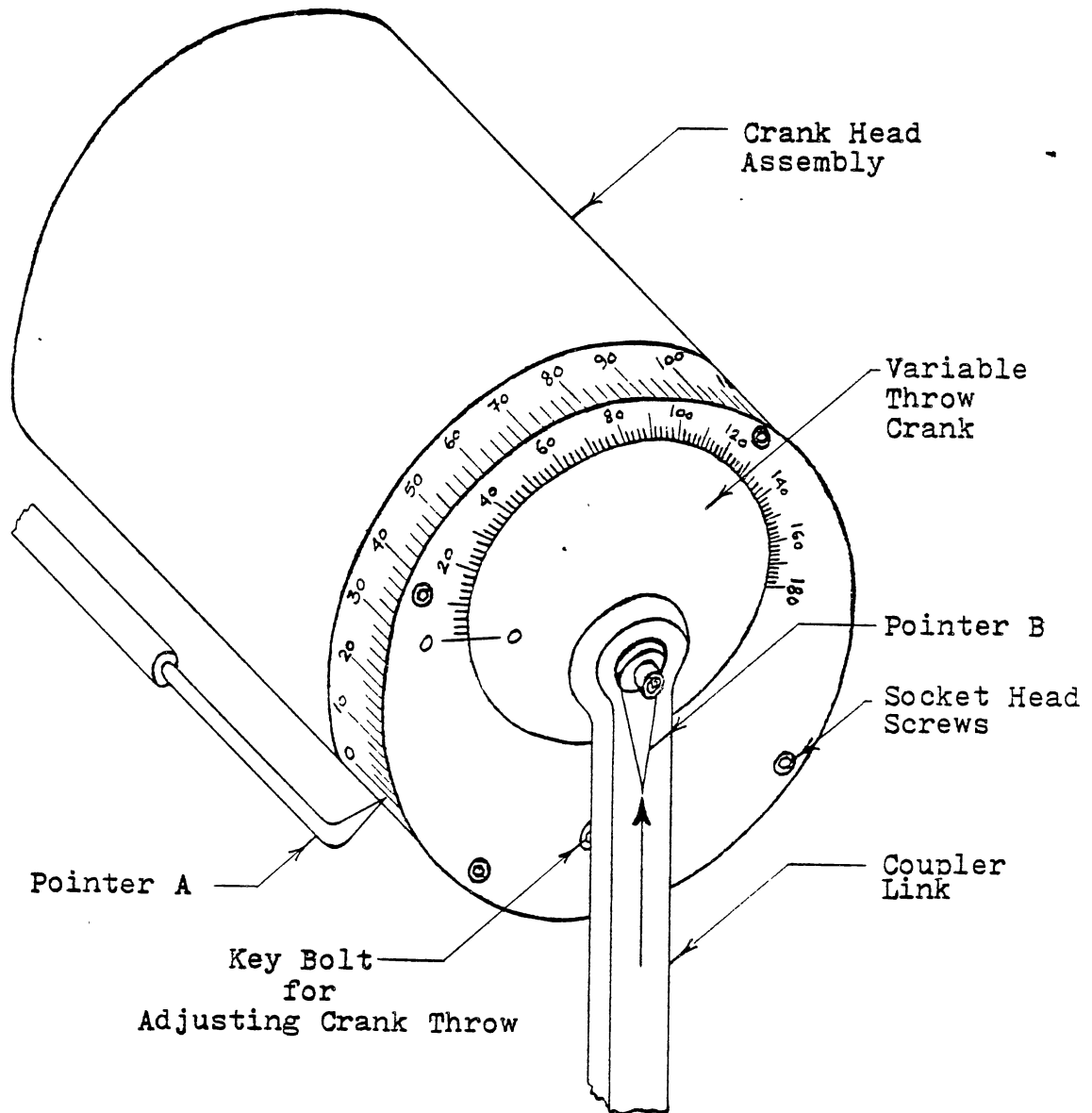
halves of bending angles $\angle N_1MN_0$ and $\angle N_2MN_0$ at which the specimen would be tested with its cyclic reversal of bending load.

C. Setting the Machine

The machine needs to be set up accurately before doing any kind of experiment on it. The correct setting of the machine is dependent on the positions of the pointers A and B (Figure 27). The pointer A should always be at the same height level as with the center of the crank head assembly. And the pointer B should point vertically downward when the variable throw crank is set to its zero value and the circular position of the crank head assembly is set to zero at the pointer A as shown in the figure. The positions of the pointers A and B are already set in the machine and should not be distributed by any means. However, the positions of the pointers should be checked from time to time and if any deviations found, they must be reset to their correct positions with the help of a height gauge and a spirit leveling gauge.

STEP 1: Loosen the two socket head bolts (pivot joints) on the two sides of the supporting frame on the horizontal vise and adjust the length of the output link (i.e., the driven link) by sliding it back and forth into the slots provided in both the arms of the output link. The length of the output link is decided upon by the required amount of bending angle at which the specimen

Figure 27. The Correct Positions of the Pointers. The Pointer A should always be at the Same Height as With the Center of the Crank Head Assembly. The Pointer B Should Point Vertically Downward When the Variable Throw Crank is Set to Its Zero Value and the Circular Position of the Crank Head Assembly is Set to Zero at the Pointer A.



would be tested. The length of the output link for any required bending angle could be found in Figure 28, corresponding to a fixed length of the crank throw. The linear scale engraved on the front arm of the output link, measures the effective length of the output link. The pivot position of the output link is then fixed by tightening the bolts of the pivot joints.

STEP 2: Loosen the four socket head screws around the face of the crank. Figure 29 gives the angle through which the variable throw crank should be eccentrically rotated for any desired crank length corresponding to the required bending angle already decided upon in Step 1. Bring the 0-mark on the variable throw crank to its required position by looking at the half circular scale as shown in Figure 30. Tighten the four screws and the crank length would remain fixed at this set value till it is not changed again.

STEP 3: Rotate the variable throw crank head assembly clockwise or anti-clockwise and at the pointer A, set it to an angular position, as given in Figure 29, corresponding to the crank length already fixed in Step 1. This position of the crank head assembly should not be disturbed till the completion of Step 4.

STEP 4: Loosen the clamping strips of the horizontal vise and slide the supporting frame back and forth until the pointer B, on the front end of the variable throw crank is pointing towards the arrow-head mark on the coupler link

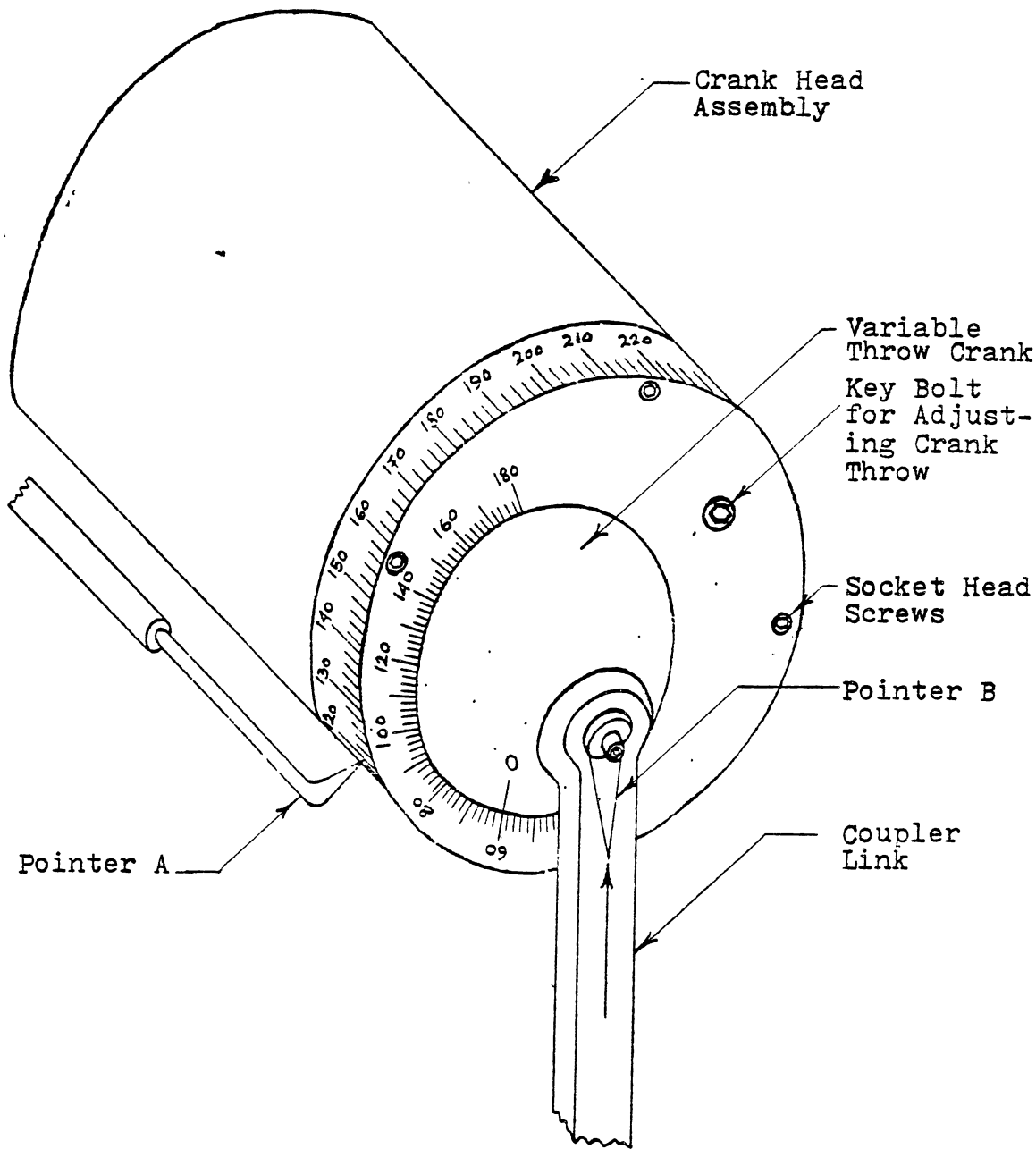
Figure 28. The Length of the Output Link for a Desired Bending Angle Corresponding to any Fixed Length of the Crank Throw (See also Figure 36, Appendix C for output link length calculation).

Reqd. Bending Angle	Length of the output link corresponding to any fixed length of the crank throw													
	1/16"	1/8"	1/4"	3/8"	1/2"	5/8"	3/4"	7/8"	1"	1 1/8"	1 1/4"	1 3/8"	1 1/2"	
1°	3.581													
2°		3.582												
3°			4.777											
4°			3.584	5.376										
5°			2.868	4.303	5.737									
6°				3.588	4.783	5.979								
7°				3.077	4.103	5.128								
8°					3.593	4.491	5.389							
9°					3.196	3.995	4.794	5.593						
10°					2.879	3.599	4.319	5.039	5.759					
11°						3.276	3.931	4.586	5.241	5.896				
12°						3.006	3.607	4.209	4.810	5.411				
13°							3.334	3.890	4.445	5.001	5.557			
14°							3.100	3.617	4.134	4.650	5.167	5.684		
15°							2.898	3.381	3.864	4.347	4.830	5.313	5.796	
16°								3.174	3.628	4.081	4.535	4.988	5.442	
17°								2.993	3.420	3.848	4.275	4.703	5.130	
18°								2.832	3.236	3.641	4.045	4.450	4.854	
19°									3.072	3.455	3.839	4.223	4.607	
20°									2.924	3.289	3.655	4.020	4.386	
21°										3.139	3.488	3.837	4.186	
22°										3.003	3.337	3.671	4.004	
23°										2.879	3.199	3.520	3.839	
24°											3.073	3.381	3.688	
25°											2.958	3.254	3.549	
26°											2.851	3.137	3.422	
27°												3.029	3.304	
28°												2.929	3.195	
29°												2.836	3.094	
30°													3.000	

Figure 29. Eccentric Rotation of the Variable Throw Crank Corresponding to Different Values of Crank Throw (i.e. Input Link Length). Also Gives the Angular Position of the Crank Head Assembly Required for Setting the Machine. (See also Figure 37, Appendix D).

Crank Throw a	Eccentric Rotation of the Variable Throw Crank $\phi = 2\text{Sin}^{-1}\left(\frac{a}{1.5}\right)$	Angular Position of the Crank Head Assembly $\psi = 180^\circ - \left(\frac{180^\circ - \phi}{2}\right)$
1/8"	9.56°	94.78°
1/4"	19.19°	99.60°
3/8"	28.96°	104.48°
1/2"	38.94°	109.47°
5/8"	49.25°	114.63°
3/4"	60.00°	120.00°
7/8"	71.37°	125.69°
1"	83.62°	131.81°
1 1/8"	97.18°	138.59°
1 1/4"	112.89°	146.45°
1 3/8"	132.89°	156.45°
1 1/2"	180.00°	0 or 180°

Figure 30. Setting the Machine for a Crank (or Input Link) Length of $3/4$ ". The Pointer B Should Point Directly Towards the Arrowhead Mark on the Coupler Link, When the Crank Head Assembly is Set to Its Required Angular Position at Pointer A as Shown in Figure 29.



(Figure 30). Clamp the horizontal vise tight at this position.

The four-link mechanism as set by the above procedure, will now have all its three points O, A, and B on the same vertical line in its two limiting positions as shown in Figure 24. The output link will now oscillate up and down with a constant amplitude from the horizontal neutral line.

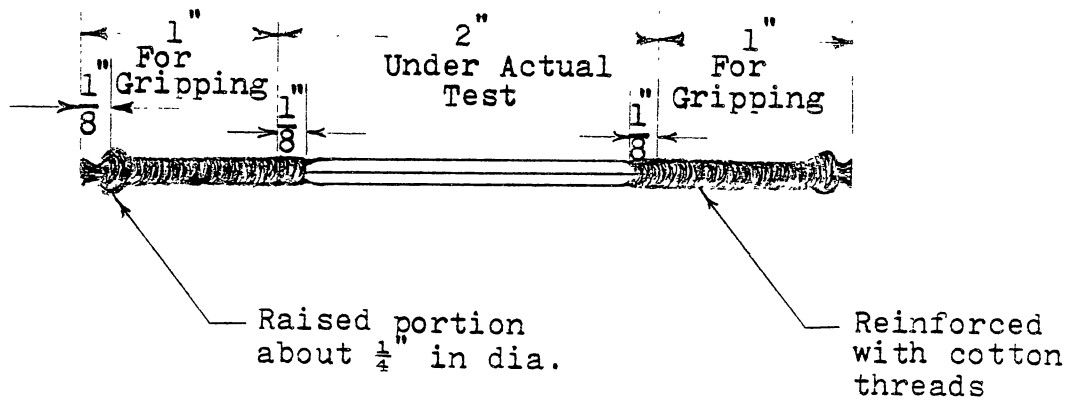
The machine is now completely set for the required amount of bending angle at which the specimen would undergo the cyclic testing.

D. Preparation and Fixing the Specimen

The fixture for mounting the specimen has been made to accommodate both a shorter and longer length of specimen. Natural ligaments are not very long, but, the artificial ligaments could be cut into various sizes of lengths to fit into the machine. However, for comparative reasons, the artificial ligaments also should be cut into smaller lengths. A length of four inches has been fixed here for doing the subsequent fatigue testing. Of the total length of the specimen, two inches would be required for gripping at the two ends and the other two inches would be under actual test.

In order to prevent any possible slippage of the specimen in the gripper, the ends of the specimen are wound and reinforced with good quality cotton threads as shown in Figure 31. Near the ends, a small portion is raised in

Figure 31. Minimum Size of the Specimen. The Ends are Reinforced and Raised by Winding Cotton Threads.



diameter with enough number of turns of cotton threads so that the ends of the specimen will get locked into the inner ends of the grippers. To prevent failure of the specimen at the ends of the the grippers due to twisting effect of the torsional load, full $1\frac{1}{8}$ " portion from both the ends are tightly wound with cotton threads and covered fully as shown in Figure 31.

Before inserting the ends of the specimen into the grippers, the respective positions of the grippers are to be set approximately by moving the spindles holding the grippers, backward or forward as needed. Apart from the screw motion, the gripper on the left has a linear movement of $\frac{3}{8}$ " between the two dead ends. Normally, when the specimen is not loaded, it is set to the left hand dead position by the spring force. When the specimen is loaded more than 25 pounds in tension, the spindle is set to its right hand dead position. This free motion of the spindle is to be accounted for while setting the position of the left hand gripper from the vertical roller support at which point the specimen would be bended. A minimum distance of $1\frac{3}{8}$ " from roller support is taken so that when the specimen is pre-loaded in tension, the front end of the left hand gripper comes to a dead end position, one inch from the supporting rollers. At this point, the jaws are opened and one end of the specimen with the raised portion is inserted into the left hand gripper. The jaws are then closed

loosely and keeping some tension on the specimen by hand the jaws are tightened fully.

In order to insert the other end of the specimen, the right hand gripper is brought forward by rotating the spindle on the knurled sliding nut at the end of the output link. The end of the specimen with the raised portion is slipped inside the gripper and the jaws are closed lightly as before. Some tension is applied to the specimen by pulling the spindle from the right end and turning the knurled sliding nut forward until it engages firmly on its seat. The jaws of the right hand gripper are then tightened fully. After the grippers are tightened, the specimen should be checked for any twisting along its length between the grippers. If there is any, the right hand spindle should be rotated clockwise or anti-clockwise as needed, holding the sliding nut. When the specimen is fully straightened, the sliding nut should be set firmly back to its seat, holding the spindle in its new position. The top vertical roller support should now be brought down to meet the bottom vertical roller support, which is already set and locked in its correct position. Bending of the specimen would occur at the inner groove between these two supporting rollers with an equal cyclic bending reversals. The specimen is now ready for pre-loading it, first in tension, and then in torsion.

E. Loading the Specimen

The specimen could be pre-loaded in tension and torsion as dictated by the requirements of the experiment for fatigue testing. Pre-loading the specimen in tension is done by putting dead weights on the weight pan suspended on a steel rope which is being carried through a pulley and hooked to the end of the right hand spindle. As more and more weights are put on the weight pan, the specimen would be elongated and the sliding nut would come apart from its seat. After adding the required amounts of weight, the sliding nut is turned back to its seat, holding the spindle by a wrench at the end. A locking nut is provided with the spindle which should now be turned to lock the spindle at this position. The steel rope with the dead weights may then be unhooked and taken off from the spindle. Since the spindles are locked in their respective positions, the specimen would remain in proper tension during the on-going of the experiment as long as the specimen does not tear off due to fatigue failure.

Pre-torsional load is applied to the specimen with the help of a torque transducer fixed on a sliding base at the left. A lever arm is fixed to the left of the rotor shaft and the torque may be applied either manually or through Instron Testing machine. In the later case, the lever arm is connected to the Instron Testing machine by some linkages. The digital voltage meter which is connected to the

torque transducer, reads the amount of torque applied in terms of voltage. A calibration chart is prepared with known amount of torque which then gives the torque equivalence of the voltage output shown on the meter during the actual experiment. The position of the lever arm is kept fixed at a pre-determined amount of torque to be applied to the specimen. The specimen is now pre-loaded both in tension and torsion.

The fatigue testing may now be started by switching on the machine and setting the variable speed motor at a pre-determined number of cycles per second. The specimen will now undergo stress reversals at its bending point until failure. The bending angle at which the specimen is bent upon is already set into the machine. The total number of stress reversals may then be read from the cycle counter.

CHAPTER VII

FATIGUE TESTING, RESULTS, AND DISCUSSION

This chapter presents the actual fatigue testing of the ligament prosthesis. The specimen was prepared in four inches of length from the original size of an artificial dacron ligament. In order to prevent possible slippage of the specimen in the grippers due to tensile load and also to prevent failure at the ends of the grippers due to twisting effect of the torsional load, the ends of the specimen were tightly wound with cotton threads as described in the previous chapter.

Before mounting the specimen, the machine was set for 30° bending angle at which the specimen would be tested. The bending arm had its total 60° angle of oscillation and during each revolution of the driving shaft, the specimen would undergo a complete reversal of bending load from maximum tension to maximum compression. The specimen was then mounted properly on the machine and pre-loaded in tension and torsion.

A tensile load of 100 lbs was applied to the specimen by putting equal amounts of dead weights on the weight pan and then locking the device at this load. A pre-torsional load of 3 in-lbs was then applied to the specimen through

the torque transducer. The machine was then set ready for cyclic testing in bending with the above setting of pre-tensile and pre-torsional load.

The cycle counter was set to its zero reading and the machine was turned on to run at a speed of two complete revolutions per second. Each complete revolution thus imposed two load-unload bending cycles of opposite direction on the specimen. The experiment was carried out in room temperature. After 200,000 stress reversals, the test was stopped and no sign of failure was observed in the specimen. The specimen was then put to a tensile test until failure in the Instron Testing machine. The load-elongation curve of this tensile test is shown in Figure 32.

A fresh specimen was also put to tensile test until failure and the load-elongation curve for this test is shown in Figure 33.

Results and Discussion

As already mentioned, no sign of failure could be seen visually in the fatigue test after 200,000 of load-unload bending cycles of opposite direction on the specimen. However, the tensile test on the fatigued implant showed a marked reduction in its ultimate strength. As shown in Figures 30 and 31, the ultimate tensile load for the fatigued specimen is 635 pounds and that for the fresh specimen is 731 pounds. This amounts to a reduction of 13.13 percent of the original strength of the implant.

Figure 32. Load-Elongation Curve of the Fatigued Implant
After 200,000 Stress Reversals.

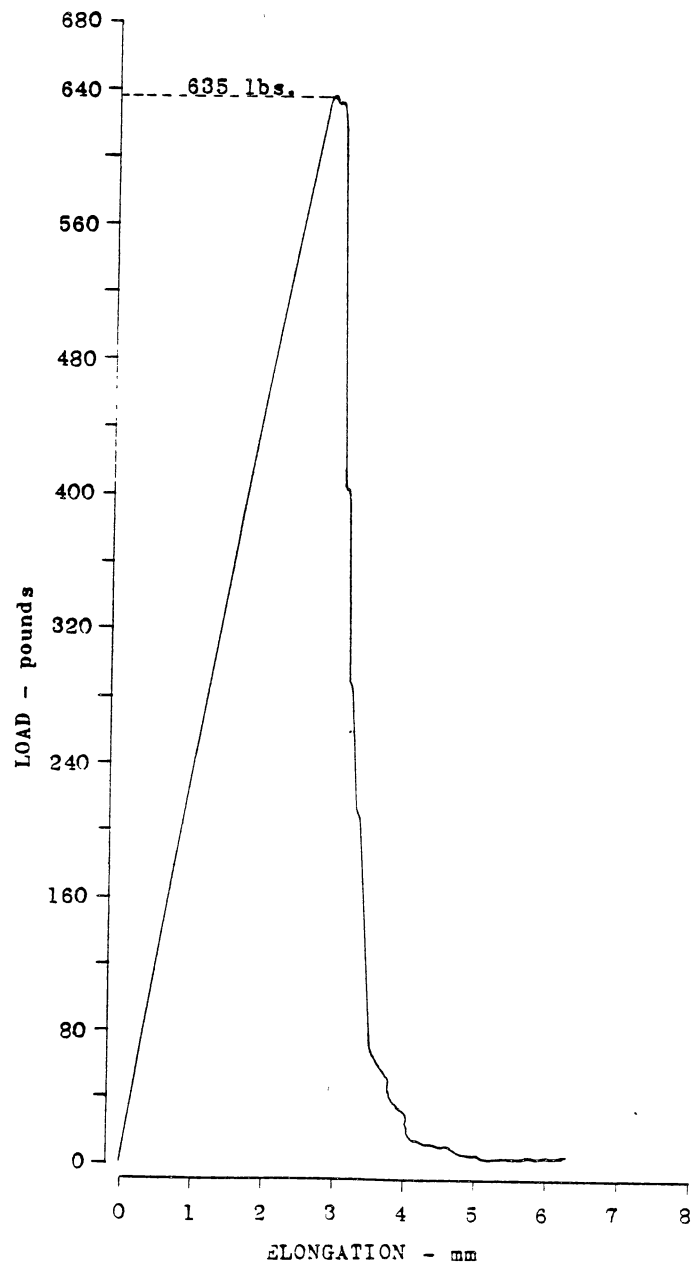
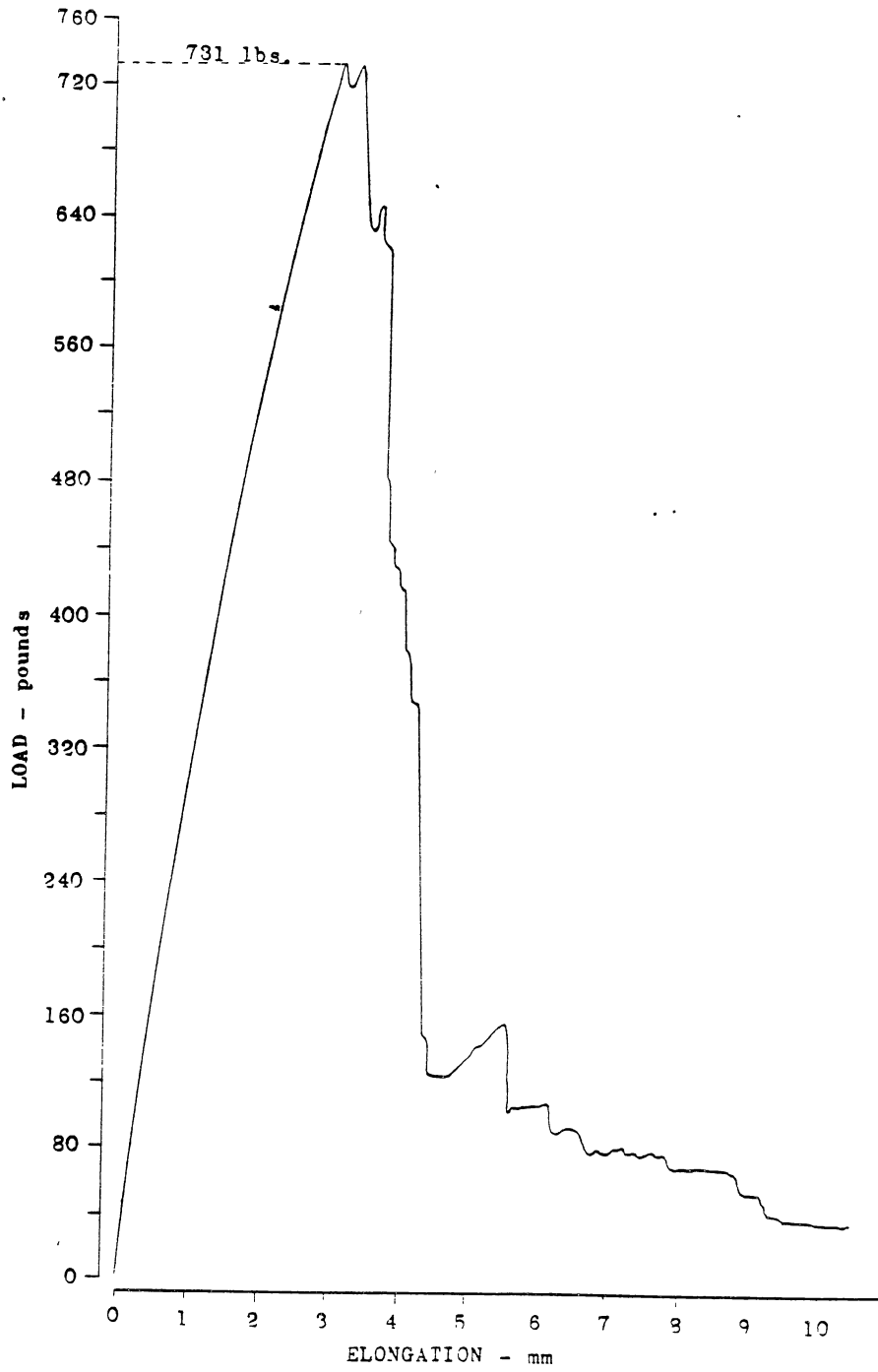


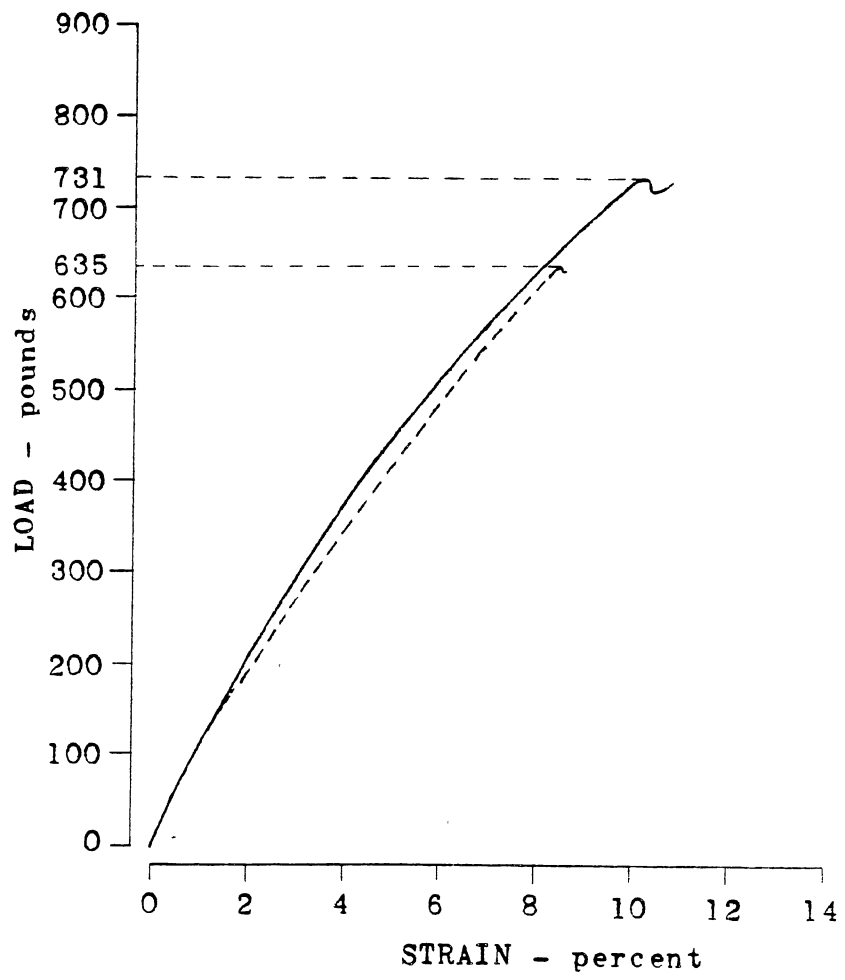
Figure 33. Load-Elongation Curve of Previously Unstressed
Dacron Ligament Implant.



Load-strain diagrams as obtained from the load-elongation curves are shown in Figure 34. As could be seen, the ultimate failure points are well defined in both the fresh and fatigued implant. However, the elastic range of the curves do not show a perfect linear relationship between load and strain. As evident from the load-elongation curves and also from the physical appearance of the specimens after the completion of the tensile tests, the fatigued implant showed a sudden and sharp decrease in load compared to the fresh implant after the final rupture of the specimens. The fatigued implant showed a clean breaking apart sign at the point where fatigue stress in bending was applied. As compared to this, the fresh implant showed a gross disruption of the fibers at certain point within the effective length of the specimen. At the point of ultimate failure, the fatigued implant had a total elongation of 10.83 percent as compared to 12.80 percent for the fresh implant. Structurally, after the final rupture, the fiber bundles of the fatigued implant disintegrated into its constituents of a thin, fluffy, and wavy material interwoven together into a fine mesh. However, some fibers in the fresh implant were still intact after the final rupture of the specimen. The gross disruption of the fibers showed an irregular pattern of failure around the neck of the specimen where the failure occurred.

Several data on the strength of human cruciate ligaments are now available. Kenedy et al. (31), reported a

Figure 34. Load-Strain Diagram of Dacron Ligament Implant. Tests Were Performed at a Strain Rate of 2 Centimeter Per Minute at Room Temperature. The Solid Line Indicates a Previously Unstressed Implant. The Interrupted Line Indicates an Implant After 200,000 Stress Reversals in Bending With the Implant Being Pre-loaded in 100 Pounds of Tension and 3 Inch-Pounds of Torsion.



maximum load of 63 ± 2.3 kg for the human anterior cruciate ligament at a strain rate of 50 cm/min. The ages of the specimen tested, ranged from 20 to 75 years with a mean of 62 years. No statistical correlation was found between age and failure strength. In contrast, Noyes and Grood (39) reported that a statistically significant correlation exists between age and the failing strength. The tensile tests as carried out by them on human anterior cruciate ligament showed a much higher value of maximum force for younger human than older human. The maximum load for specimens from older human (43 to 86 years) was 0.734 ± 0.266 kilo-newtons and for that of younger human (16 to 26 years), it was 1.73 ± 0.66 kilo-newtons. The maximum load values as given by Noyes and Grood are much higher than those given by Kenedy et al. However, the tensile tests on the artificial dacron ligament showed a much higher value of maximum load than any of the available data on real anterior cruciate ligament. The tensile test on the fatigued implant showed a reduction of maximum load from 731 to 635 pounds. Apparently, even after 200,000 stress reversals, the implant showed a greater value of maximum load than those of real ligament. A lower value of strain rate, 2 cm/min was applied during the tensile tests for both the fresh and fatigued implant. This rate is much lower than those applied by Kenedy et al. or Noyes and Grood while doing the tensile tests to failure for real ligaments. At higher

strain rates, one could expect much higher values of maximum loads for the dacron ligament implant.

Assessing the adequacy of the strength of the dacron ligament implant depends on the knowledge of the magnitudes of forces to which the anterior cruciate ligament or its replacement would be subjected during various activities. Only with the knowledge of these forces, can definitive statements be made as to the mechanical suitability of the implant. For normal gait, Morrison (35, 36, 37), used a combined experimental and analytical approach and calculated the forces acting on each cruciate ligament during various activities. This has already been described in Chapter V (Page 77), and the maximum force obtained, for anterior cruciate ligament was 445 newtons while descending stairs, and for posterior cruciate ligament was 1215 newtons while ascending stairs.

The maximum failure strength (731 lbs), as obtained in the tensile test of the fresh dacron ligament implant serves to demonstrate that the dacron implant has been designed to function under much higher stresses than those imposed by normal activities. However, the result of the fatigue test in bending with the implant pre-loaded in tension and torsion indicates a reduction in mechanical strength and this needs further investigation on the amount of tensile and torsional load to which the specimen was pre-loaded. A torsional load of 3 in-lbs as applied to the specimen

was probably a much higher value than the actual torsion to which the real ligaments in-vivo are subjected during normal activities. No quantitative report could however, be found, indicating the amount of torsion which any of the individual cruciate ligaments in the knee joint is subjected to during normal gait. The rotational movement of the tibia with respect to the femur in the horizontal plane reaches about 10° from full flexion to full extension. This small rotational movement occurs in the last few degrees of extension and usually, the maximum torque in normal gait develops when the knee is near full extension. As per Morrison's report (37), in normal knees the maximum torque developed during level walking is approximately 100 in-lbs and this happens at about 75 to 80 percent of the stance phase (i.e., 45 percent of a complete gait cycle). According to Morrison, a greater part of this inward torque at the knee during the stance phase is balanced by the tension produced in the oblique posterior fibers of the medial collateral ligament. Also that the tension in this ligament required to produce equilibrium would not significantly increase the values of the joint force as the force in the posterior cruciate which is also in tension at this part of the cycle, would tend to be reduced in order to maintain the equilibrium of the forces in the anterior-posterior direction.

From the consideration of the above, it could be concluded that it is not the cruciate ligaments which are being subjected to any significant part of the increasing torque

acting at the knee during the stance phase of walking. However, any abnormal medial rotation of the tibia on the femur is prevented by both the collateral and the cruciate ligaments. At this stage, the anterior and the posterior cruciates act together by twisting on themselves to resist a part of the torque acting at the knee.

In view of the above, 3 in-lbs of pre-torsional load applied to the dacron ligament implant during the fatigue test was probably a much higher value which in combination with 100 lbs of pre-tensile load, ultimately affected the mechanical strength of the ligament implant. However, after 200,000 stress reversals, the ultimate failure strength of the dacron implant was still high enough as compared to the maximum load which the human anterior cruciate ligament is subjected to during normal activities. But no definitive statements can now be made about the mechanical suitability of the dacron implant with respect to its fatigue strength. The fatigue test which continued for 200,000 stress reversals is considered to be only a preliminary investigation on the mechanical strength of this dacron ligament implant. Further investigation should be made on the nature and magnitude of the forces which the real ligaments in-vivo are subjected to during normal activities, and with a more precise knowledge of these forces, a prolonged fatigue experiment (up to a minimum of 2,000,000 stress reversals) should be carried out on the dacron implant. Only then a definitive statement could be made as to the adequacy of

the strength of the dacron implant when used as permanent prosthetic replacement of the cruciate ligaments in the human knee joint.

CHAPTER VIII

SUMMARY AND CONCLUSION

The decron ligament implant when subjected to fatigue test in bending, showed a reduction of 13.13 percent of its ultimate failure strength after 200,000 stress reversals. As compared to millions of fatigue cycles which a ligament prosthesis must survive without any substantial change in mechanical strength, the dacron ligament implant, however, showed a substantial decrease in its mechanical strength. Apart from the inherent material properties, there are several other factors which could be attributed to the cause of reduction of mechanical strength of the dacron implant. The amount of pre-tensile load and the ligament twist associated with the pre-torsional load, the bending angle, the rate of stress reversals are some of these factors. As mentioned earlier, the magnitude and exact nature of these forces in-vivo in the human knee joint during normal gait is not yet fully known. However, the amount of ligament twist associated with the pre-torsional load as applied to specimen may be analysed here with respect to the ligament twist in-vivo in a knee joint during normal gait. When applying 3 in-lbs of pre-torsional load in the fatigue test, the specimen was twisted approximately

900° ($2\frac{1}{2}$ turns) about its longitudinal axis. This naturally is an abnormal figure since the rotational movement of the tibia with respect to the femur reaches about 10° from full flexion to full extension of the normal knee joint. Also, as reported by Maiya (33), when the tibia is subjected to high torsional load, it is the capsular and collateral ligaments which are damaged and not the cruciates which remain intact even after the capsular ligaments were completely torn.

The nature of the above work was largely experimental. But, sufficient number of tests could not be performed because of non-availability of enough number of specimens. Besides, in order to do the fatigue experiment with the combined loading system in tension, torsion, and bending, a specially designed Fatigue Testing machine had to be built. The total work involved was therefore, divided into three phases: first, designing the machine; second, building the machine; and third, doing the actual fatigue experiment on dacron ligament implant. A major portion of the work done was in designing and building the Fatigue Testing machine. It is anticipated that by building the Fatigue Testing machine, an experimental base has been set up, whereby many future experiments could be performed on any artificial or real ligaments which require cyclic loading in bending combined with pre-tensile and pre-torsional load.

However, as a preliminary investigation, the fatigue test performed on the dacron ligament implant showed some

definite result of some changes in its mechanical strength. The ultimate failure strength was reduced substantially after 200,000 stress reversals; yet the value was much higher than any of the reported values of the maximum load which the real cruciate ligaments in-vivo are subjected to during normal activities. No definitive conclusion could be made here as to the adequacy of the mechanical strength of the dacron ligament implant which should survive millions of fatigue cycles when used as a permanent prosthetic replacement in human knee joint. Prolonged fatigue tests up to a minimum of 2,000,000 stress reversals should be carried out on a number of specimens. And a further investigation should also be made towards the exact nature and magnitude of the forces which the real ligaments in-vivo are subjected to during normal activities.

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APPENDIXES

APPENDIX A

GLOSSARY

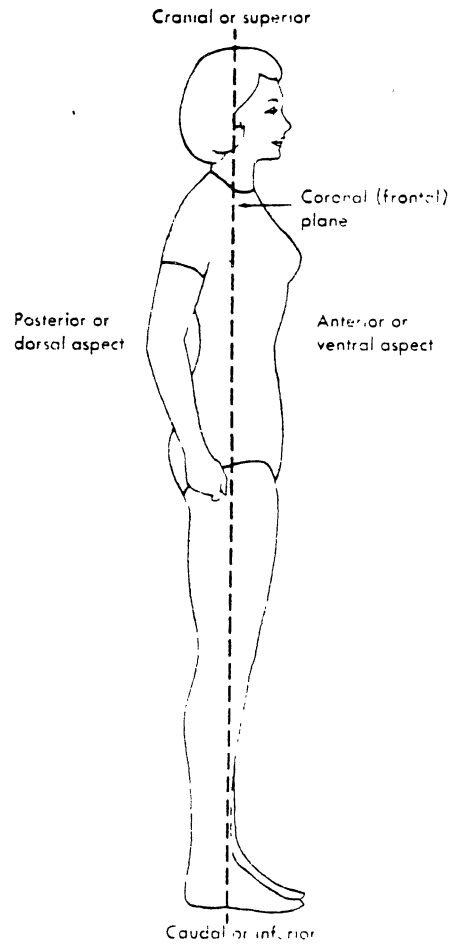
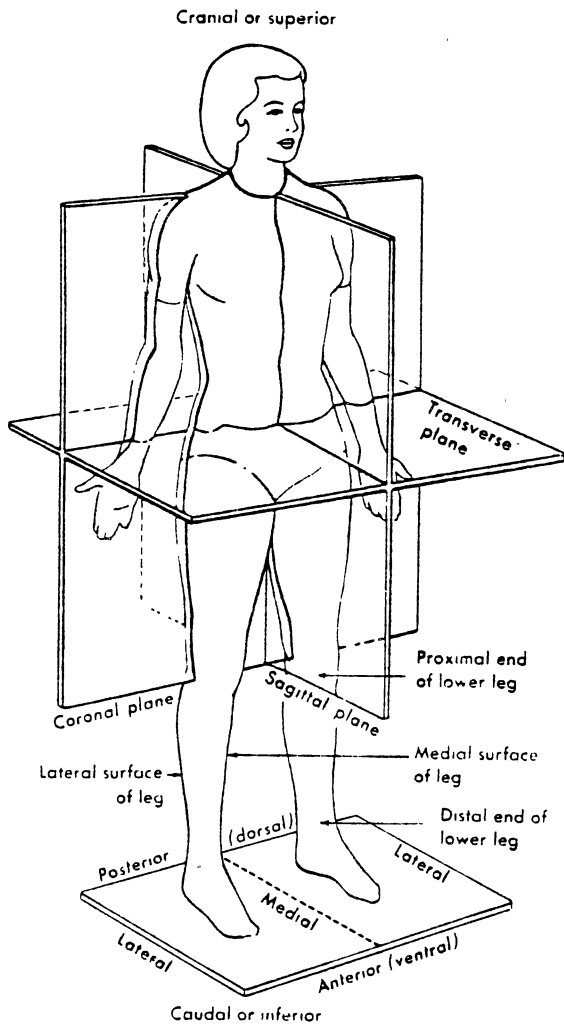
GLOSSARY

ABDUCT:	To move away from the middle of the body, on one of its parts.
ADDUCT:	To draw toward or beyond the median line of the body or its parts.
ANTERIOR:	In front of or in the front part of.
CARTILAGE:	A translucent elastic tissue characterized by its scanty blood supply.
CONDYLE:	A rounded surface at the extrimity of a bone.
FEMUR:	Thigh bone.
FIBULA:	Smaller of the two bones in the calf.
LATERAL:	On the side (outside) opposite of medial.
LIGAMENT:	A band or sheet of fibrous tissue connecting two or more bones, and providing the integrity of the joint.
MEDIAL:	Relating to the middle or center.
MENISCUS:	A crescent or disk shaped cartilage found in certain joints.
POSTERIOR:	Behind - in the back opposite of Anterior.
SYNOVIA:	A clean viscus fluid secreted by a synovial membrane.
TIBIA:	Larger of the two bones of the calf.
TRANSVERSE:	Crosswise - lying across the long axix of the body.
TUBERCLE:	A bump or protrusion such as the tibial eminence.
VALGUS:	Contact between lateral condyles is lost.
VARUS:	Contact between medial plateau is lost.

APPENDIX B

THREE PERPENDICULAR PLANES

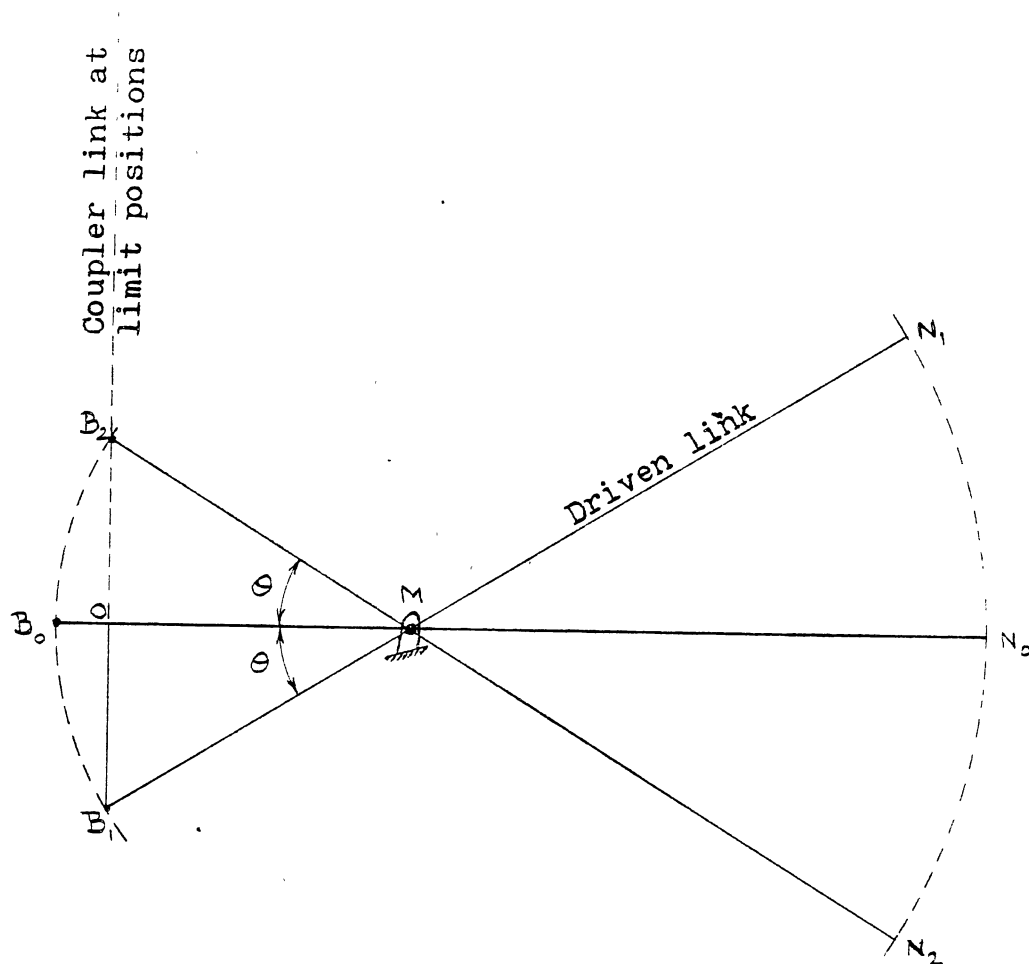
Figure 35. Anterior and Lateral View of the Human Figure.



APPENDIX C

OUTPUT LINK LENGTH CALCULATION
FOR DESIRED BENDING ANGLE

Figure 36. Two Limit Positions of the Four-link Crank-Rocker Mechanism.



θ = Required bending angle

$OB_1 = OB_2$ = Crank throw or input link length

$MB_1 = MB_2$ = Output link length

$$\sin \theta = \frac{OB_1}{MB_1} = \frac{\text{Crank throw}}{\text{Output link length}}$$

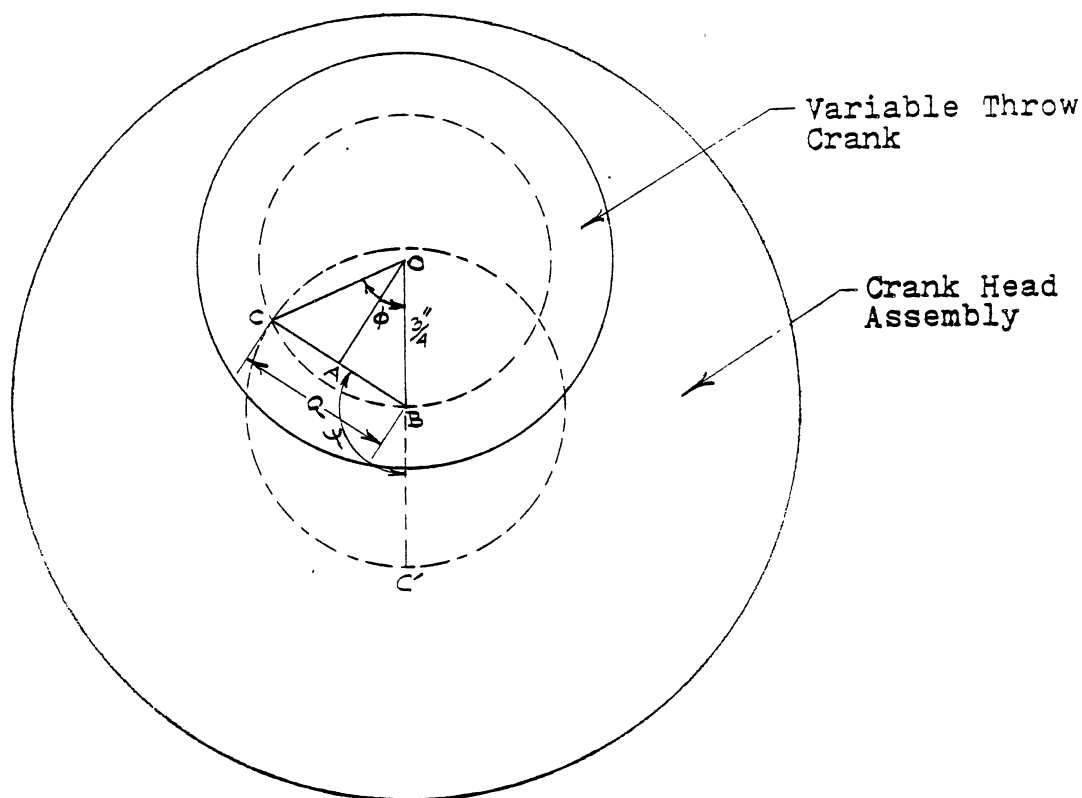
Hence,

$$\text{Output link length} = \frac{\text{Crank throw}}{\sin \theta}$$

APPENDIX D

ECCENTRIC ROTATION OF THE VARIABLE THROW
CRANK AND ANGULAR POSITION OF THE
CRANK HEAD ASSMEBLY

Figure 37. Rotation of the Variable Throw Crank and Crank Head Assembly Corresponding to a Particular Value of Crank Throw for Correct Setting of the Machine.



ϕ = Eccentric rotation of the variable throw crank

a = Required crank throw (i.e., input link length)

OB = Distance between the centers of the variable throw crank and crank-head assembly ($3/4$ inch constant)

γ = Angular position of the crank-head assembly

$$\sin\left(\frac{\phi}{2}\right) = \frac{AB}{OB} = \frac{a/2}{3/4} = \frac{a}{1.5}$$

Hence, $\phi = 2 \sin^{-1}\left(\frac{a}{1.5}\right)$ and $\gamma = 180^\circ - \left(\frac{180^\circ - \phi}{2}\right)$

APPENDIX E

FATIGUE TESTING MACHINE
AS DESIGNED AND BUILT

Figure 38. The Fatigue Testing Machine, When the Variable Throw Crank is Set to its Zero Value. The Crank Head Assembly Rotates but There is no Transmission of Motion to the Connecting Rod.

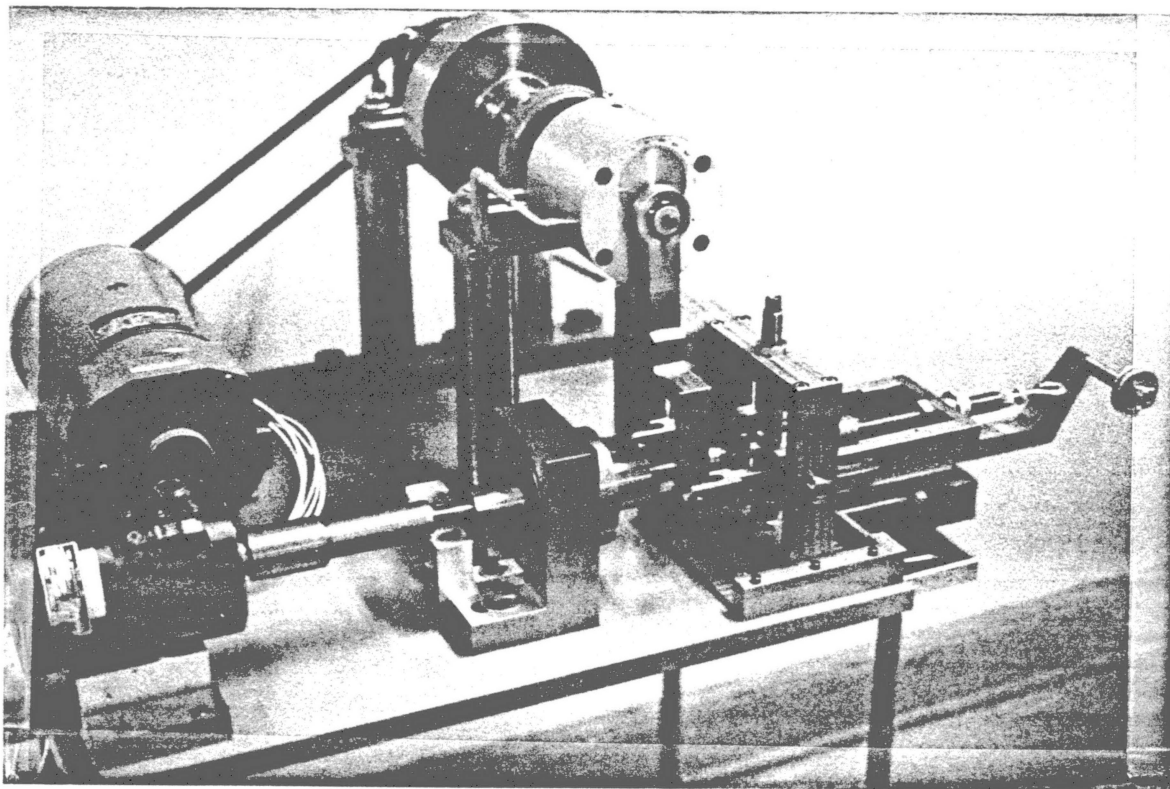


Figure 39. The Machine, When the Variable Throw Crank is Set to its Maximum Value of $1\frac{1}{2}$ inch, and the Output Link Length to 4 inches. The Bending Angle at This Position is 22 degrees, i.e., the Total Angle of Oscillation is 44 degrees.

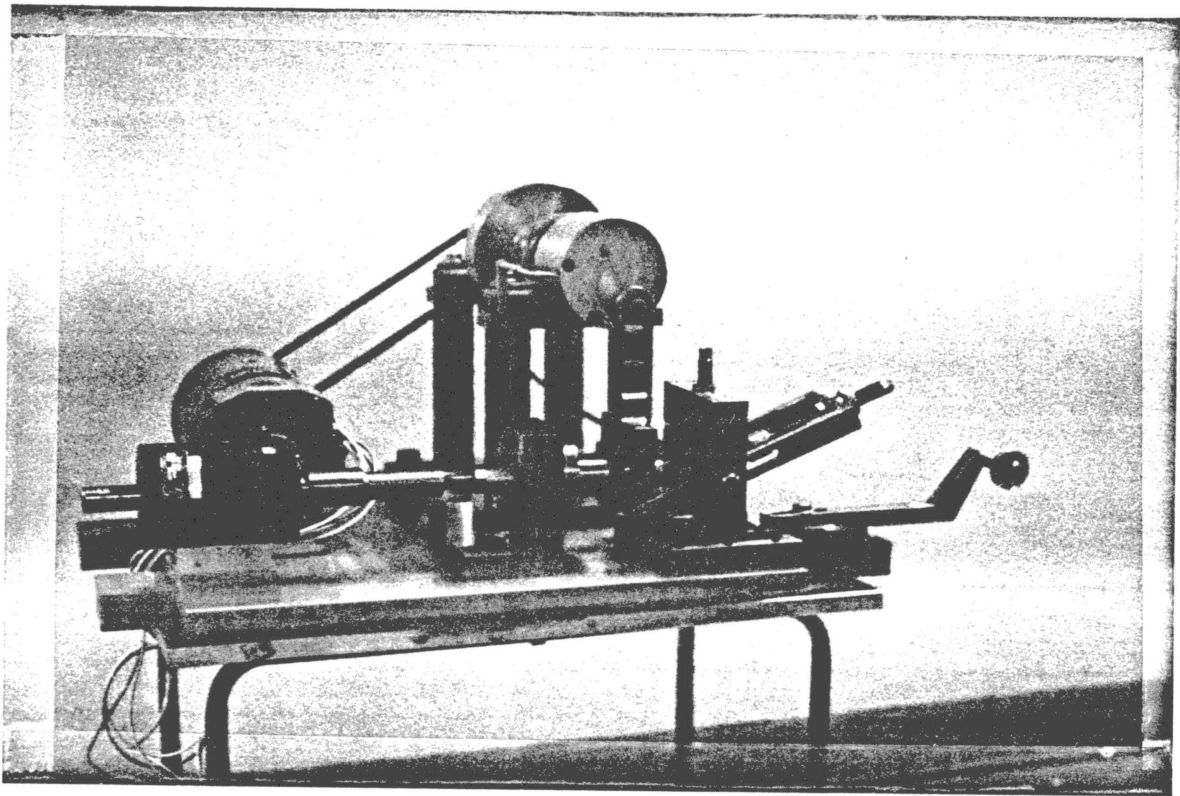


Figure 40. Loading the Specimen in Tension. This is done by Putting Dead Weights on the Weight Pan (not visible in dark shade), Suspended on a Steel Rope. The Rope is Taken off After Locking the Right Hand Spindle and the Specimen Remains Loaded at its Set Value.

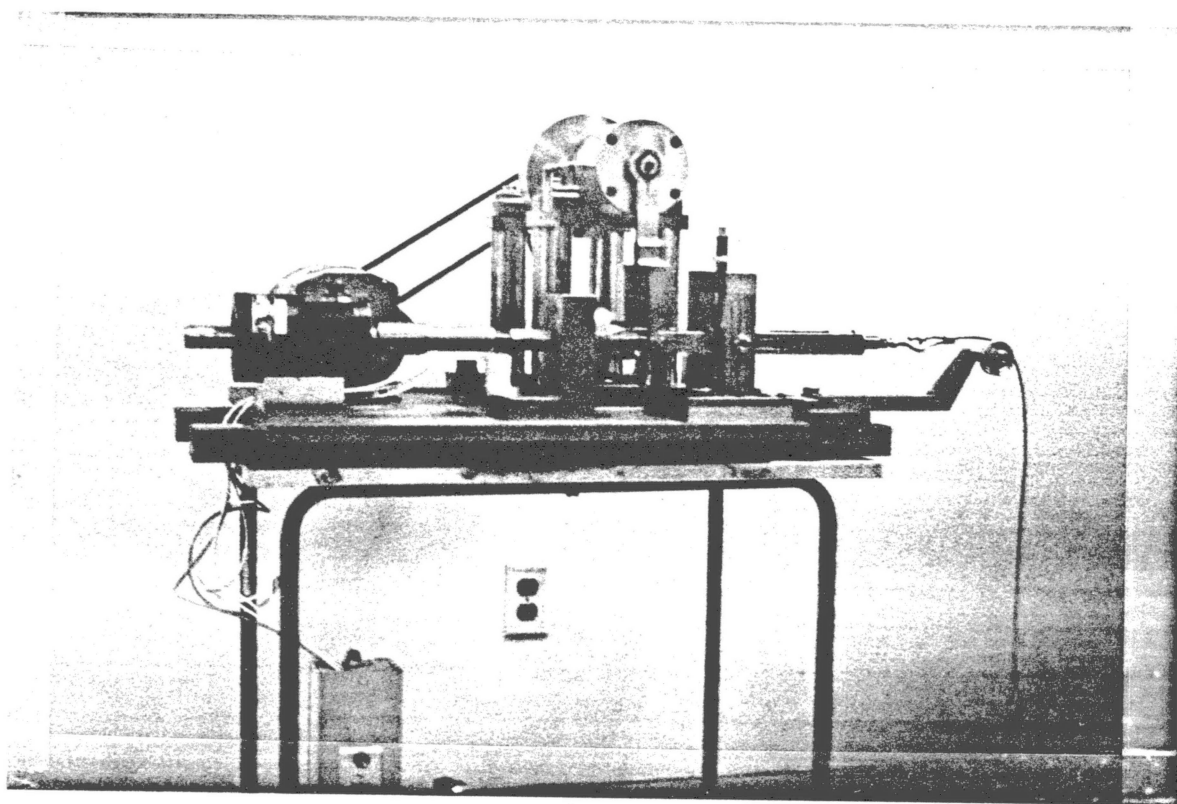
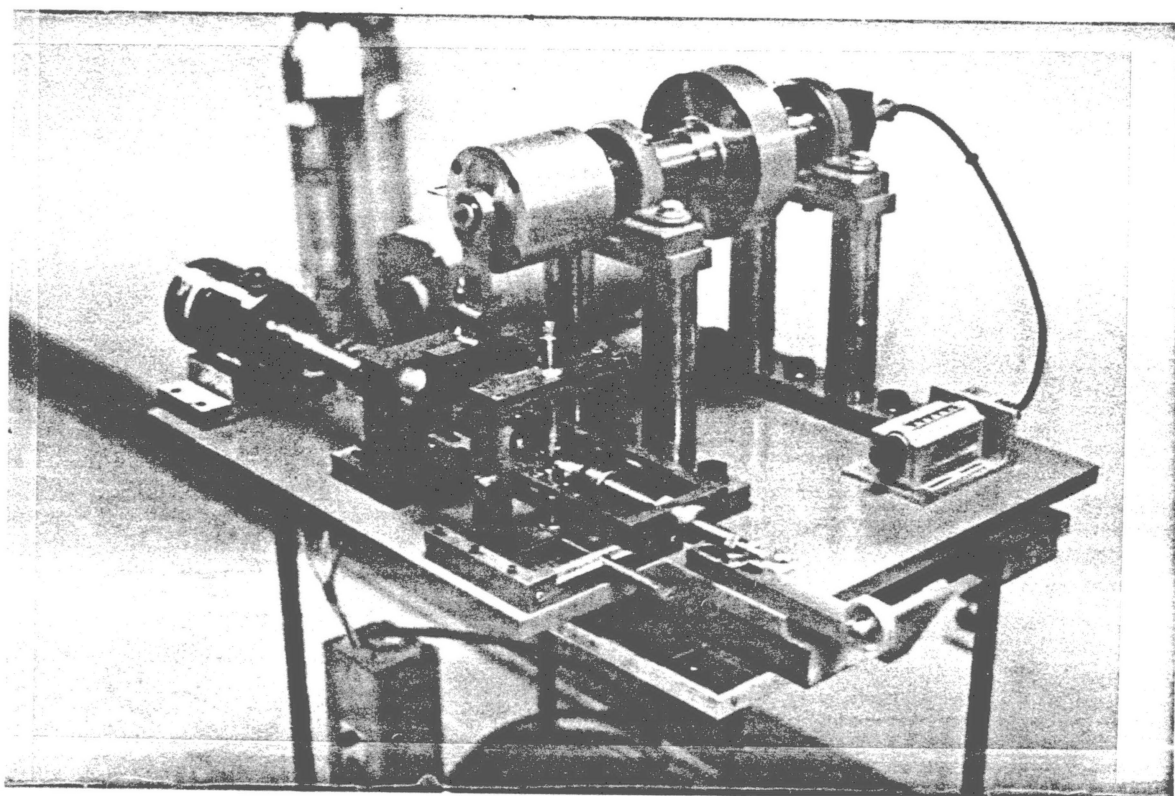


Figure 41. The Cycle Counter Which Reads the Total Number
of Cycles During and After the Fatigue Test.



VITA 2

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