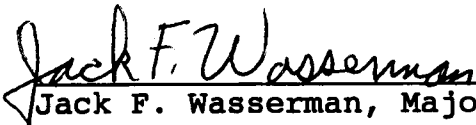
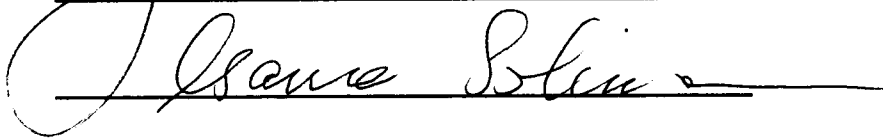


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**THE MECHANICAL BEHAVIOR OF THE
HUMAN FEMUR SUBJECTED TO
IMPACT LOADING**

**A Thesis
Presented for the
Master of Science
Degree
The University of Tennessee, Knoxville**

Guy V. Tucker

May 1991

DEDICATION

I would like to dedicate this thesis to my parents, Guy and Mary Frances Tucker, and to my best friend, Abigail Page. Without the love and support of these three, I could have accomplished nothing.

ACKNOWLEDGEMENTS

The research presented in this thesis was performed while working on the research project entitled "Dynamic Response of the Human Leg to Impact Loading." The project is sponsored by the Japan Automobile Manufacturers Association, Inc. This is an on-going project at the University of Tennessee with support from the University of Louisville.

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The contributions of my good friend Cindy Demarest and my laboratory partner, Tyler Kress, are greatly appreciated. And if it was not for the brilliant editorial, technical and financial help of Abigail Page and my parents, this thesis would be completely unintelligible.

ABSTRACT

The objective of this thesis was to perform a definitive study of the mechanical behavior of the human femur subjected to loading conditions similar to those experienced by a motorcyclist whose leg is impacted by an automobile. The results of this report will then be used in the development of artificial bones for crash dummies.

To meet this objective, nearly one hundred femurs and other long bones have been broken in a specially designed laboratory. The principle component of this laboratory is a machine which simulates an automobile/motorcycle collision. The testing was divided into four sections: (a) bending strength of the femur, (b) femurs under torsional loads, (c) femurs under axial loads, and (d) fresh tissue testing.

Given the number of different types of test performed in compiling this thesis, there are many results and conclusions. The femur appears stronger when impacted in the anterior-posterior direction than in the lateral-medial. In addition, the fractures produced by these a-p impacts provide interesting clinical information. It was found that even very small torsional preloads can greatly diminish the femur's breaking strength. Axially loading the femur allowed the mapping of the stress along the femur to

accurately predict fracture locations. Fresh bones tend to be stronger than embalmed bones, and impacted bones are stronger than slowly loaded bones.

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CHAPTER I

INTRODUCTION

A. SUPPORTING PROJECT

This thesis presents data on the impact response of the human femur under several loading conditions. It represents a portion of the research completed over the past five years at the University of Tennessee for the Japan Automobile Manufacturers Association (JAMA). The research has revolved around the high-speed impact of human, animal and artificial leg bones. Effective quantification of the damage to hard and soft tissue under loading conditions similar to that which occurs when an automobile impacts a human leg provides insight into protection mechanisms and provides input to computer and physical models.

This is the second Master's thesis reflecting research on the JAMA project. The first was written by Kress (1989) entitled *Mechanical Behavior of Lower Limbs in Response to Impact Loading: Facility Development and Initial Results*. Kress's thesis deals primarily with the response of the human tibia and will be cited often in this report.

B. OVERVIEW OF PAST RESEARCH

During the first year, the primary concern of the researchers was facility development. A pneumatically-operated machine was designed for impacting both intact legs and bare bones of humans and animals. A major component of the machine is a pressure tank which accelerates a cart down a guide-rail system. Various bumpers and impacting surfaces can be mounted on the cart. The cart travels at selected velocities ranging from 3.0 to 13.4 meters per second, and the velocity remains essentially constant throughout the impact.

After the testing facility was developed, the initial laboratory work involved cadavers. This testing provided insight into the mechanisms of failure of the lower-leg bones during impact loading. The types of fractures that result from conditions such as automobile/motorcycle and automobile/pedestrian collisions typically result in amputation due to severe neurologic and vascular damage. Previously, it had been argued that leg entrapment must be present for such a fracture to occur, but impact research with high-speed film documentation showed that this is not necessarily the case. Inertial fixation from the foot and the body provides an adequate support condition to produce the aforementioned damaging fractures (Kress, 1989).

The second year of research was spent developing force measurement capabilities. From these measurements, force/time history through bone fracture could be obtained. The motivation behind this research was to guide in the design of breakable limbs for impact test dummies. To satisfy this end, the impactor is fixed to the cart with slide pins in such a manner that the entire load felt by the bone is transmitted through a force transducer. Tests that year concentrated on the breaking strength of human tibias, the breaking strength of intact cadaver legs and the differences in preserved and living tissue.

During the third year, the author of this thesis joined the JAMA project. Construction of a full-service biomechanics laboratory was begun and tibial work was refined to separate the bone material properties and geometric considerations. Towards the end of year three, the testing of artificial bones, which were commissioned by the JAMA project, became possible. Initial results indicated that the development of artificial, breakable bones was a real possibility.

Fourth and fifth year research is the subject of this thesis.

C. RESEARCH OBJECTIVE

The ultimate goal of the JAMA project is to develop a crash-test dummy made of artificial, breakable bones. Following the comprehensive evaluation of the response of the human tibia under impact loading (Kress, 1989), it became clear that such information should be obtained for the other major bones of the body. The objective of this thesis, therefore, is to perform a definitive study on femur impact testing.

To meet this objective, nearly one hundred human femurs and other long bones have been broken. The testing can be divided into the four following sections:

1. Bending Strength of the Femur. Femurs were impacted at high-speed with the previously-mentioned cart. Tests included (a) lateral-medial loading and (b) anterior-posterior loading.
2. Femurs under Torsional Loads. Femurs were tested (a) under slowly-applied torsional loads and (b) under lateral-medial bending impacts with a simultaneously-applied torsional preload.
3. Femurs under Axial Loads. Femurs were tested (a) subjected to a low strain rate, axial load (compression) and (b) subjected to an axial impact or high strain rate load (knee impact).

4. Fresh Tissue Testing. Femurs and other long bones were impacted to determine the effects of embalming on the properties of the bones.

CHAPTER II

LITERATURE REVIEW

A. INTRODUCTION

The study of the human femur has been extensive. Living tissue, however, rarely behaves consistently by engineering standards. Therefore, there can never be too much study performed on one area of the body.

In order to fully understand the concepts presented in this thesis, three topics must be addressed. These are (1) the structural and anatomic characteristics of the femur, (2) the past work on mechanical behavior of femur, and (3) the reported relevance of this thesis.

B. FEMUR STRUCTURE AND ANATOMY

1. Structure

Bone is connective tissue. Connective tissue, a group of similarly specialized cells, is the support for various structures of the body. Bone's primary function is load carrying. The composition of bone is shown in Figure 1.

There exist two major forms of bone tissue: compact (or cortical) and cancellous (or trabecular). The outer surface of the bone is referred to as the periosteum, and

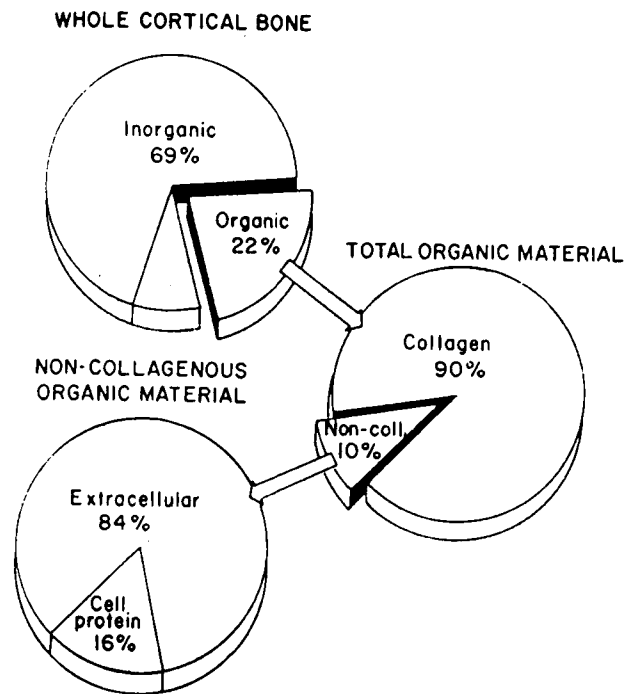


Figure 1. Distribution by weight of components of whole cortical bone (Park, 1984, p. 131).

the inner surface is referred to as the endosteum. These are illustrated on a femur in Figure 2.

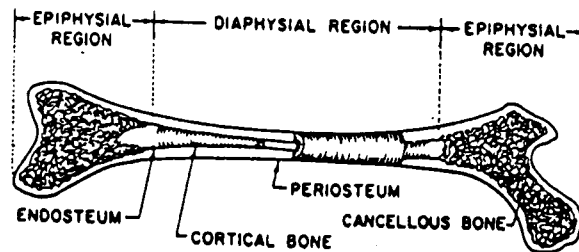


Figure 2. Illustration of femur showing different bone types and sizes (Covin, 1987, p. 2.2).

Bone is a composite material. Its basic structure is shown in Figure 3. Bone is the only hard connective tissue, owing to its extracellular, collagenous matrix, which is impregnated with hydroxyapatite, as shown in Figure 4. The basic unit of bone is the osteon or the Haversian system. In the center of the osteon is the Haversian canal housing an artery or vein. Volkmann's canals connect these blood vessels.

The composite materials of bone, therefore, are hydroxyapatite and collagen. The apatite crystals are nearly as strong and stiff as steel. Young's modulus for

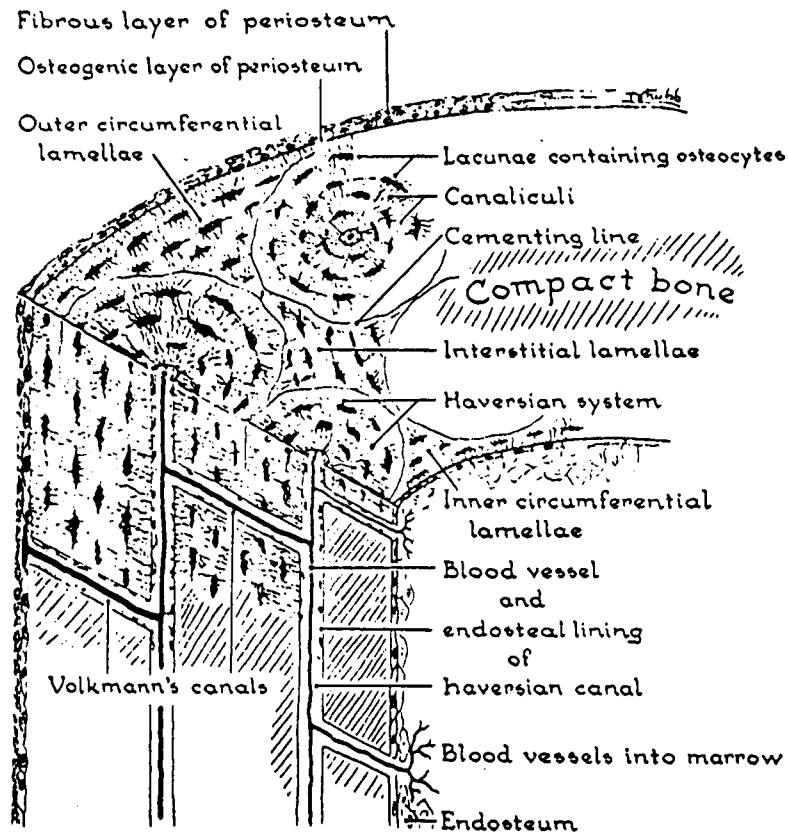


Figure 3. Structure of compact bone
(Fung, 1981, p. 388).

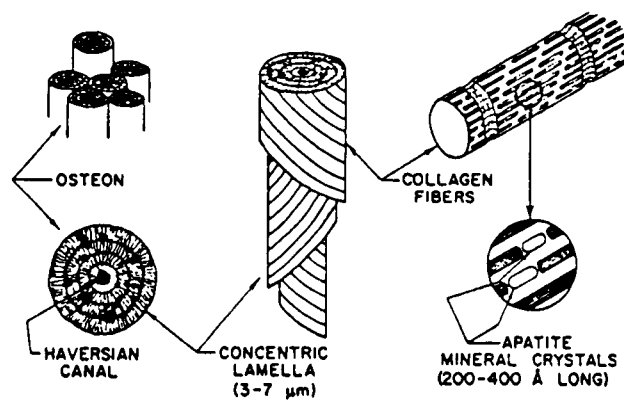


Figure 4. Detailed structure of an osteon
(Cowin, 1987, p. 2.3).

apatite is 165 GPa. Young's modulus for collagen, on the other hand, is 1.24 GPa. Bone's modulus falls in between these two at approximately 18 GPa, but the strength of bone is higher than the strength of either. The collagen prevents the apatite from brittle cracking, and the apatite prevents the collagen from yielding (Fung, 1981).

2. Anatomy

The bone and surface anatomy of the femur and hip can best be described in conjunction with Figures 5, 6, and 7. The hip is defined as the region between the iliac crest and the greater trochanter of the femur (Moore, 1985).

The femur is the longest, strongest and heaviest bone in the body (Moore, 1985). The femur is divided into three sections - the proximal end, the shaft, and the distal end. The proximal end is composed of the head, neck, greater trochanter, and lesser trochanter. The distal end broadens and articulates with the tibia and patella to form the knee joint. The shaft of the femur has a slight anterior bow and is narrowest at its midpoint. The cross-section is approximately circular through most of the shaft as can be seen in Figure 8 (Moore, 1985).

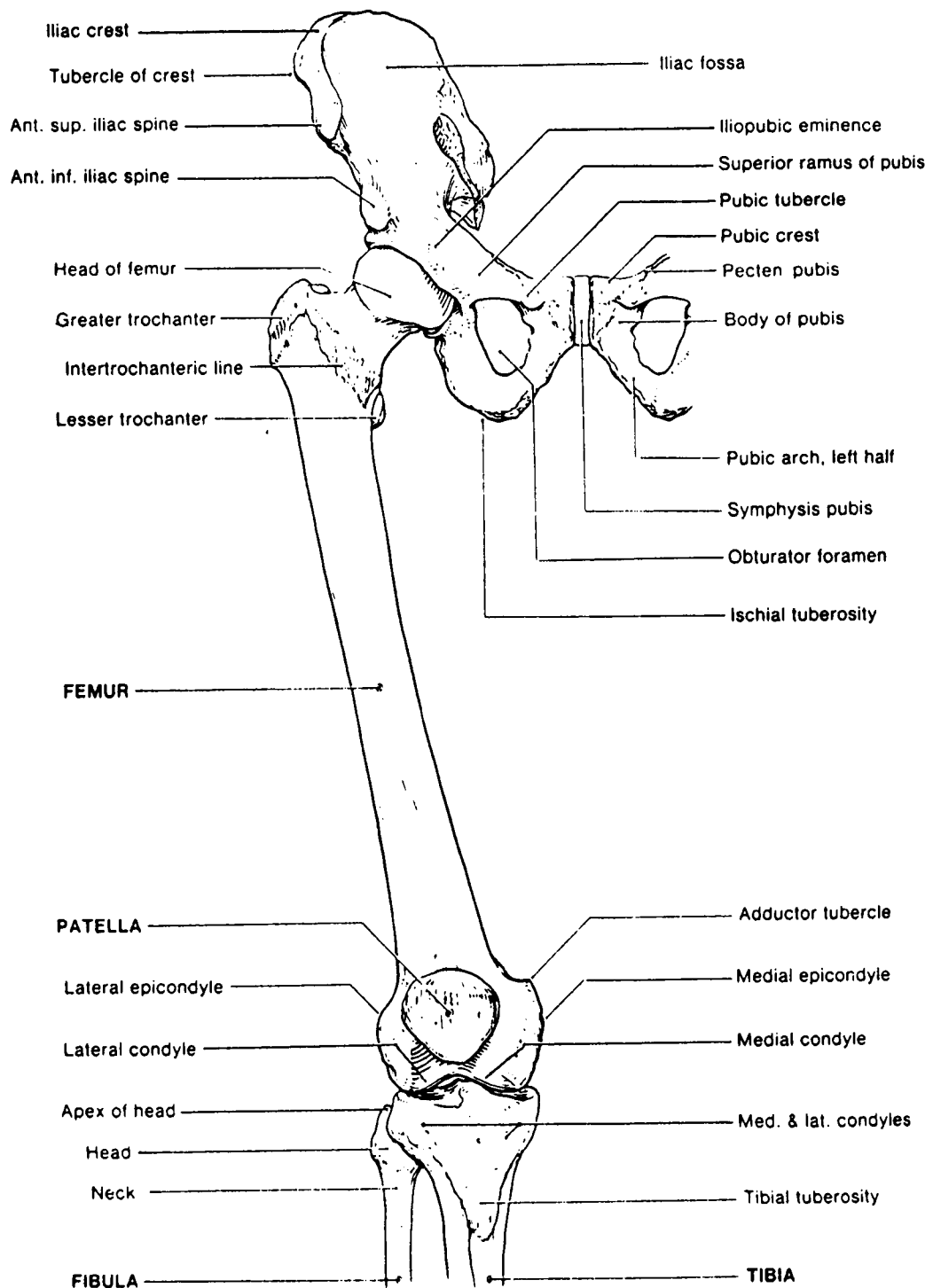


Figure 5. Anterior view of bones of lower limb (Moore, 1985, p. 398).

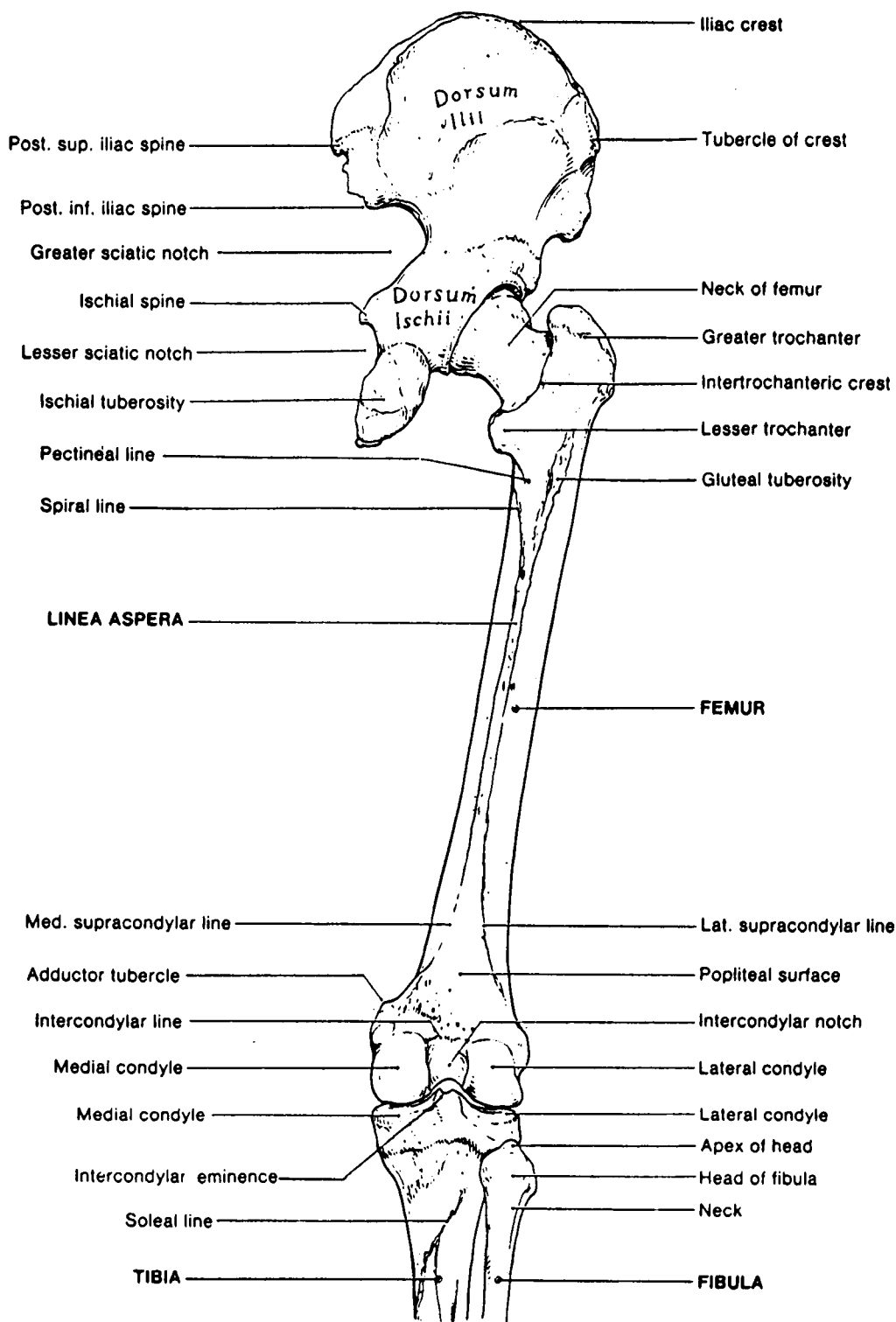


Figure 6. Posterior view of bones of lower limb (Moore, 1985, p. 399).

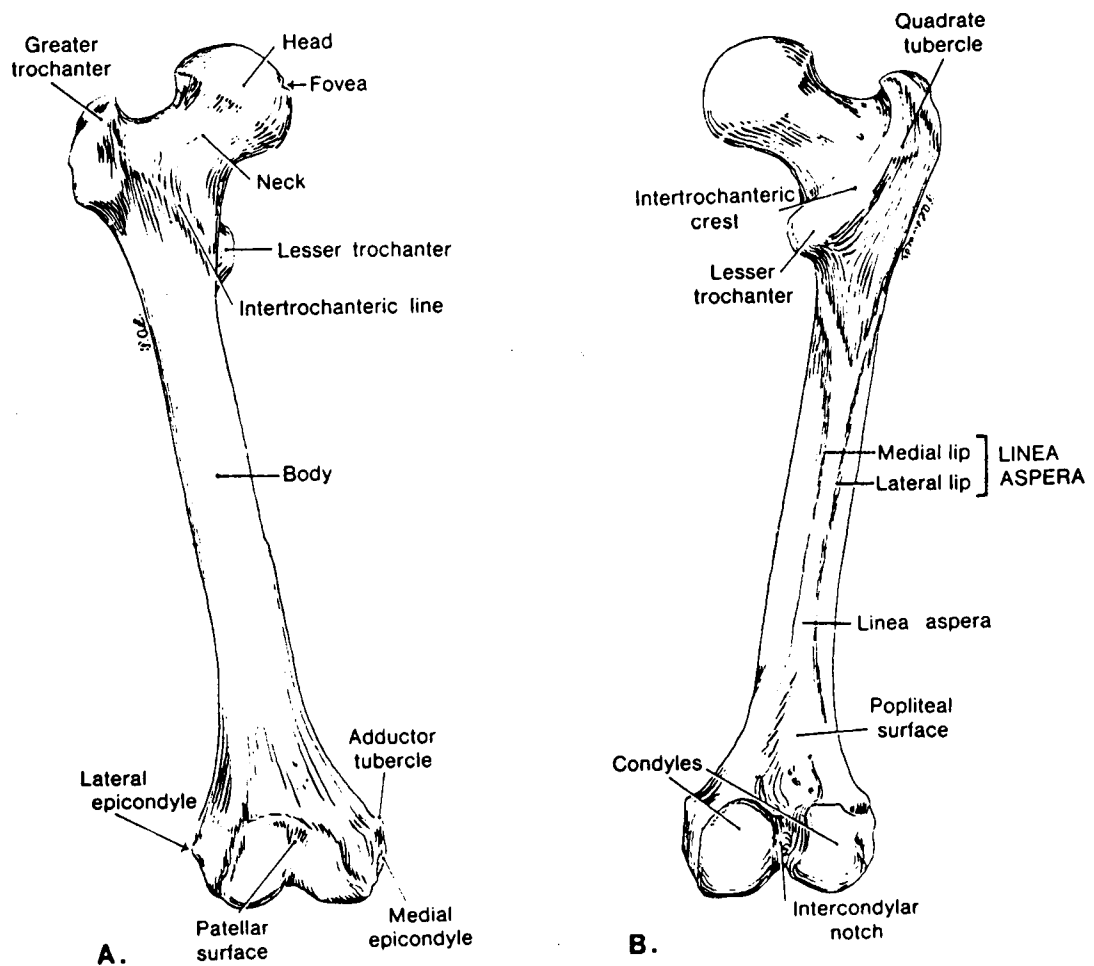


Figure 7. Right femur, A, anterior aspect, B, posterior aspect (Moore, 1985, p. 403).

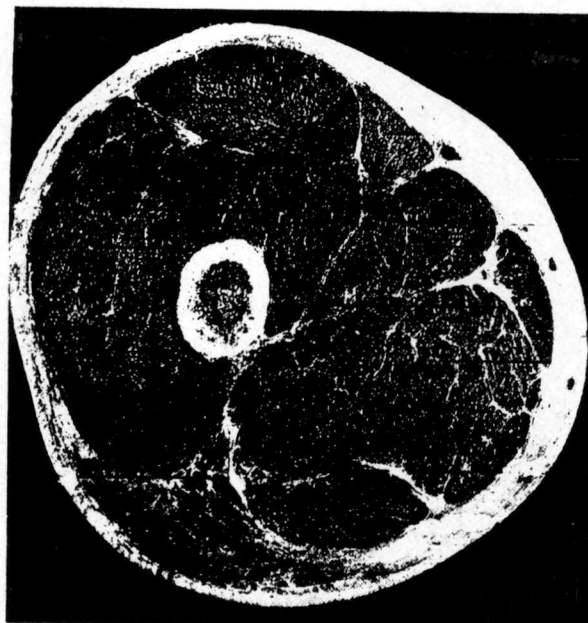


Figure 8. Cross-section of right thigh
(Rohen, 1988, p. 419).

C. MECHANICAL BEHAVIOR OF FEMUR

There is so much to consider when trying to evaluate the mechanical characteristics of the human femur that concrete, specific solutions are difficult. Among the variables to consider when studying bones are direction of loading, preservation of specimen (fresh or embalmed, wet or dry), age of individual, sex of individual, loading conditions (type of support, static, impact, etc.), and the geometry of the specimen.

The following is a summary of recent work performed in this area. Each study has its merits and its liabilities, and only taken as a whole can the data be meaningful.

1. Static Strength of Femur

Researchers have taken many approaches to determining the static properties of the femur. Whole bones, sections of bone, and machined specimens are commonly used. Information has been gathered on compressive strength, tensile strength, bending strength, and torsional strength (see Table 1).

Yamada (1973) tested sections of fresh, wet bones from 28 individuals in compression and found ultimate compressive strength to be 143 MPa. McElhaney (1966) used machined specimen from a femur of a 24-year-old male and found the compressive strength to be 150 MPa for the embalmed bone.

Table 1

Summary of Data on the Static Strength of Femur

Property	Yamada (1973)	McElhaney (1966)	Evans (1951)
Compr. Strength, fresh (MPa)	143		
Compr. Strength, embalmed (MPa)		150	
E (compr), embalmed (GPa)		15	
% Strain, embalmed		1.65	
UTS, fresh, wet (MPa)			82
E (tension), fresh, wet (GPa)			15.6
UTS, fresh, dry (MPa)			106
E (tension), fresh, dry (GPa)			18.4
Max. Bend. Load, fresh (N)	2453		
Bend. Strength, fresh (MPa)	186		
Average deflection (d/l)	0.032		
Torsion Breaking Moment, fresh (N·m)	137.3		
Torsion Strength (MPa)	45		

McElhaney also found Young's modulus to be 15 GPa and calculated the strain to be 1.65 percent.

Evans (1951) found the ultimate tensile strength of fresh, wet (kept in water) and dry (dried in air) bone using machined specimens. The UTS was 82 MPa for wet bone and 106 MPa for dry bone. Young's modulus was 15.6 GPa dry and 18.4 GPa dry.

Yamada (1973) loaded whole, wet bones in three-point bending in the anterior-posterior direction. The average breaking load was 2453 N. The average ultimate bending stress was 186 MPa, and the average deflection per length was .032. Yamada (1973) also found the torsional breaking moment of a whole, wet femur to be 137.3 N·m and the ultimate torsional stress to be 45 MPa.

The static strength of the femur information is summarized in Table 1.

2. Other Factors Affecting Femoral Strength

As mentioned before, several factors influence the strength of the femur. The effects of the method of preparation of specimens have already been seen in Table 1 (and will be discussed more later), as has the direction of loading. The next two most important considerations are the age of the individual who donated the bone and the rate of loading.

Yamada (1973) demonstrated how the strength of the human femur varies with age. This data can be seen in Table 2.

Table 2

Relationship Between Age and Femur Strength

Property	Age Group			Avg
	20-39	40-59	60-89	
Compressive Strength (MPa)	154	146	130	143
% change from 20-39	0%	5.2%	15.6%	7%
Ultimate Bending Load (Newtons)	2717	2413	2237	2452
% change from 20-39	0%	11.2%	17.7%	9.8%

Source : Yamada, 1973.

On average, the femur of a 50-year-old is 94.8 percent as strong as a femur of a 30-year-old, and that of a 70-year-old is only 84.4 percent as strong as the 30-year-old in compression. In bending the change gets worse with a 70-

year-old only being 82.3 percent as strong as the 30-year-old.

Mather (1968) developed a regression formula for the prediction of tensile breaking load. His formula is given as

$$\text{Load (N)} = 4280.9 - 16.5 \times \text{age (male)}$$

$$\text{Load (N)} = 3239.6 - 16.5 \times \text{age (female)}$$

Bone is highly strain rate-dependent. Figure 9 and Table 3 present data from McElhaney (1966) demonstrating this dependence. As can be seen, bone is stiffer and stronger at higher strain rates.

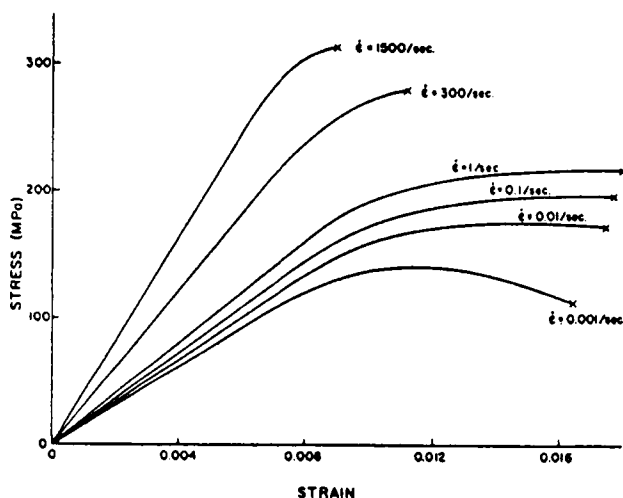


Figure 9. Illustration of strain rate dependence of the stress-strain curve of embalmed femur (McElhaney, 1966, p. 29).

Table 3

Strain Rate Effect on Properties of
Embalmed Human Femur

Strain Rate	Ultimate Compressive Stress (MPa)	Young's Modulus (GPa)	Strain to Failure (%)
0.001	150	15	1.65
0.01	179	17	1.75
0.1	200	18	1.80
1.0	221	22	1.78
300	279	30	1.10
1500	317	41	0.95

Source : McElhaney, 1966.

Mather (1968) made two important findings. He documented that the right and left femurs from the same individual will absorb statistically equal amounts of energy as long as both legs are equally operable. He also showed that the energy absorbing capacity of a bone is 50 percent greater under high strain rates than it is under low.

Ultimate tensile stress under impact loading was found to be 126.3 MPa by Saha (1976). This is significantly greater than the 82 MPa and 106 MPa found by Evans (1951)

under slow loading.

Finally, Sammarco (1971) illustrated a 46 percent increase in torsional strength of the femur when loaded 3000 times faster.

3. Effect of Embalming vs. Fresh, and Wet vs. Dry

Embalming effects have been discussed to a certain extent. One other study, however, merits additional mention. McElhaney (1964) performed a test on beef bones to quantitate the effect. Results are presented in Table 4.

Table 4

Comparison of Fresh and Embalmed Beef Bone
(low strain rate)

Property	Fresh	Embalmed	% Change
UTS (MPa)	91.7	88.6	-3.4
Maximum Tensile Strain	.00503	.00480	-4.6
E, tension (GPa)	20.5	20.3	-1.3
UCS (MPa)	133.5	117.1	-12.3
Max Compressive Strain	.00482	.00461	-4.4
E, compression (GPa)	28.8	27.0	-6.2

Source : McElhaney, 1964.

Figure 10 depicts the difference in dry and wet bones. The wet bone, whose characteristics are most like bones in living individuals, absorb more energy and elongate more before fracture. Dry bones on the other hand have a higher modulus of elasticity and compressive strength (Evans, 1952).

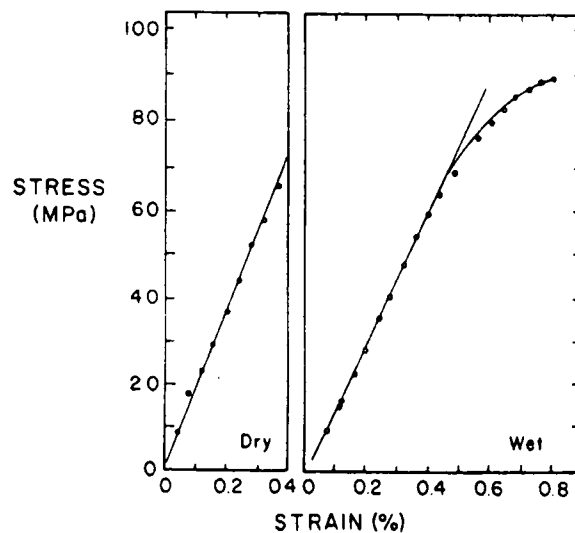


Figure 10. Comparison of dry and wet stress-strain curves (Evans, 1952, p. 136).

4. Axial Knee Impact Tests

Impact compression loading onto the whole femur has been limited to intact leg/knee impact studies. The list of researchers included Viano and Stalnaker (1980), Cooke and Nagel (1969), Horsch and Patrick (1976), Kramer et al. (1973), Kulowski (1964), Lister and Wall (1970), Melvin and Stalnaker (1975), Patrick et al. (1965), Powell et al. (1975), and Wilson (1969).

Nearly all of these papers concentrate solely on force tolerance levels. All use low mass strikers. Patrick (1965) initially found force levels of 6200 Newtons. More recently, Powell et al. (1975) found an average force level of 10,500 N. Age and osteoporosis may reduce the axial strength of the impacted femur by as much as a factor of three. Viano and Khalil (1976) showed that femur responds as a column-like structure under compressive loads.

In terms of fracture study, relatively little has been documented. Powell et al. (1975) reported four femoral shaft fractures, two femoral neck fractures, and one trochanteric fracture.

Viano and Stalnaker (1980) describe the knee impact event as being composed of three main features: knee joint compression, femoral bending and compression, and femoral displacement with load transfer to the hip. Each of these result in a particular mode of injury. Knee joint

compression may cause condyle fractures. Femoral bending may lead to shaft fracture. Transfer of load to hip may lead to femoral neck fracture. Viano and Stalnaker go on to report that neck fractures occur frequently in rigid impact experiments due to tensile strains on the posterior side of the femoral neck. Few pelvic fractures have been reported.

D. FEMUR FRACTURES IN THE REAL WORLD

Femoral fractures are most commonly the result of motor vehicle accidents, motorcycle accidents, falls, and age-induced osteoporotic failures. A distinction is typically made in the literature between femoral fractures and hip fractures, regarding the former solely as femoral shaft fractures. Since both of these types of fractures occur on the femur bone, they will both be addressed in this report. A small percentage of the reported hip fractures actually occur on the pelvis.

Clearly the greatest cause of femoral compression fractures occurs during automobile head-on collisions where the knee impacts against the automobile dashboard. Most simulation research (discussed in the previous section) has centered around this type of loading. In terms of the fractures that occur, Daffner et al. (1988) studied 500 patients involved in high-speed, head-on collisions. Two-hundred-sixty-five of these suffered femur fractures, and

nearly half of these were in the hip region (from the head of the femur to the greater trochanter). Chapman (1984) also reports fractures of the hip concomitant with shaft fractures are common in dashboard impact injuries.

Numerous studies have reported on injuries sustained during motorcycle accidents (Deaner et al., 1975, Drysdale et al., 1975, Zettas et al., 1979, Aldman, 1984). But little has been done to simulate this accident, owing to its high degree of irregularity. Deaner et al. (1975) reveal a 17 times greater chance of fatality in motorcycle accidents than in automobile accidents. An individual who owns a motorcycle for 10 years risks a 35 percent chance of experiencing a serious collision (Drysdale et al., 1975).

The femur is fractured in 12.7 percent of motorcycle accident patients according to Deaner et al. (1975) and in 13.5 percent according to Drysdale et al. (1975), but Zettas et al. (1979) report this percentage to be as high as 31 percent. Deaner et al. calculate 17.2 percent of these femoral fractures to occur at the neck. The number of neck fractures increases significantly with age. Of 57 injured cyclists over 65 years of age, 11 suffered femoral neck fractures (McCoy et al., 1989). No other femoral fracture location was reported.

This increase of neck failures with age is expected, especially in women. Postmenopausal estrogen deficiency causes bone loss of one percent to two percent per year. Fifty percent of women have bone mineral content below fracture threshold by age 65, and 100 percent are below the threshold by age 85. A different cause but the same effect is also seen in men (Hofeldt, 1987).

The fracture of bones resulting from a fall from height has been studied by Scalea et al. (1986). The results of his study of 161 adults who fell one to seven stories can be seen in Table 5. At every height, the hip fractures total more than the shaft fractures.

Table 5

Femur Damage in Patients Who Fell 1-7 Stories

Type of Fracture	Number of Stories							total
	1	2	3	4	5	6	7	
Hip (%)	10	11	16	35	55	67	60	22
Shaft (%)	3	7	11	24	33	17	40	13

Source : Scalea et al., 1986.

In conclusion, several investigative studies have been performed on the femur, but several mysteries still exist. Very little work has concentrated on stresses in the axially impacted bone or on the mechanisms which determine fracture locations.

CHAPTER III

DESCRIPTION OF TESTING DEVICES

This research utilizes a biomechanics research laboratory, the center of which is a device that simulates an automobile impact. This machine is referred to in this thesis as the crash simulator. This device was the most frequently used tool in producing the data used in this thesis.

The crash simulator is a pneumatically-powered machine which simulates a car/motorcycle or car/pedestrian collision. A cart of significant mass is propelled down a rail system where it impacts a test specimen (e.g. a human bone, a human leg, an animal bone or an artificial bone). The cart is instrumented with a force measurement system enabling the user to obtain much information about the specimen.

The three principle parts of the crash simulator are (a) the accelerator and cart, (b) the specimen holding device, and (c) the force measurement system. These are discussed below.

A. ACCELERATOR AND CART

The cart accelerator system (Figure 11) is composed primarily of an air tank which applies pressure on a piston within the tank. The piston acts like a ram and pushes an impact cart (Figure 12). The cart travels down a rail system and impacts the test specimen. The cart is decelerated by impact with bales of energy-absorbing wood fiber.

The mass of the cart is sufficient for an insignificant velocity change through impact. This fact separates the crash simulator from most impact testing devices. The velocity of the cart is directly related to the amount of pressure in the pneumatic tank. 347×10^3 Pascal (50 psi) produces a cart velocity of 7.44 m/s (26 ft/sec) (Kress, 1989). The details of this system are presented by Kress (1989). All impact tests in this report were performed at this speed.

B. SPECIMEN HOLDING DEVICES

The specimen holding device is one of the most important parts of the crash simulator. It must be strong and rigid enough to withstand tens of thousands of Newtons of force without significant movement, and it must provide the desirable loading conditions for the test to be performed.

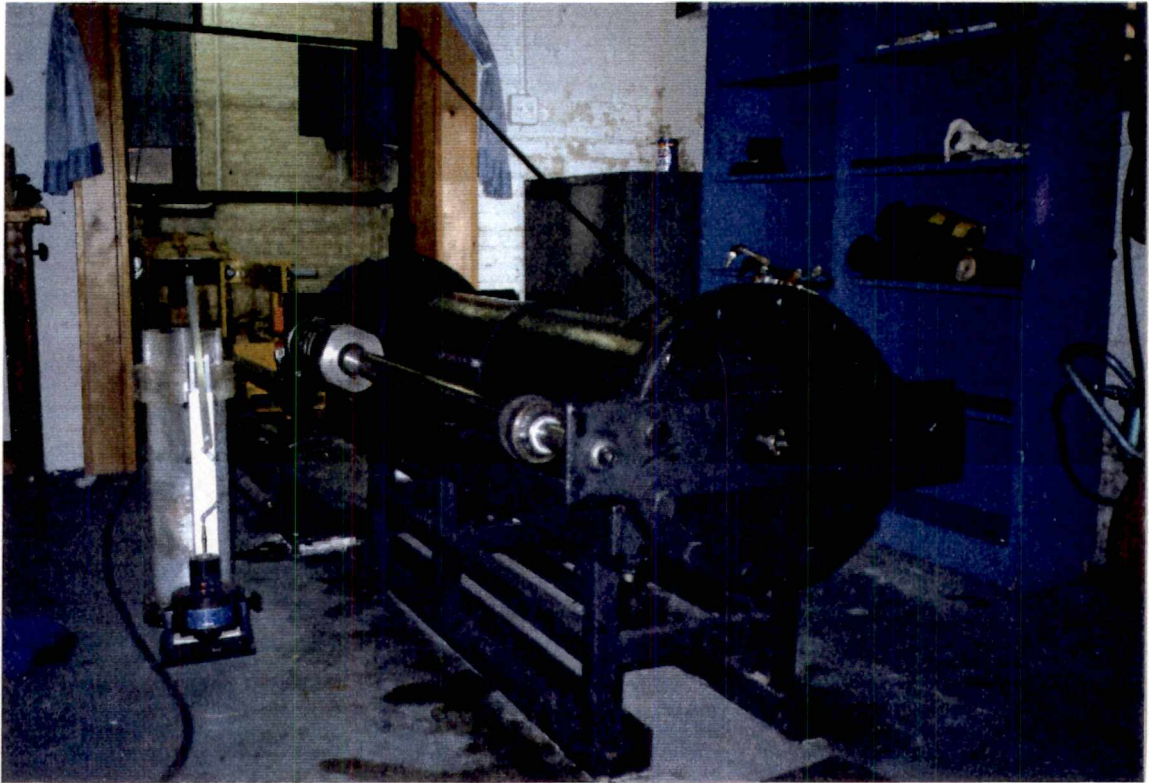


Figure 11. Cart accelerator system.

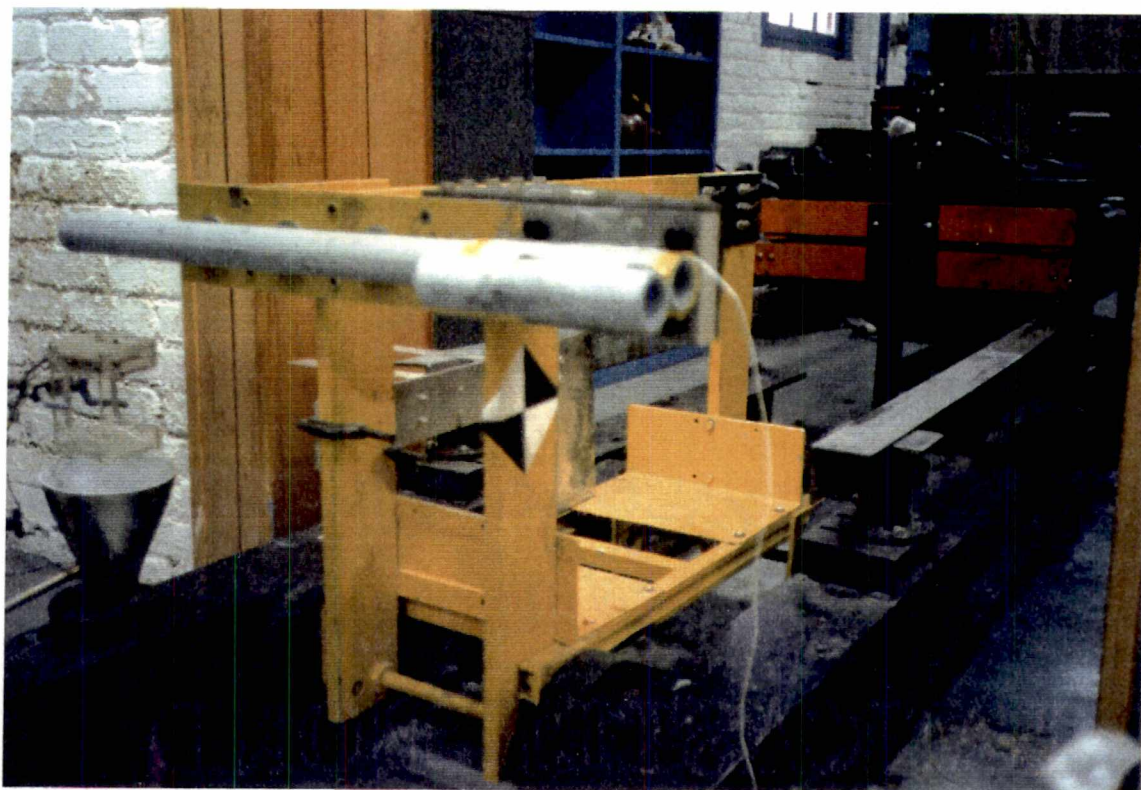


Figure 12. Impact cart.

Since there were four different tests run using the crash simulator, there are four holding devices which need to be discussed. All of the tests involved human femurs and are listed as follows :

1. Simply-supported (pinned-pinned) bone loaded in the anterior-posterior direction,
2. Simply-supported bone loaded in the lateral-medial direction,
3. Simply-supported bone with a torsional preload, and
4. Axially loaded bone.

1. Simple Support, A-P Loading

The a-p simple-support (Figure 13) is the least complex support of the four. The top support is a steel plate which hangs from a horizontal crossbar spanning the rail system. The bottom support is a steel plate clamped onto the left rail.

Due to the geometry of the femur, securing the bone to the holding device for this test is also quite simple. The posterior side of the distal condyle of the femur rests against the bottom support and is held in place by duct tape (much more duct tape is used than is shown in the photograph). The posterior side of the greater trochanter rests against the upper support. The posterior of the head

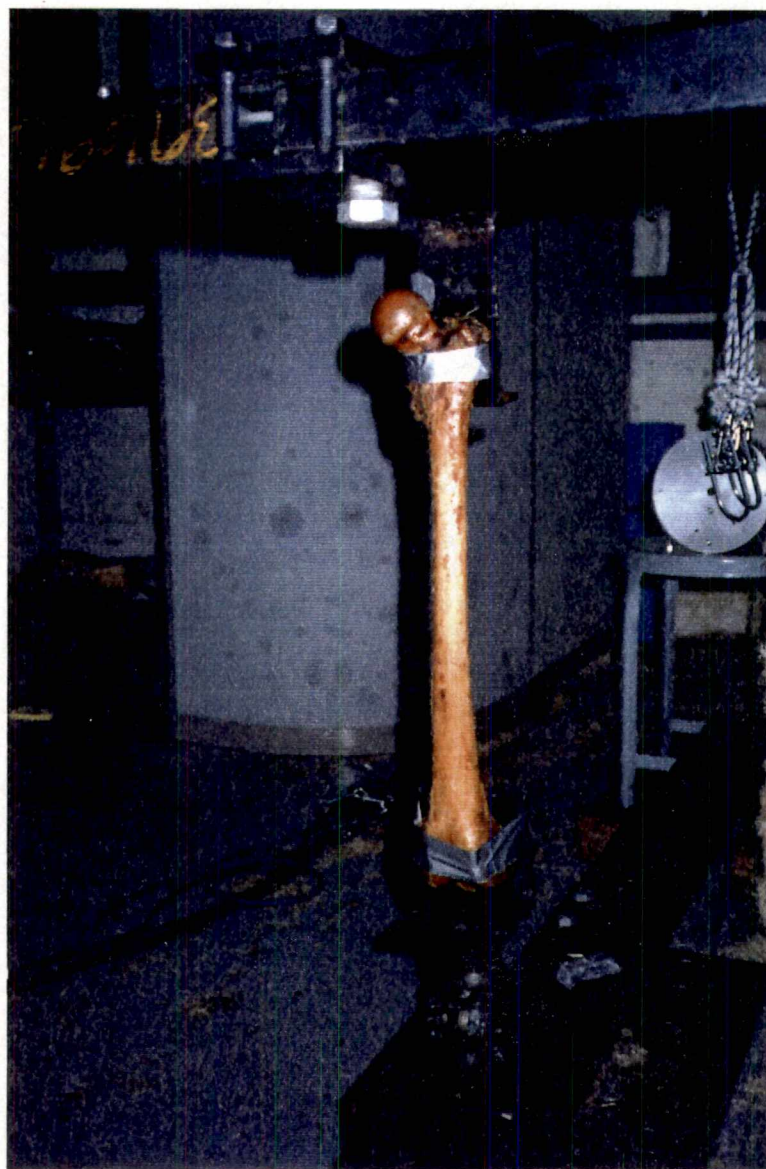


Figure 13. Anterior-posterior simple support.

of the femur usually will be displaced from the support by one to two centimeters. A lead wedge is inserted as a shim. This proximal portion of the bone is then taped to the support.

2. Simple Support, L-M Loading

Loading the bone in the lateral-medial direction means the femur has to touch the supports at its depression for ligamentum teres on the medial side of the head and at the inner tuberosity of the distal condyle. Obviously, without additional support to the holding device described in section one, the femur would rotate during impact and then be loaded in the anterior-posterior direction.

To prevent this rotation, a U-shaped piece is added to the upper support. This piece fits around the head and extends beyond the greater trochanter. Two holes in the piece allow for a rod to pass through the U and through the greater trochanter of the bone (Figure 14).

For the bottom support, an aluminum cup is clamped onto the distal condyle of the femur. The cup is connected to a square aluminum piece that rests against the bottom support preventing rotation. The cup's motion vertically is not restricted maintaining the simple support of the bone.

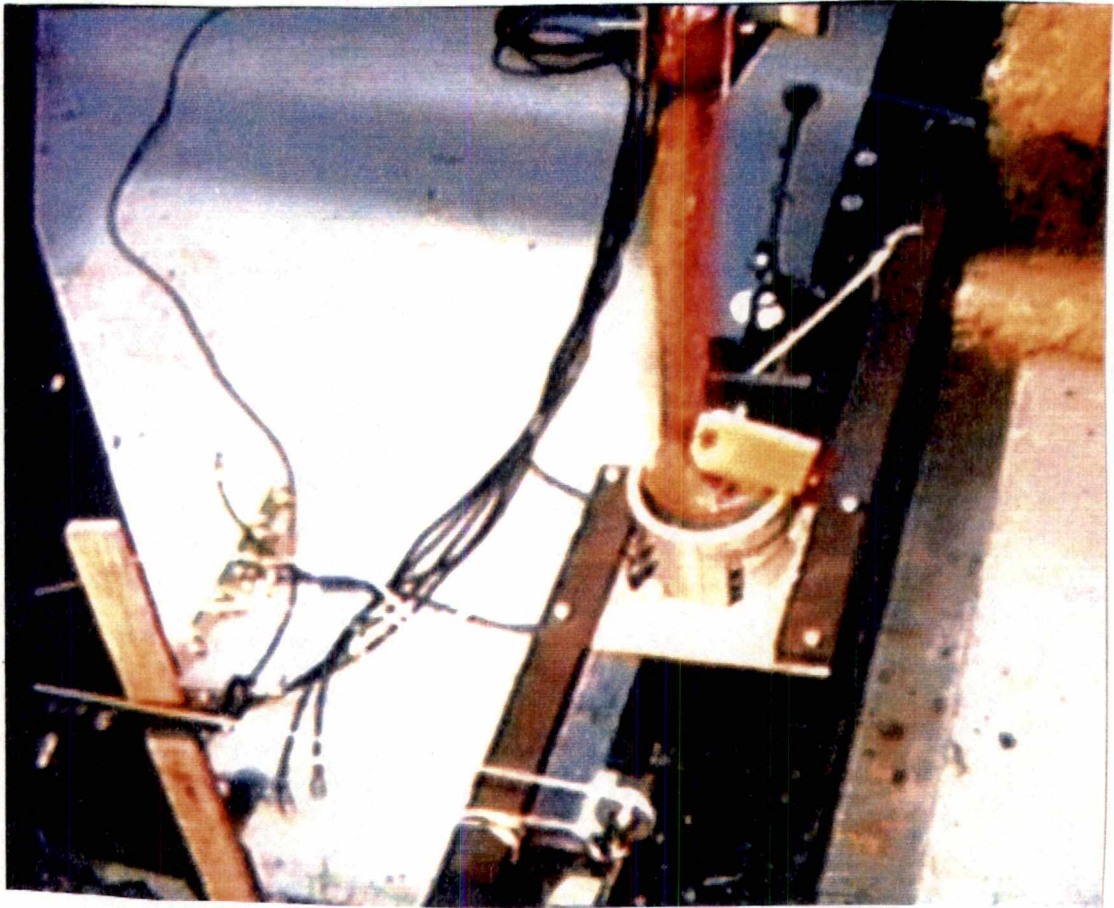


Figure 14. Lateral-medial simple support and torsional preload.

3. Simple Support, Torsional Preload

The torsional preload holding device (Figure 14) is very similar to the 1-m simple support. The cup clamped to the distal condyle is equipped with two symmetrically-opposite arms. Weights are hung to pull on these arms via a pulley system with each arm applying equal torque on the cup and subsequently on the femur.

4. Axial Loading

The axial loading specimen holding device is by far the most complicated, cumbersome and tedious of the four devices. The force must be delivered axially through the bone, and the bone must be held in a manner that does not affect the loading conditions.

A rope is hung so that it travels directly above the left rail and is easily cleared by the cart (Figure 15). From this rope hangs two other pieces of rope. These pieces hold the femur, one at the distal end and one at the proximal end. The bone now is hanging horizontally and is situated in the correct direction with respect to the rails.

An artificial acetabulum cup (Figure 16) is constructed and is attached to a large mass. The cup and mass are placed directly behind the bone. Duct tape prevents the head of the femur from becoming dislodged from the cup.

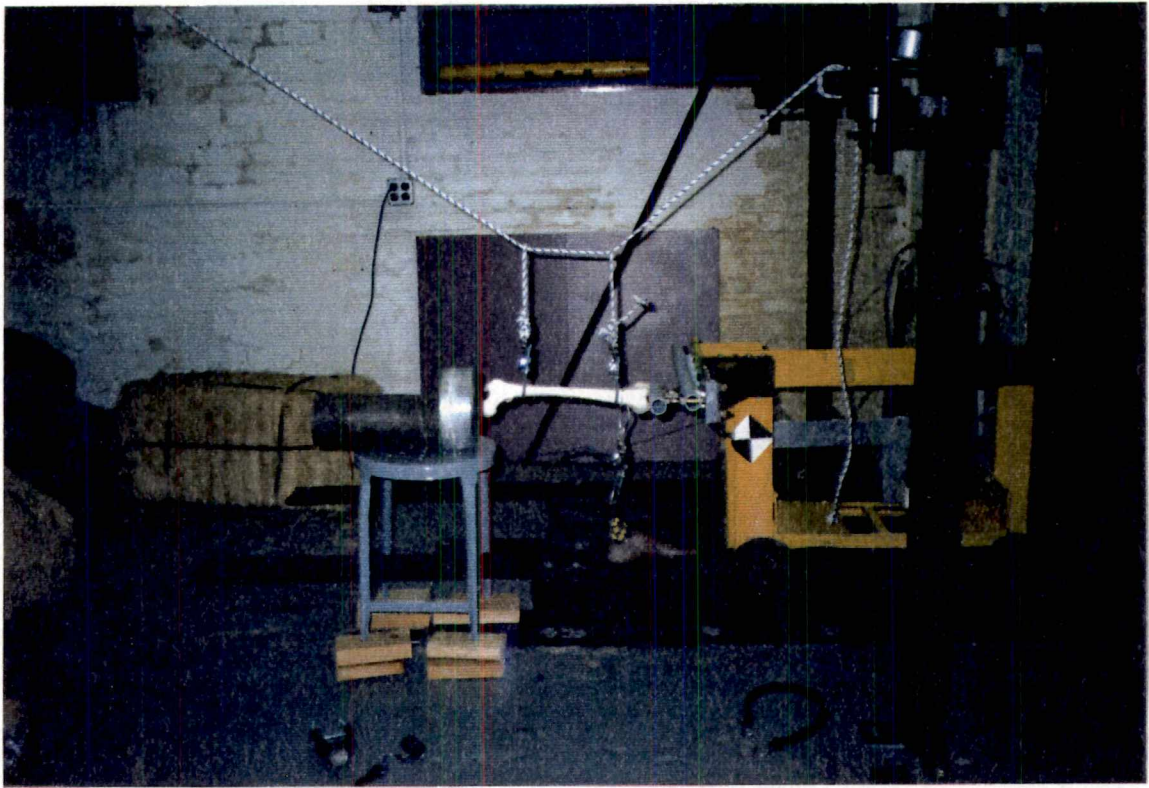


Figure 15. Side view of axial-loading specimen holding device.

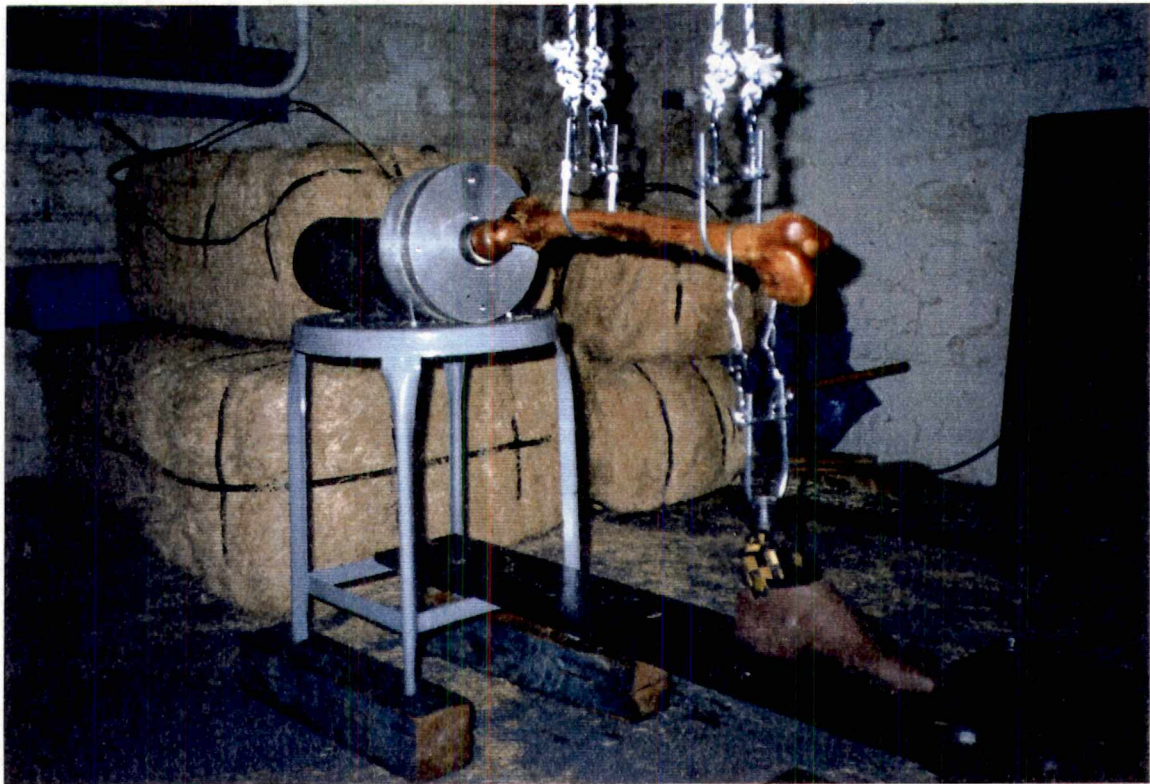


Figure 16. 45-degree view of axial-loading specimen holding device.

The mass of the lower leg is simulated by hanging an impact dummy artificial foot from the distal end of the femur.

C. FORCE MEASUREMENT SYSTEM

Data from each test using the crash simulator is obtained via a force transducer mounted on the cart. The force transducer is manufactured by PCB Piezoelectronics, model number 208A04. The sensitivity of the transducer is 1.16 kilonewtons per volt.

A 4.1275-centimeter (1 5/8-inch) pipe or a 7.62 x 20.32 x 0.3175 centimeter (3 x 8 x 1/8 inch) plate mounted on the cart serves as the impact surface. The pipe or plate is held on by slide pins (Figures 17, 19). The slide pins allow all of the force to be transferred to the force transducer (Figures 18, 20).

The signal from the force transducer passes through a PCB Power Unit and then to a Hewlett-Packard 3562a Dynamic Signal Analyzer where the force versus time history of the event is recorded, displayed and plotted (Figure 21, 22).

As is the case with all dynamic data acquisition, the time records from the event contain pertinent information and noise. Much of the noise is from vibrating structures. It is therefore important to know the natural frequencies of the surrounding structures in order to properly understand

the record taken.

Modal analysis was performed on the top support and the bottom support of the specimen holding device. The first few natural frequencies of the top support were found to be 46 Hz, 375 Hz, 465 Hz, 565 Hz, 830 Hz, and 936 Hz. For the bottom support, the only natural frequency below 100 Hz was at 241 Hz.

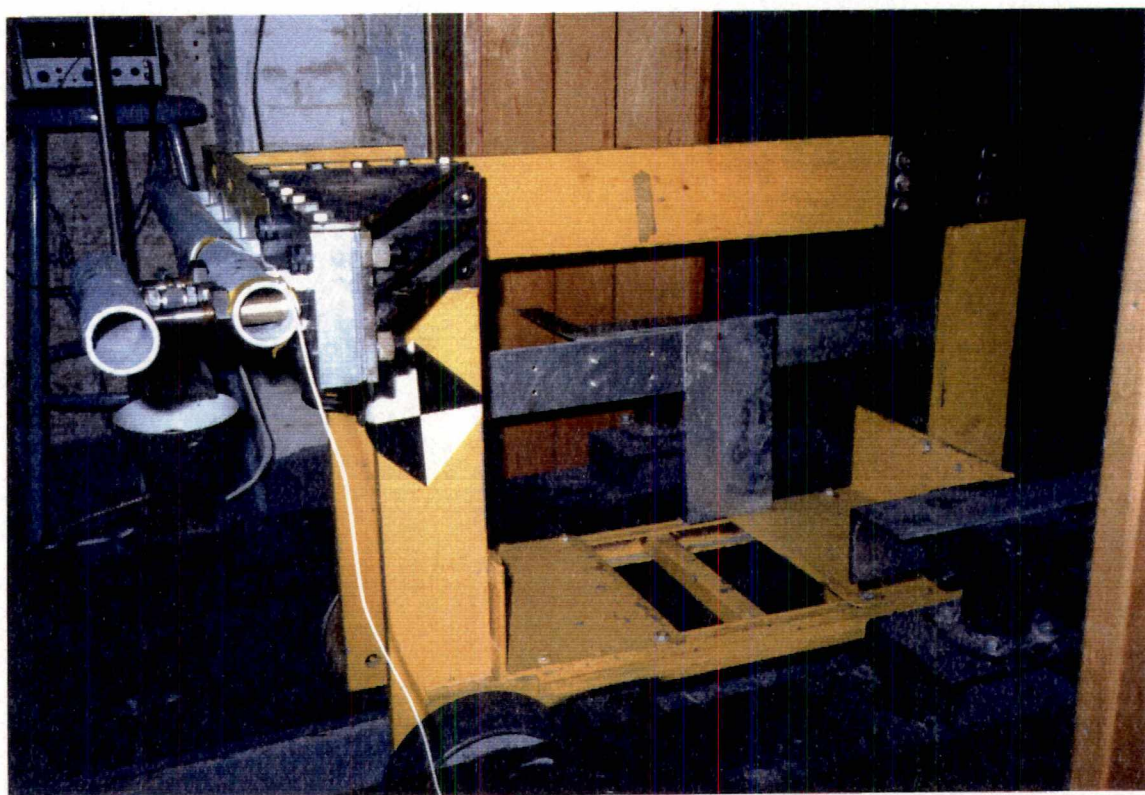


Figure 17. Impactor support cart showing pipe impactor.

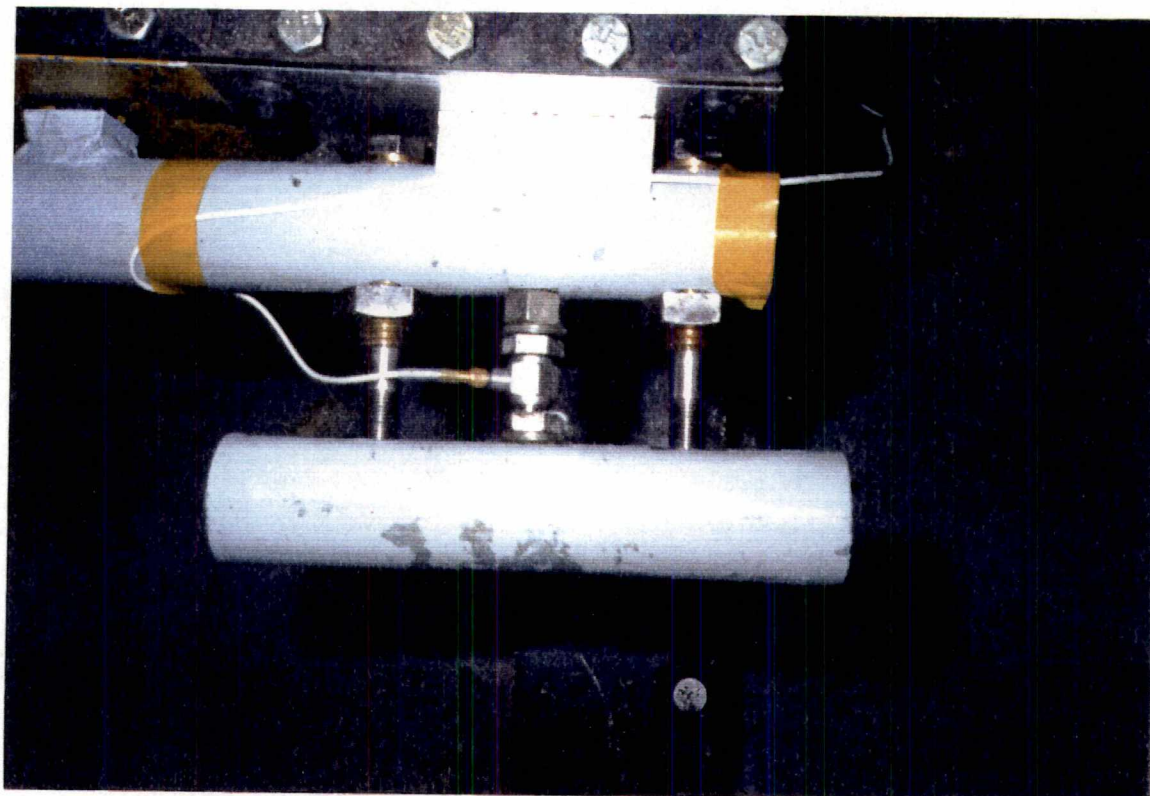


Figure 18. Top view of pipe impactor. Notice slide pins (both sides) and force transducer (center).

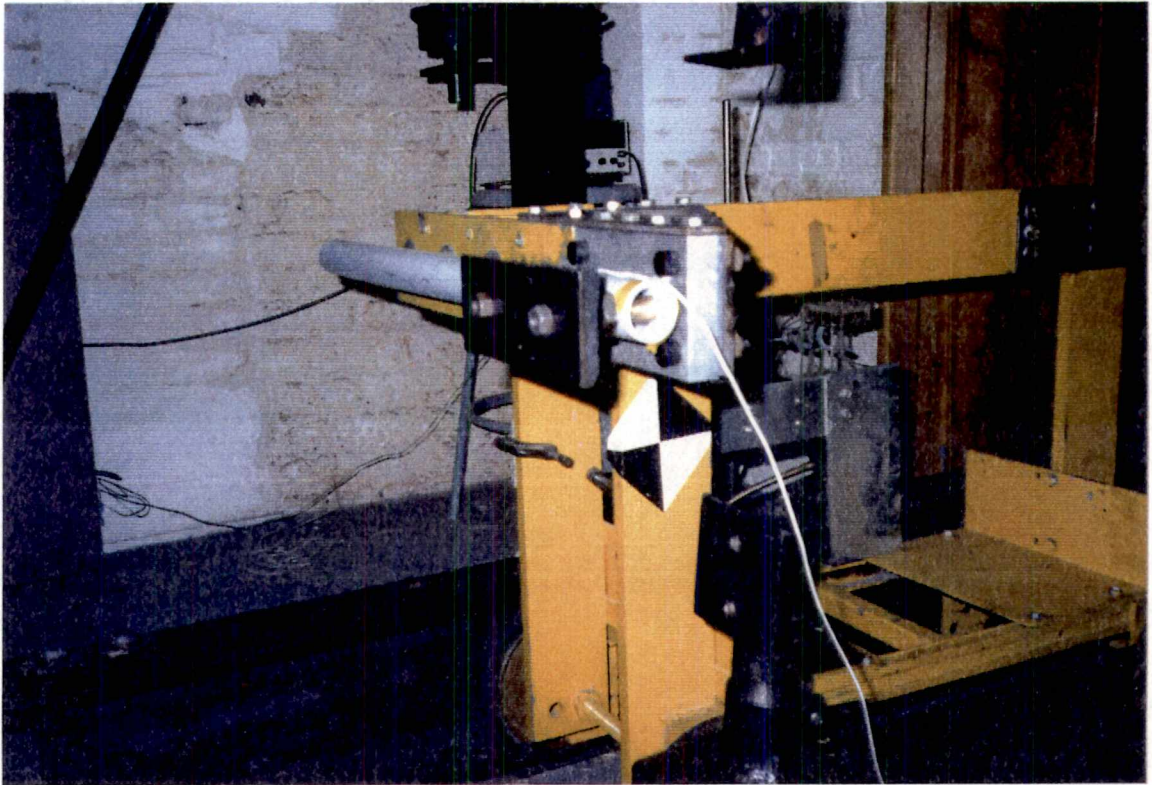


Figure 19. Impactor support cart showing plate impactor.

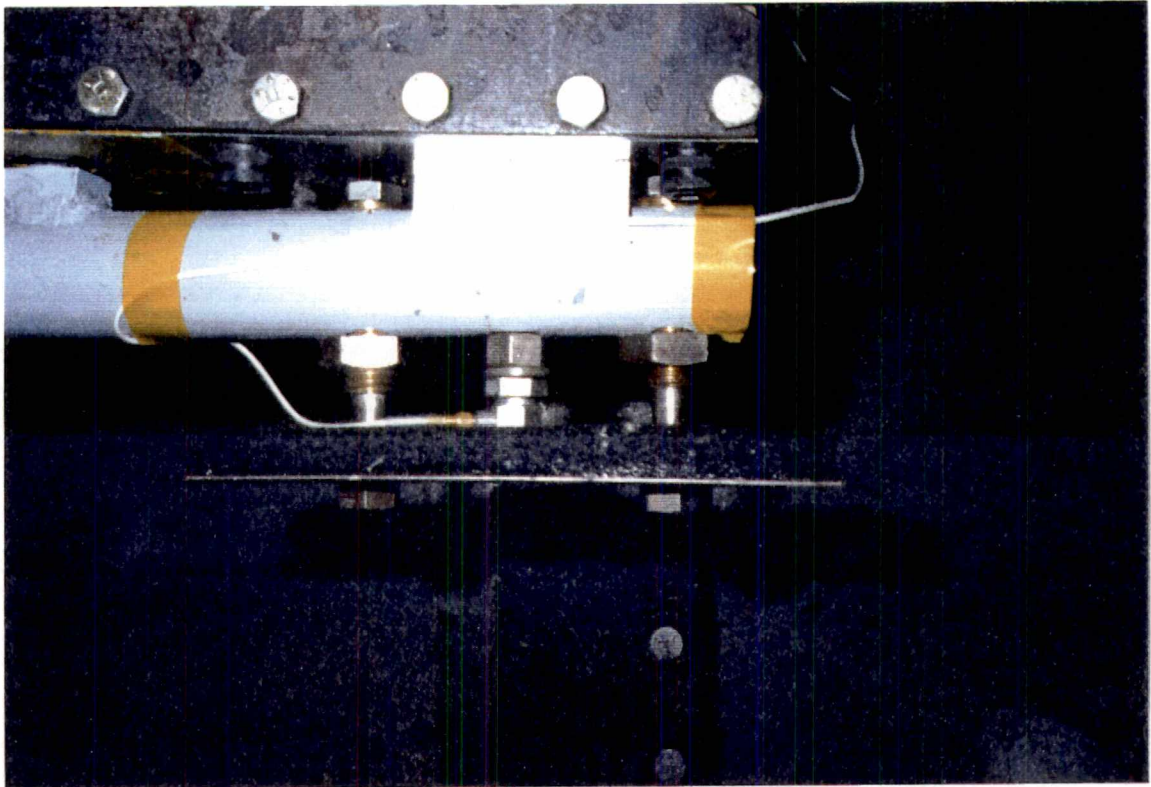


Figure 20. Top view of plate impactor. Notice slide pins (both sides) and force transducer (center).



Figure 21. Dynamic signal analyzer.

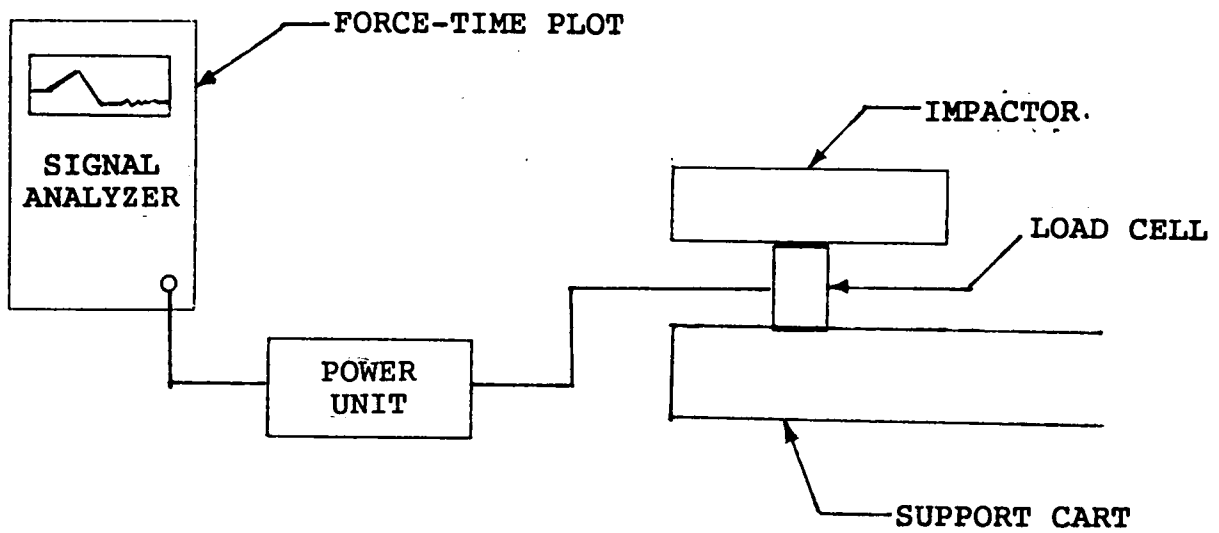


Figure 22. Block diagram of instrumentation.

CHAPTER IV

METHODOLOGY

A. MATERIALS

Ninety-four bones were obtained from the University of Louisville and the University of Tennessee Hospital in Knoxville. The University of Louisville supplied 80 embalmed femurs with soft tissue removed (see figure 23). The University of Tennessee supplied 14 fresh, cryogenically frozen long bones from two recently deceased persons. Demographic information was available for some of the bones.

Since the objective of this thesis is to completely understand the mechanical behavior of the femur when impacted, a number of different loading conditions were applied on these bones. These different conditions can be divided into four types of tests: 1) bending, 2) torsional, 3) axial, and 4) fresh tissue tests.

In all tests except the low strain rate axial tests, the crash simulator discussed in Chapter III was used. Data was recorded and presented by the crash simulator in the form of a force-time plot. An example of this plot can be seen in Figure 24.

The breaking force, the amount of time from impact initiation to fracture, and the area under the curve were



Figure 23. Sample of embalmed femurs used in tests.

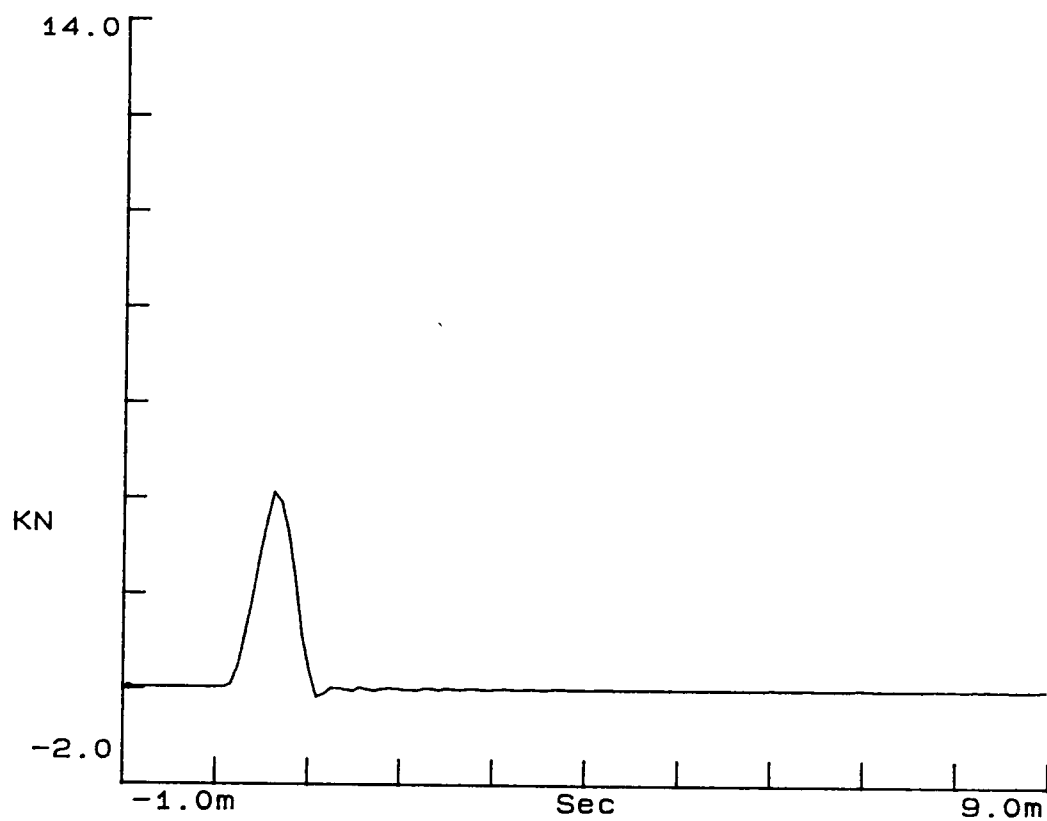


Figure 24. Sample force-time plot produced by crash simulator after impact.

obtained from this force-time curve. The breaking force was used to calculate, among other things, the ultimate stress. The time measurement allows for the calculation of strain since there is a constant velocity through impact. The area under the curve is roughly related to the amount of energy absorbed during impact and is used strictly for comparison with other tests.

Prior to testing, anatomical measurements were made on the bones (see Figure 25). Following testing, cortical thickness measurements were taken.

Protocol, justification and procedure for each test is detailed below.

B. TEST I: BREAKING STRENGTH OF FEMUR

1. Lateral-medial loading

Lateral-medial impact of the femur occurs primarily in the real world when an automobile impacts the side of a motorcycle.

Lateral-medial loading of the bare femur at impact velocities was accomplished using the crash simulator at a speed of approximately seven meters per second. The "simple support, L-M loading" holding device discussed in Chapter III was employed. Twelve femurs were tested.

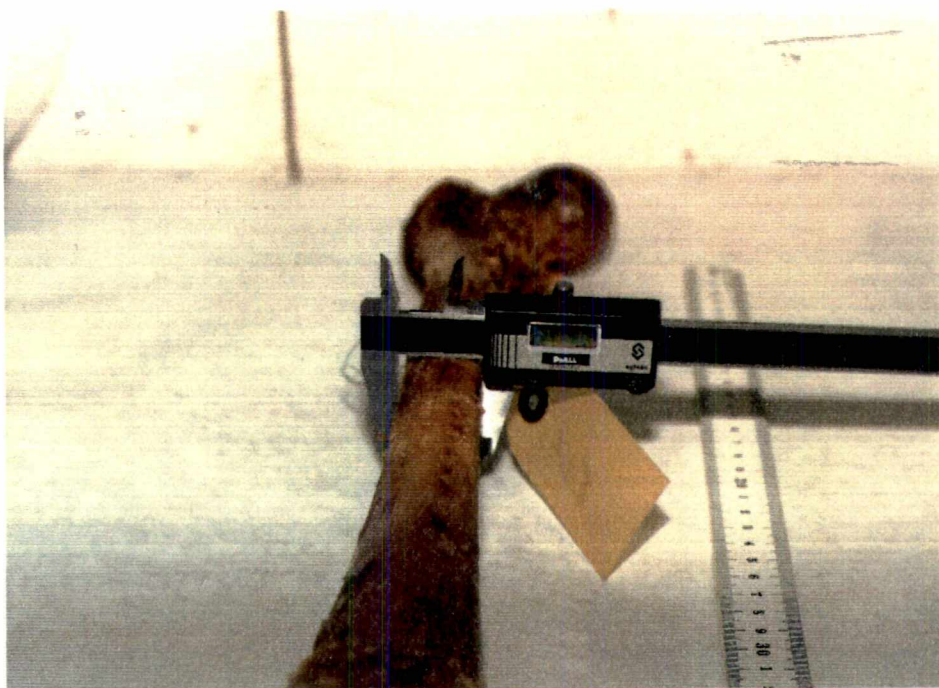


Figure 25. Anatomical measuring devices.

The breaking force was determined for each bone from the force-time plot. The breaking forces for each bone is averaged to determine a bone tolerance level. Kress (1989) reported a good correlation is breaking strength to cortical thickness. This correlation will be improved on by using a higher order curve-fitting technique.

Using the breaking force and the anatomical measurements taken for each bone, the ultimate bending stress can be approximated using beam theory.

The formula for calculating bending stress is

$$\sigma_{b(max)} = \frac{M_{(max)} C}{I}$$

where

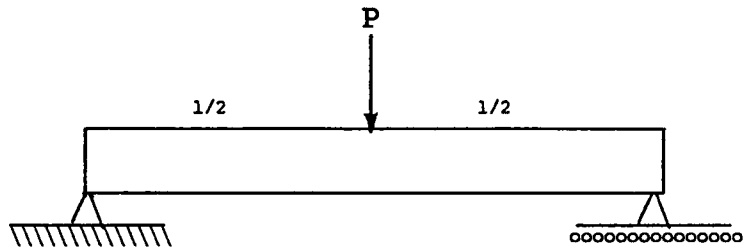
σ_b = bending stress

M = bending moment

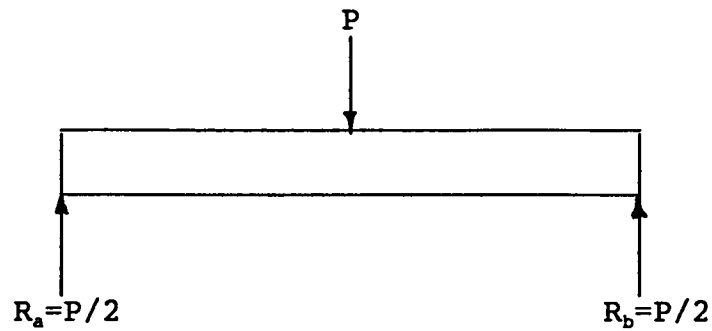
c = distance from centroid to edge of beam

I = moment of inertia

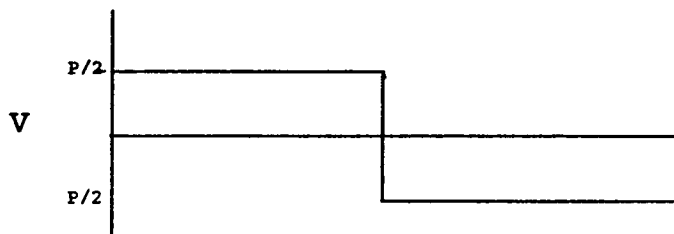
For a simply supported beam loaded in the center, the maximum moment is (see Figure 26).



Free Body Diagram:



Shear Diagram:



Moment Diagram:

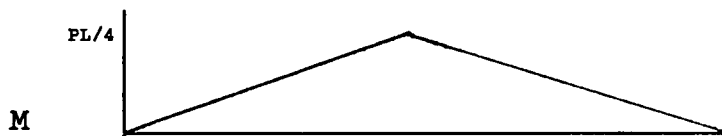


Figure 26. Determination of maximum bending moment for simply-supported beam.

$$M_{(max)} = \frac{PL}{4}$$

where

P = breaking force

L = distance between support

For the femur calculations, the shaft will be considered a perfect cylinder with an outer radius, r_o , and an inner radius, r_i , with

$$r_i = r_o - t$$

where t is the cortical thickness. The cortical thickness is measured at six points at midshaft and averaged. This is an acceptable method (Viano and Khalil, 1976; and Moore, 1985). The distance from centroid to edge of bone, c , is simply r_o , and I is given by

$$I = \frac{\pi}{4} (r_o^4 - (r_o - t)^4)$$

In addition to bending stress, Young's modulus can be approximated using beam theory. The equation for maximum defection of this beam,

$$\delta_{\max} = \frac{PL^3}{48EI}$$

can be written as

$$E = \frac{PL^3}{48\delta I}$$

where E = Young's modulus

δ = maximum defection

The maximum deflection is found by multiplying time of contact by striker velocity, since there is no change in the striker velocity through the event. This calculation is very crude and is only an approximation of Young's modulus. The equation used is only valid for uniform cross-sectional bodies.

The area under the force-time curve is measured to approximate the relative energy absorption. The types of fractures that occur are analyzed.

2. Anterior-posterior loading

Anterior-posterior loading of the femur occurs often in the real world. It is most associated with flying bodies.

Testing the femur in this direction is also of academic interests, since bone is non-homogeneous, anisotropic and has complex, varying geometry.

Much of the procedure in this test is the same as the lateral-medially loaded femurs. The "simple-support, a-p loading" holding device was used, and 32 femurs were tested.

Demographic information was available for these bones, making it possible to study age effects on the properties of the bone. In addition, this testing allowed for a comparison to be made between the properties of left and right femurs of the same person.

C. TEST II: TORSIONAL STRENGTH OF FEMUR

1. Steady-state torsional loads

Six femurs were available for determination of the maximum, slowly-applied torque on the bone. The torsional loading device discussed in Chapter III was used. Gradually increasing forces were applied to the arms of the device until failure occurred.

Knowing the force applied to the arm and the length of the arm, the maximum torque is easily calculated. The maximum torsional stress is calculated as (remember the femur shaft is considered a hollow circular cylinder)

$$\tau_{(max)} = \frac{T_{(max)} c}{J}$$

where τ = shear stress
 T = torque
 c = distance from centroid to outer
 edge of bone (r_o)
 J = polar moment of inertia and

$$J = \frac{\pi}{2} (r_o^4 - (r_o - t)^4)$$

2. Torsional preload with l-m impact

It is suspected that motorcycle riders' legs undergo multiple loading configurations when suffering a collision.

To begin to understand the effects of these multiple loads, this combined torsion and bending test was performed. A torsional preload was placed on a simply-supported femur (as it was described in the previous section). The femur was then impacted at high-speed in the lateral-medial direction with the crash simulator.

To best understand the effect of the torsional preload, matched pairs of femurs were used. The right femur was

struck in the l-m direction with no torsional preload. The left femur was struck similarly but with a torsional preload.

D. TEST III. COMPRESSIVE STRENGTH OF FEMUR

Although several researchers have investigated the compressive strength of the entire thigh, few, if any, have loaded a whole, bare femur in the axial direction at low and high strain rates. Eighteen bones were tested under such conditions.

1. Steady-state axial load

A materials testing machine (see Figure 27) was employed to test nine bare, embalmed femurs in axial compression to failure. Cups simulating the acetabulum and the tibial plateau were designed to fit the machine and allow a distributed load on the condyles and head of the femur. The only data taken from the machine was the breaking force. Video documentation allowed for the analysis of fractures.

Using the breaking force and the anatomical measurements, breaking strength can again be approximated using beam theory. Due to the geometry of the femur, an axial load is not truly an axial load. A bending element is also involved. The stress when bending and axial loads are

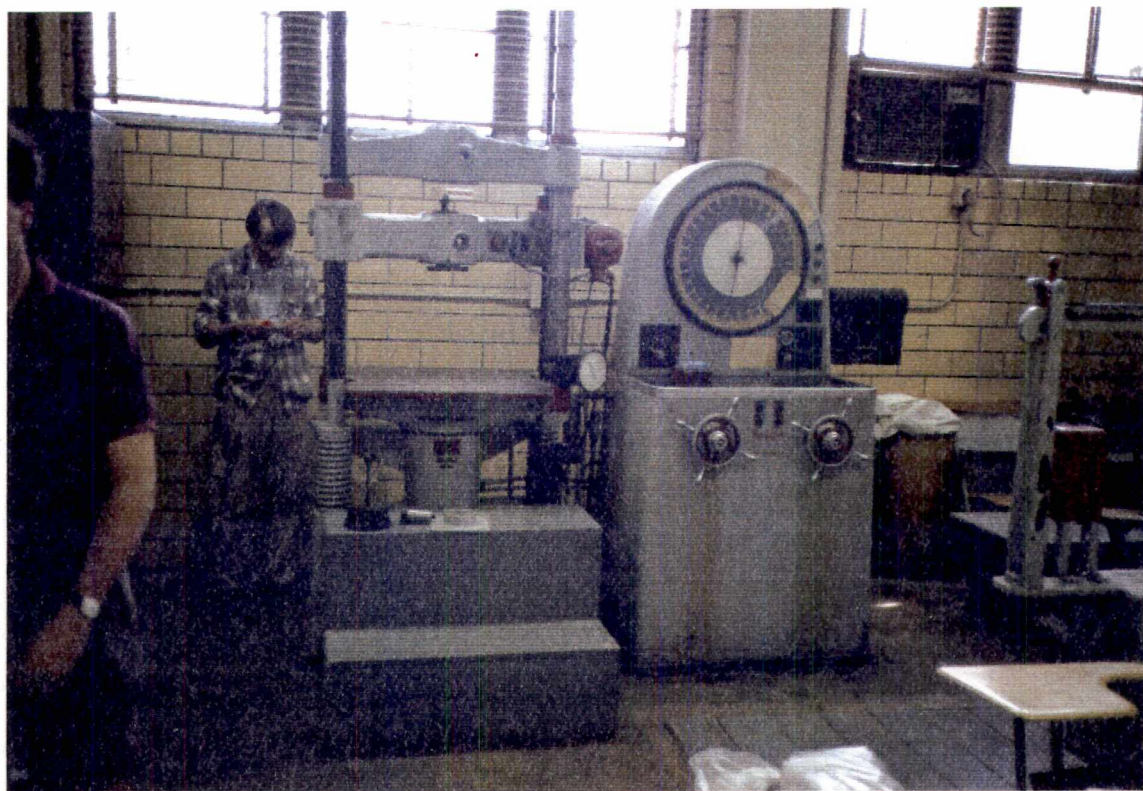


Figure 27. Materials testing machine

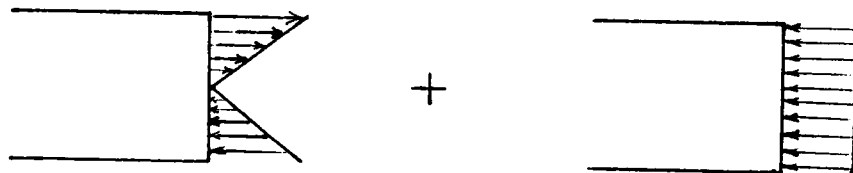
involved is given by

$$\sigma_x = (\sigma_x)_{centric} + (\sigma_x)_{bending}$$

$$= \frac{P}{A} \pm \frac{Mc}{I}$$

where P = force
 A = cross-sectional area
 M = bending moment
 c = distance from centroid
 I = moment of inertia

A side view of the cross-section shows the action of the stresses.



The top half of the cross-section is in tension and the bottom half is in compression due to the bending element. The entire cross-section is in compression from the centric element. The stress in the bone, therefore, is given by

$$\sigma_x = \frac{P}{A} + \frac{MC}{I} \quad (\text{bottom half})$$

$$\sigma_x = \frac{P}{A} - \frac{MC}{I} \quad (\text{top half})$$

It should be noted that these calculations are approximations. Mc/I does not hold when E , tension does not equal E , compression. The E 's are close enough in the bone (four percent according to Evans, 1951), however, that this does not significantly change the results. Also, the neutral axis is not the same as the centroidal axis.

Stresses were calculated at three cross-sections on the bone: at midshaft, just below the greater trochanter, and at the neck of the femur (see Figure 28). The cross-section of minimum inertia was chosen at each area.

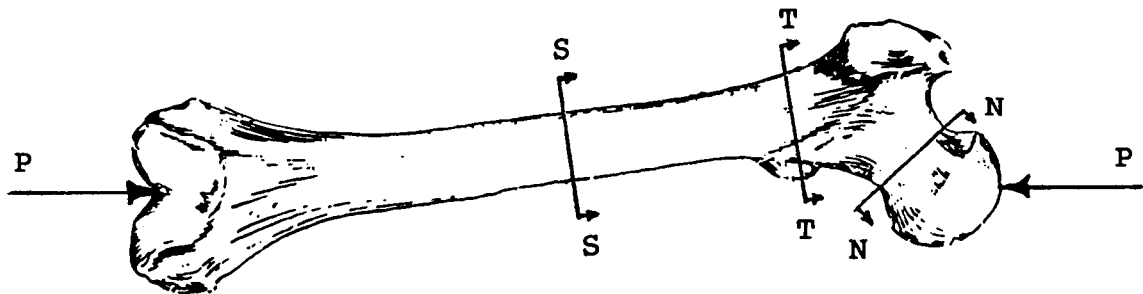


Figure 28. Diagram of axially loaded femur showing cross-sections of stress calculation: S = shaft, T = Trochanter, N = neck.

The breaking force, P, and cross-sectional areas are taken directly from measurements. The bending moment, M, is equal to

$$M = Pd$$

where

d = moment arm.

d is measured as shown in Figure 29. A hollow cylinder cross-section is assumed as before.

2. High-speed axial load (knee impact)

In order to examine the difference in low and high strain rate loading of the femur in the axial direction, nine femurs have been impacted in the crash simulator. The axial loading, specimen holding device was used. The flat plate impactor was used instead of the pipe.

Measurements and calculations for these tests were the same as for the statically loaded bone.

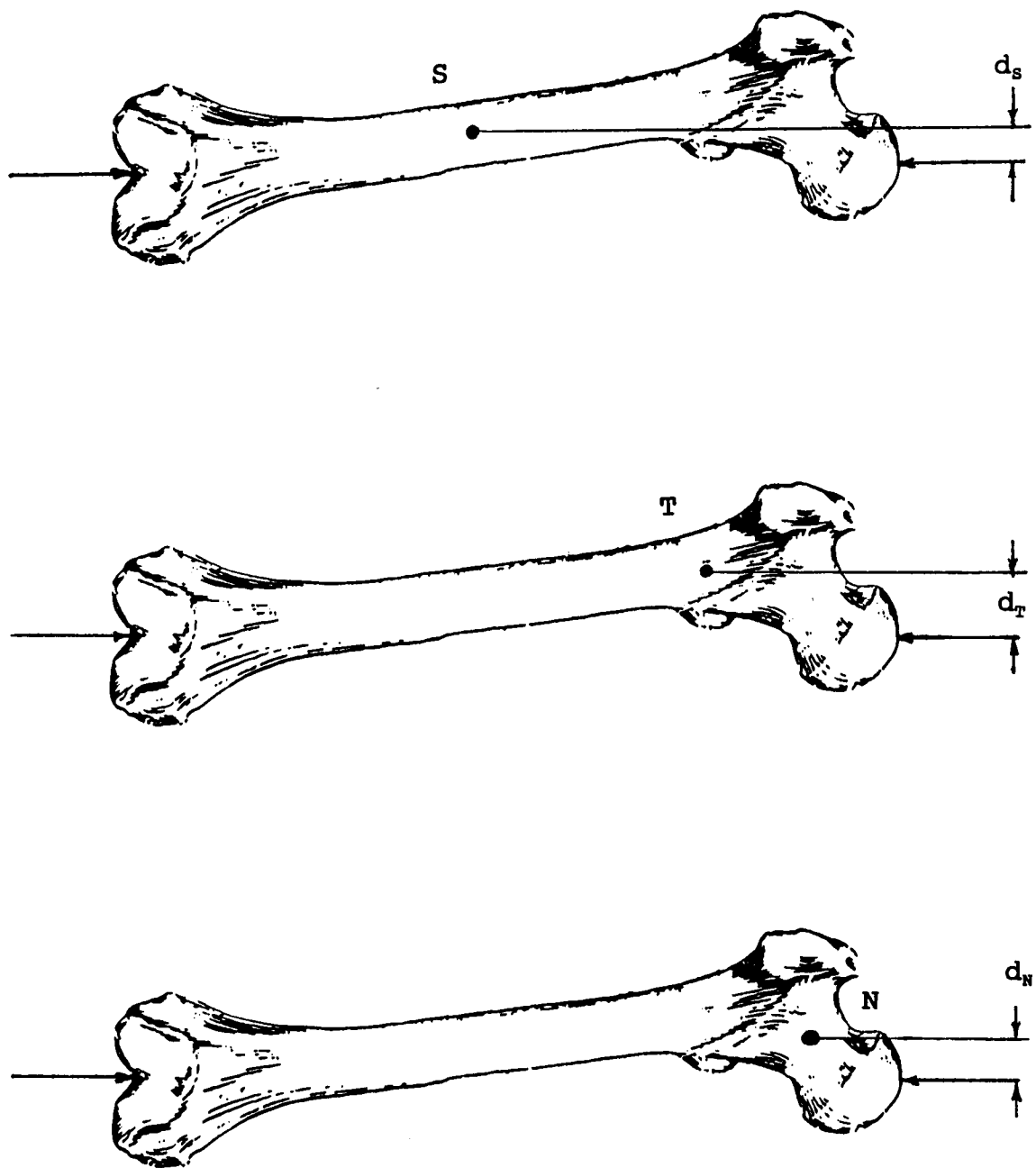


Figure 29. Diagram showing moment arm, d , for each desired stress calculation.

E. TEST IV: FRESH TISSUE TESTING

As has been stated consistently throughout this thesis, the type of preservation technique used on the tested bones effects the properties of the bones. In an effort to begin to examine thesis effect, 14 fresh bones were tested in the crash simulator. These bones were cryogenically frozen just after death and were wet-thawed shortly before testing.

Breaking force and area-under-curve data was obtained for each bone. Anatomical measurements were not taken due to diseases in the bones.

CHAPTER V

RESULTS AND DISCUSSION

Results from 94 bone-breaking tests are presented. Embalmed, bare femurs have been broken at high speeds (approximately seven meters per second) in the lateral-medial, anterior-posterior, and axial directions. They have been broken at low strain rates in the axial direction and in torsion. They have also been impacted in the lateral-medial direction while being subjected to a torsional preload. Fresh bones have also been tested.

A. TEST I: BREAKING STRENGTH OF FEMUR

1. Lateral-medial loading

Twelve femurs have been impacted in the crash simulator. A sample output plot from the crash simulator can be seen in Figure 30. The initial peak represents the event, the top of the peak being the breaking force. As can be seen, the entire event occurs in less than one millisecond. The peaks following the event are artifacts.

Table 6 shows the cortex thickness, support length, and breaking force for each of the femurs tested. Anatomical measurements were not available for three of the bones, and a force signal was not obtained for three bones. Breaking

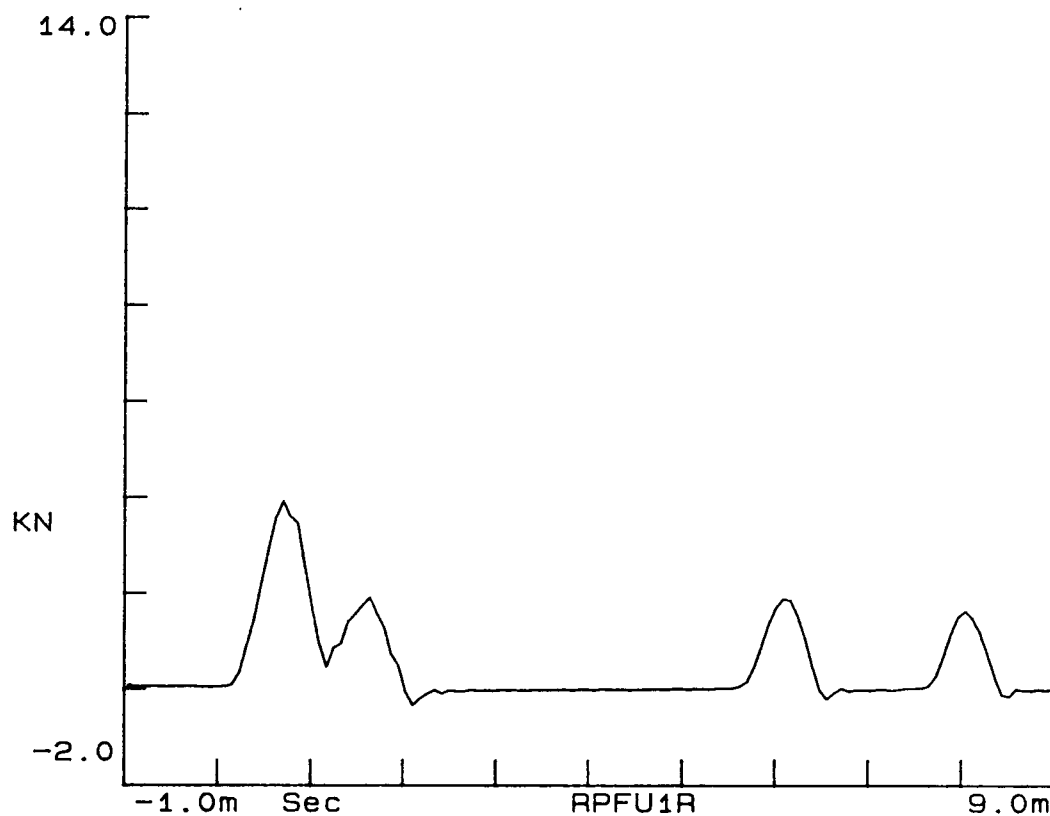


Figure 30. Sample force-time curve for femur impacted in l-m direction.

Table 6

Independent Variables and Force Data
for Femurs Loaded in Lateral-
Medial Direction

Bone	Cortex Thickness (meters)	Support Distance (meters)	Breaking Force (Newtons)
RPF698L	.00725	.432	---
RPF746R	---	---	---
RPF17R	---	---	---
RPF698R	---	---	4572
RPFU1R	.00673	.406	3928
RPF704R	.00580	.330	1832
RPFU2R	.00620	.330	1535
RPFU3R	.00839	.330	4262
RPFNO2R	.00602	.330	1696
RPFU4R	.00713	.343	1703
RPFU5R	.00779	.356	2169
RPFU6R	---	.381	<u>5779</u>
AVERAGE			3053

force vs. average cortex thickness is plotted in Figure 31. Linear regression reveals a relationship of

$$\text{Breaking Force (Newtons)} = -2510 + 722 \times \text{Cortex (mm)}$$

The linear correlation coefficient is 0.61. The second order correlation coefficient improves to 0.82. The least squares relationship is

$$\text{Breaking Force (N)} = -22 + 52 \times \text{cortex}^2 \text{ (mm)}$$

The ultimate bending strength and Young's modulus were calculated by the method described in Chapter IV. Table 7 contains this data. The average breaking force was 3053 Newtons; the average bending strength was 147 Megapascals; and the average Young's modulus was 30 Gigapascals. These values compare favorably to values found in literature (see Table 8).

Yamada (1971) found a static breaking force of 2453 newtons when testing in the anterior-posterior direction. The difference in these two values can be explained. Yamada's test was performed at low strain rates, while the 3053 value was performed at a strain rate of eight (Kress, 1989). Also, Yamada's test were performed in the a-p direction, and he did not use whole femurs.

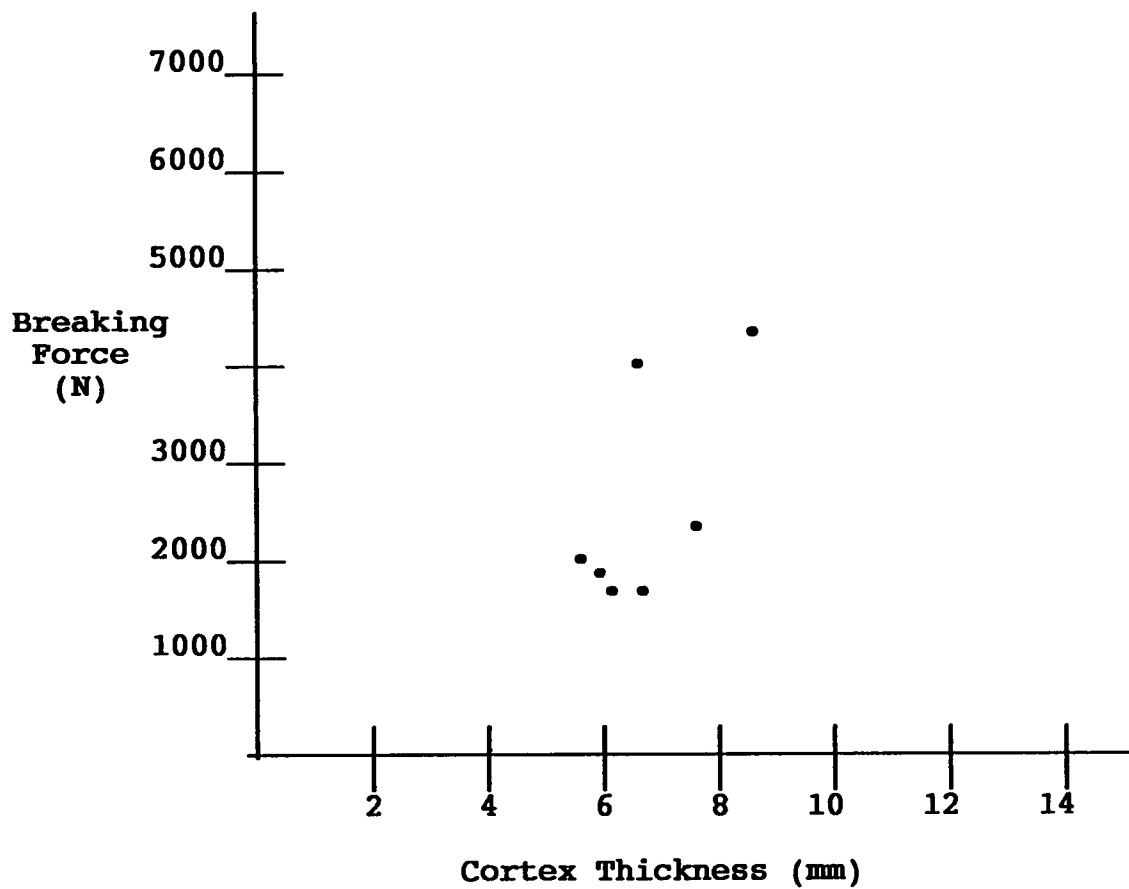


Figure 31. Plot of force vs. cortex thickness for femurs loaded in l-m direction.

Table 7

**Breaking Load, Ultimate Bending Stress,
and Young's Modulus for Femurs Loaded
in Lateral-Medial Direction**

Bone	Breaking Force (Newtons)	Ultimate Bending Stress (Mpa)	Young's Modulus (Gpa)
RPFU1R	3928	232	65
RPF704R	1832	112	17
RPFU2R	1535	119	20
RPFU3R	4262	194	35
RPFNO2R	1696	121	20
RPFU4R	1703	142	27
RPFU5R	<u>2169</u>	<u>110</u>	<u>24</u>
AVERAGE	3053	147	30

Table 8

**Comparison of Results from L-M Impact
with Those of Previous Researchers**

Property	Value	
Breaking Force		
Tucker (strain rate=8)	3053	N
Yamada, 1971 (static)	2453	N (a-p)
Ultimate Bending Stress		
Tucker	147	MPa
Yamada, 1971 (static, fresh, not whole bone)	186	MPa (a-p)
Young's Modulus		
Tucker	30	GPa
McElhaney, 1966 (sr=1-300)	22-30	GPa (a-p)

Yamada (1971) also found the ultimate bending stress of the femur to be 186 MPa. The discrepancy here cannot be explained by strain rate since the impact result is lower than that for the slowly loaded tested bone. The explanation instead lies in the direction of loading. As will be reported later, the 186 MPa value is correct for an anterior-posterior load. Bone, however, appears to be much weaker in the l-m direction than in the a-p direction. Little work has been performed loading on the femur in the l-m direction.

The calculated Young's modulus of 30 GPa is exactly in the range McElhaney (1966) reports for similar strain rates.

Table 9 lists the area under the force-time curve for each tested bone. The average area is 2236 N·ms. This value is of little real significance. It is, however,

Table 9

**Area Under the Force-Time Curve
for L-M Impacted Femurs**

Bone	Area (N·ms)
RPFU1R	2711.1
RPF704R	1725.3
RPFU2R	965.5
RPFU3R	5727.6
RPFNO2R	1808.9
RPFU4R	1294.0
RPFU5R	<u>1368.8</u>
AVERAGE	2236

roughly related to the amount of energy absorbed during impact and can be compared to other area under force-time curve calculations.

Six of the twelve fractures were comminutions (see Table 10). Fractures are important, because they determine

Table 10

Types of Fractures Occurring in
L-M Loaded Femurs

Fracture Type	Number	Percentage
Comminution		
Tension Wedge	5	41.7 %
Compression Wedge	1	8.3
Oblique	3	25.0
Spiral	2	16.7
Shatter	1	8.3

soft tissue damage. Most of the comminutions produced tension wedges, that is the wedge was shot out of the side of the bending bone in tension. Oblique and spiral fractures occurred. The spiral fractures were probably caused by the specimen holding device which also served as a torsional delivery system. Its configuration alone may have encouraged a spiral fracture. One bone shattered on impact. This was a result of severe osteoporosis.

2. Anterior-posterior loading

Directional changes in the properties of the bone were investigated by turning the femurs 90 degrees and striking them in the anterior-posterior direction. The crash

simulator was used to break 32 femurs by this method. More femurs were used in this test than in the lateral-medial test for several reasons. First of all, this set up was much simpler. The specimen loading device for lateral-medial hits was quite cumbersome. More importantly, age information was available on most of these 32 bones. It was therefore important to perform the same test on the bones, enabling the dependence of bone properties on age to be studied.

A sample force-time plot can be viewed in Figure 32. As can be seen, this signal is very clean, owing to the constant improvements made in instrumentation on the crash simulator. Anatomical measurements, support distance, and breaking force data for all of the bones tested are listed in Table 11. Figure 33 illustrates the cortex thickness relationship with breaking force. The linear regression works out to be

$$\text{Breaking Force (Newtons)} = 277 + 740 \times \text{cortex (mm)}.$$

with a linear correlation coefficient of 0.40. The second order correlation of 0.42 only improves the relation slightly. This polynomial curve fit is

$$\text{Breaking Force (N)} = 2827 + 52 \times \text{cortex}^2 \text{ (mm)}.$$

Note that the slope of this line is the same as the slope of the l - m regression line. And the second order coefficients for the two sets are equal.

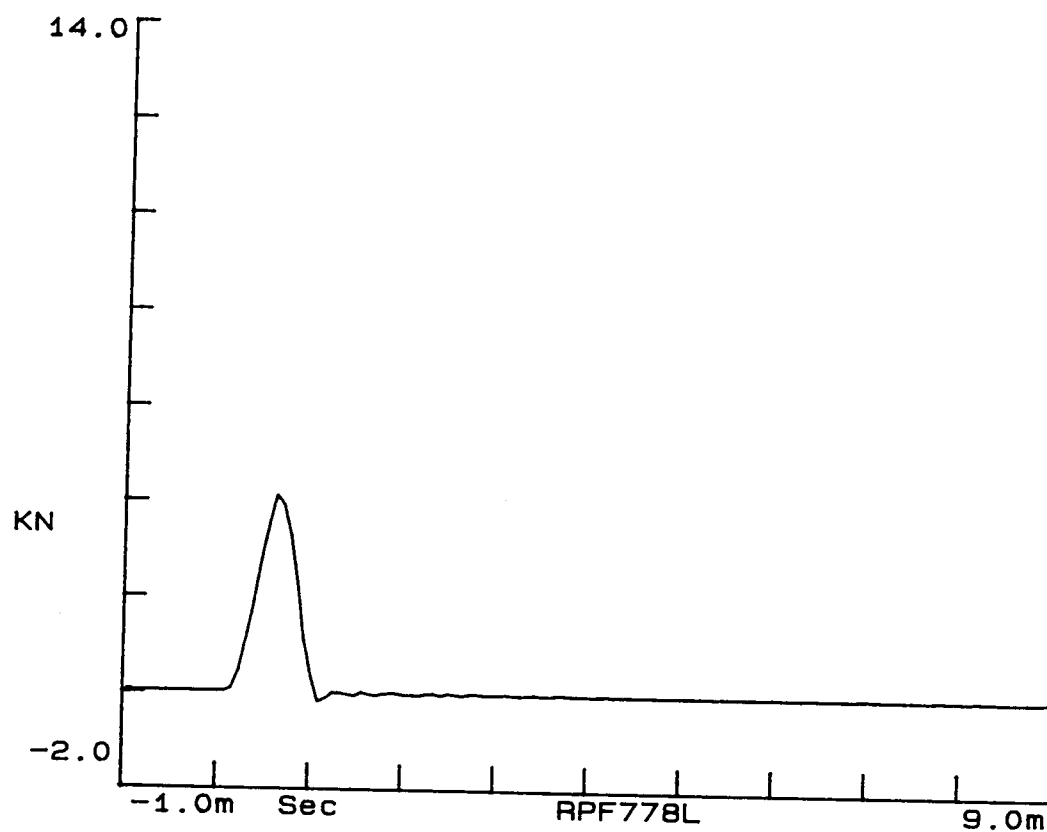


Figure 32. Sample force-time plot for femur impacted in a-p direction.

Table 11

Independent Variables and Force Data
for Femurs Loaded in Anterior-
Posterior Direction

Bone	Cortex Thickness (meters)	Support Distance (meters)	Breaking Force (Newtons)
RPF783R	.00930	.430	7814
RPF756R	.00698	.390	3185
RPF723R	.00744	.420	4819
RPF738R	.00454	.390	5484
RPF752R	.00868	.400	6050
RPF781R	.00588	.410	3658
RPF798R	.00785	.490	8911
RPF778R	.00701	.410	4092
RPF754R	.00891	.430	6378
RPF720R	.00995	.420	7626
RPF760R	.00659	.420	4423
RPF828R	.00816	.415	6751
RPF779R	.00820	.420	6608
RPFU1R	.00809	.375	5213
RPF553L	.00831	.455	5065
RPF783L	.00824	.430	8205
RPF738L	.00670	.400	4568
RPF709L	.00680	.385	6138
RPF776L	.00629	.400	--
RPF828L	.00889	.410	8029
RPF778L	.00689	.400	4130
RPF754L	.00685	.430	6177
RPF779L	.00602	.425	--
RPF781L	.00587	.410	5196
RPF723L	.00652	.425	6790
RPF760L	.00567	.415	4003
RPF720L	.00781	.430	1443
RPF752L	.00798	.410	6422
RPF798L	.00798	.490	<u>9946</u>
AVERAGE			5697

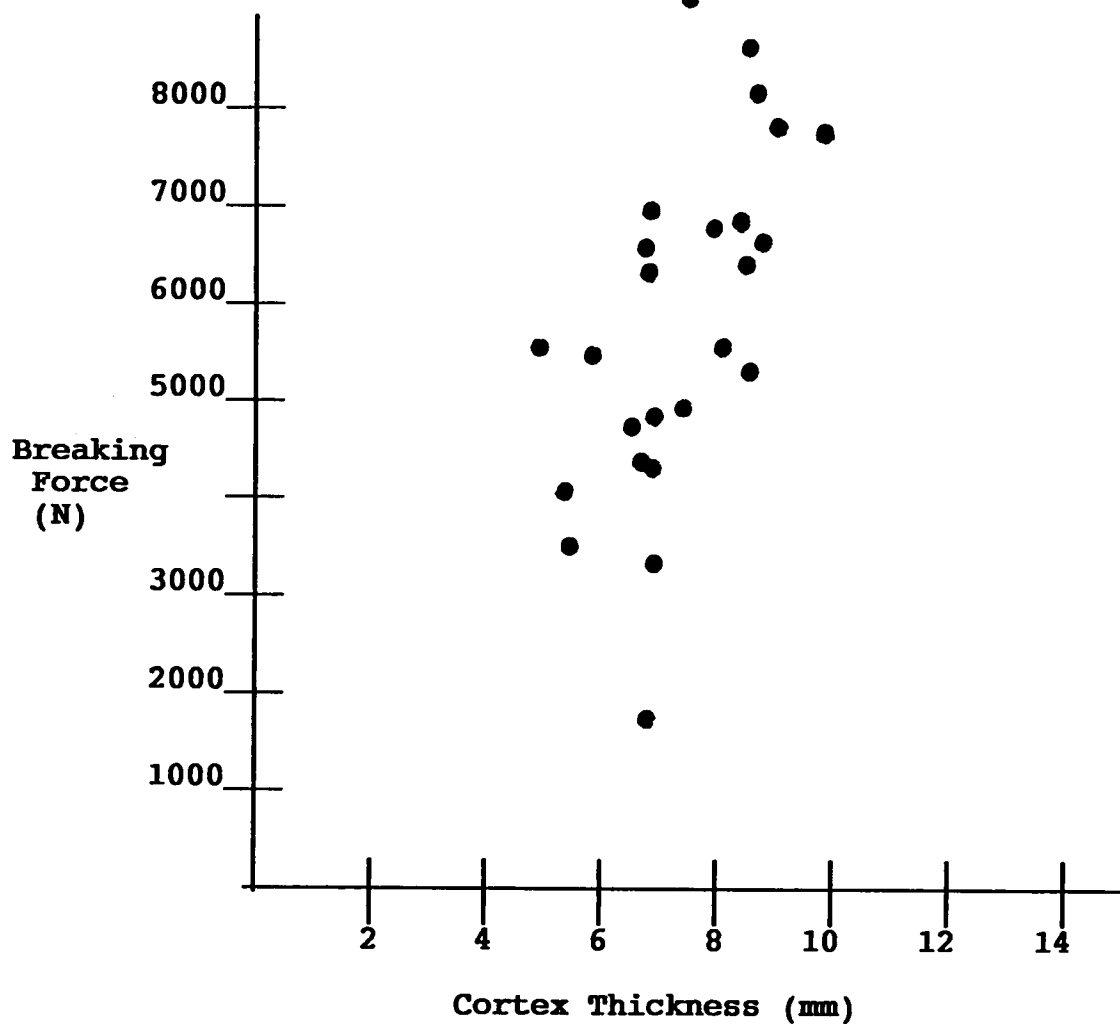


Figure 33. Plot of breaking force vs. cortex thickness for a-p impact on femur.

The ultimate bending stress and Young's modulus were calculated and are listed along with the breaking force in Table 12. The average breaking force was 5697 Newtons. The average bending stress was 284 Megapascals, and the average Young's modulus was 88 Gigapascals.

Table 13 compares these values with previous researcher's values. Of the most interest in this table is the ultimate bending stress data. McElhaney (1966) reports a 53 percent increase in bending stress in going from low strain rates to a strain rate comparable to this experiment. Using Yamada's (1973) static value of 186 MPa, this equates out to a theoretical impact value of 285 MPa. The calculated strength in this experiment was 284 MPa.

The breaking force value was much higher than Yamada's. This is most likely due to bone geometry. Both tests certainly seem valid since the strength data is so comparable. The Young's modulus calculated here is very high. This is most likely due to the method with which it is obtained being very crude.

Bending strength and stiffness are listed by age of bone donor at death in Table 14. Figure 34 plots age vs. breaking strength, and Figure 35 plots age vs. Young's modulus. Linear regression lines can be drawn through these points, but the correlation coefficient would be so low that

Table 12

Breaking Load, Ultimate Bending Stress
and Young's Modulus for Femurs Loaded
in Anterior-Posterior Direction

Bone	Breaking Force (Newtons)	Ultimate Bending Stress (MPa)	Young's Modulus (GPa)
RPF783R	7814	289	96
RPF756R	3185	159	35
RPF723R	4819	262	89
RPF738R	5484	409	124
RPF752R	6050	181	40
RPF781R	3658	210	59
RPF798R	8911	408	153
RPF778R	4092	239	76
RPF754R	6378	249	74
RPF720R	7626	304	81
RPF760R	4423	292	105
RPF828R	6751	342	102
RPF779R	6608	233	68
RPFU1R	5213	306	81
RPF553L	5065	246	88
RPF783L	8205	344	107
RPF738L	4568	274	92
RPF709L	6138	436	105
RPF828L	8029	370	113
RPF778L	4130	201	56
RPF754L	6177	282	89
RPF781L	5196	295	85
RPF723L	6790	375	138
RPF760L	4003	270	85
RPF720L	1443	62	19
RPF752L	6422	201	55
RPF798L	<u>9946</u>	<u>422</u>	<u>171</u>
AVERAGE	5697	284	88

Table 13

Comparison of Results from A-P Impact
with Those of Previous Researchers

Property	Value
Breaking Force	
Tucker (strain rate=8)	5697 N
Yamada, 1971 (static)	2453 N
Ultimate Bending Stress	
Tucker	284 MPa
Yamada, 1971 combined with McElhaney, 1966	285 MPa
Young's Modulus	
Tucker	88 GPa
McElhaney, 1966 (sr=1-300)	22-30 GPa

Table 14

Ultimate Bending Stress and Young's Modulus of
Femurs in Relation to Age

Age (years)	Ultimate Bending Stress (MPa)	Young's Modulus (GPa)
53	409	124
53	274	92
56	370	113
56	342	102
58	249	74
58	282	89
59	233	68
64	239	76
64	201	56
68	304	81
68	292	105
68	270	85
68	62	19
73	436	105
74	408	153
76	289	96
76	344	107
79	262	89
79	375	138
82	210	59
82	295	85
84	181	40
84	201	55
89	159	35

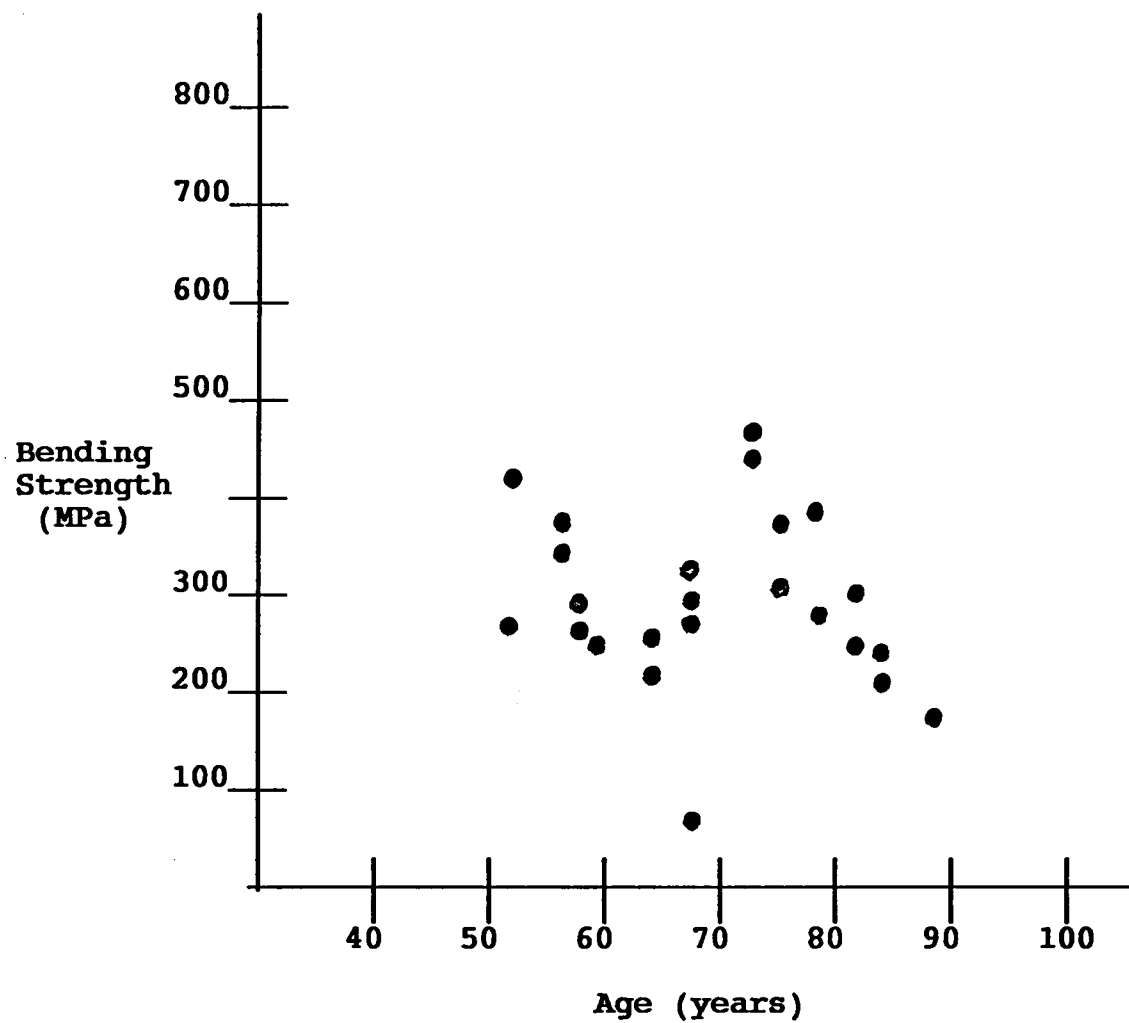


Figure 34. Plot of breaking strength vs. age for a-p impacted femur.

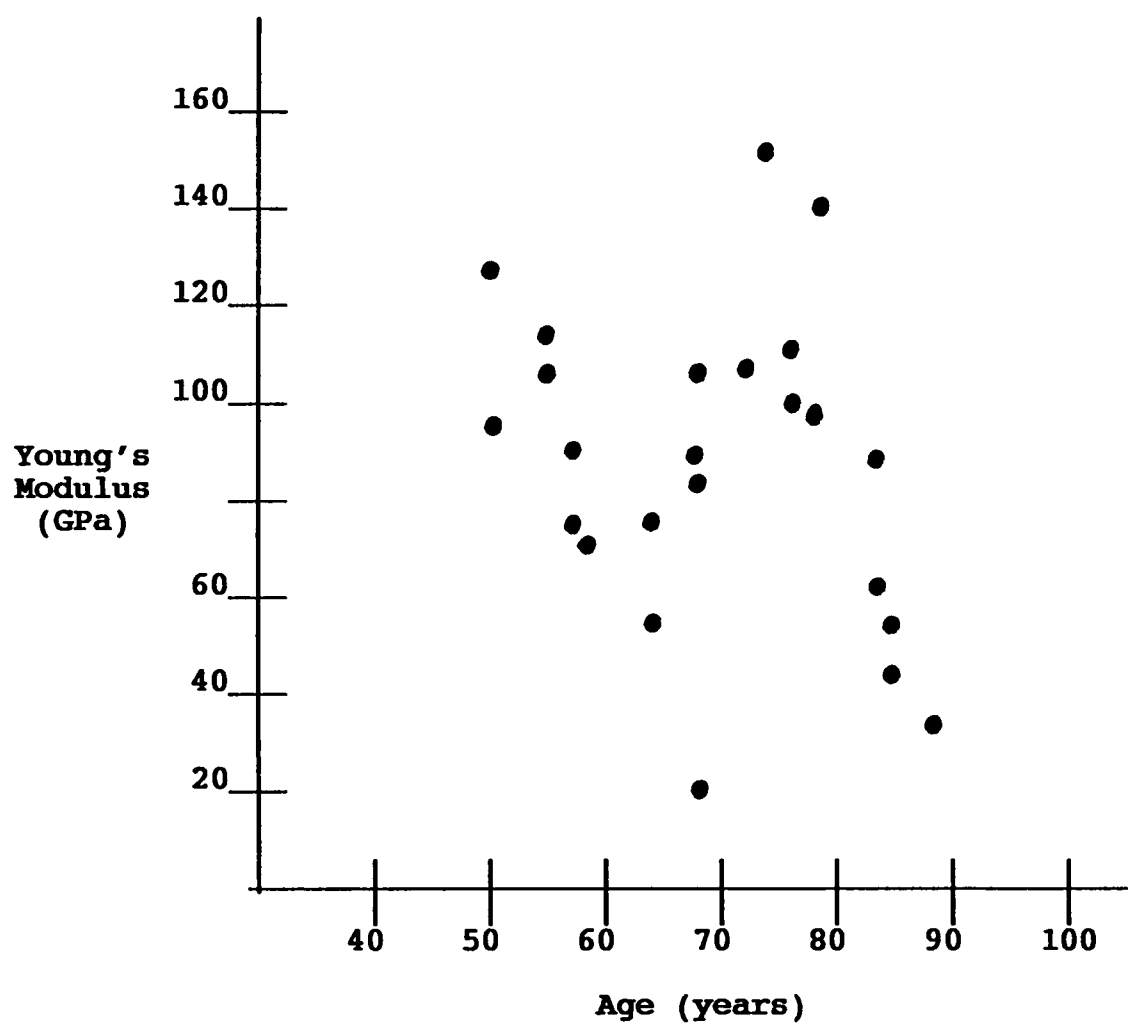


Figure 35. Plot Young's modulus vs. age for a-p impacted femur.

the lines would be meaningless. This scatter of points indicates no real age dependence in this range of age.

Table 15 compares the breaking force and the breaking stress of right and left matched pairs of femurs (two femurs belonging to the same individual). The obvious conclusion, surprisingly, is that the left bone is roughly eight percent stronger on average than the right bone. However, a closer

Table 15

Comparison of Right and Left Matched
Pairs in A-P Loading

Bone	Breaking Force (N)			Breaking Strength (MPa)		
	right	left	%	right	left	%
783	7814	8205	-5%	289	344	-19%
723	4819	6790	-41	262	375	-43
738	5484	4568	17	409	274	33
752	6050	6422	6	181	201	-11
781	3658	5196	-42	210	295	-41
798	8911	9946	-12	408	422	-3
778	4092	4130	-1	239	201	16
754	6378	6177	3	249	282	-13
760	4423	4003	10	292	270	8
828	6751	8029	-19	342	370	-8
AVERAGE			-8.4%			-8.2%

examination of the results reveals two sets of bones (723 and 781) where the left bone happened to be significantly stronger than the right for whatever reason. If these two bones are thrown out of the averaging, the right and left bones are virtually equal.

Area under the force-time curve for each bone is listed in Table 16. The average value is 2658 N.ms. The value for the l-m loaded femurs was 2254 N.ms. The a-p loaded bone, therefore, does not absorb much more energy than the l-m loaded bone, although the strength is much greater in the a-p direction.

Seventy-one percent of the fractures in this a-p test were comminuted (see Table 17). Interestingly, however, few produced tension (3) or compression (2) wedges. The vast majority (17) of the comminuted fractures were side wedges (see Figures 36 and 37). The side wedges were equally dispursed as lateral and medial wedges. Eight fractures were oblique, and one was a shatter.

Table 16

Area Under the Force-Time Curve for
A-P Impacted Femurs

Bone	Area (N·ms)
RPF783R	4853.9
RPF756R	1834.6
RPF723R	2087.6
RPF738R	2560.7
RPF752R	2809.3
RPF781R	1697.8
RPF798R	3902.1
RPF778R	1890.1
RPF754R	2691.8
RPF720R	5010.3
RPF760R	2063.6
RPF828R	2608.0
RPF779R	2852.9
RPFU1R	2429.4
RPF553L	2560.3
RPF783L	3483.0
RPF738L	2301.2
RPF709L	2601.9
RPF828L	3845.8
RPF778L	2080.8
RPF754L	2619.5
RPF781L	2201.6
RPF723L	3170.4
RPF760L	1882.3
RPF720L	814.5
RPF752L	2674.6
RPF798L	<u>3835.7</u>
AVERAGE	2658

Table 17

Types of Fractures Occurring in
A-P Loaded Femurs

Fracture Type	Number	Percentage
Comminution		
Tension Wedge	3	9.6%
Compression Wedge	2	6.5
Side Wedge	17	54.8
Oblique	8	25.8
Shatter	1	3.2

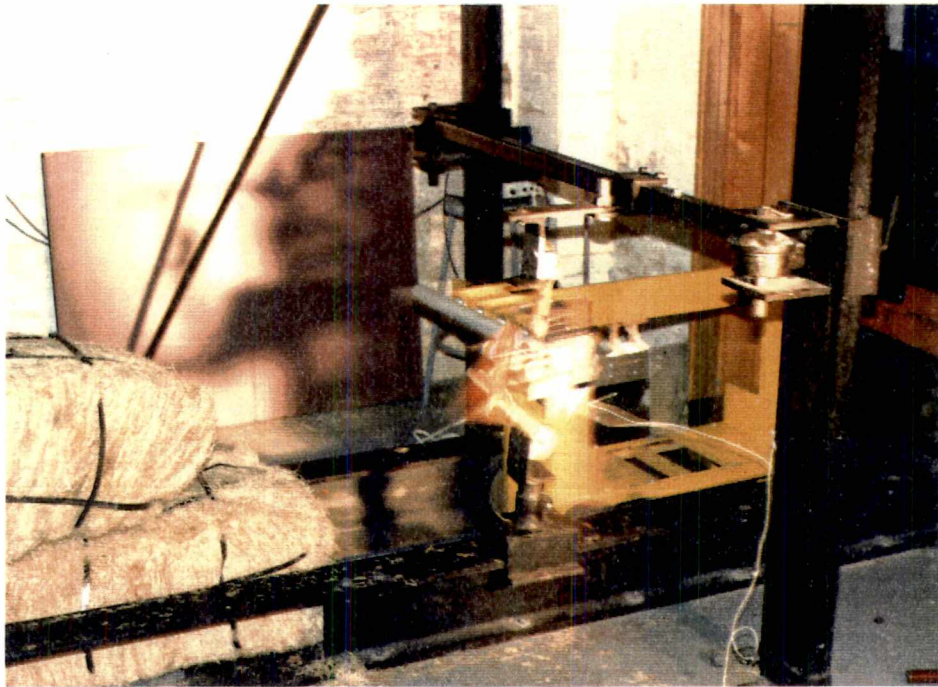


Figure 36. High-speed impact of human femur in a-p direction. Notice side comminution.



Figure 37. Human femur following high-speed impact.
Notice side comminution.

B. TEST II: TORSIONAL STRENGTH OF FEMUR

1. Steady-state torsional loads

Six femurs were loaded in torsion at low strain rates. The average cortex thickness, support distance and maximum torque for each bone is presented in Table 18. A plot of

Table 18

Independent Variables and Torque Data
on Femurs Loaded in Torsion at
Low Strain Rates

Bone	Cortex Thickness (meters)	Support Distance (meters)	Maximum Torque (N·m)
1	.00564	.381	96.0
2	.00758	.381	154.6
3	.00628	.318	113.8
4	.00894	.356	115.9
5	.00959	.387	145.0
6	.00201	.330	24.4
AVERAGE			108

maximum torque vs. cortex thickness is illustrated in Figure 38. The linear relation, with a correlation coefficient of 0.90, is given by

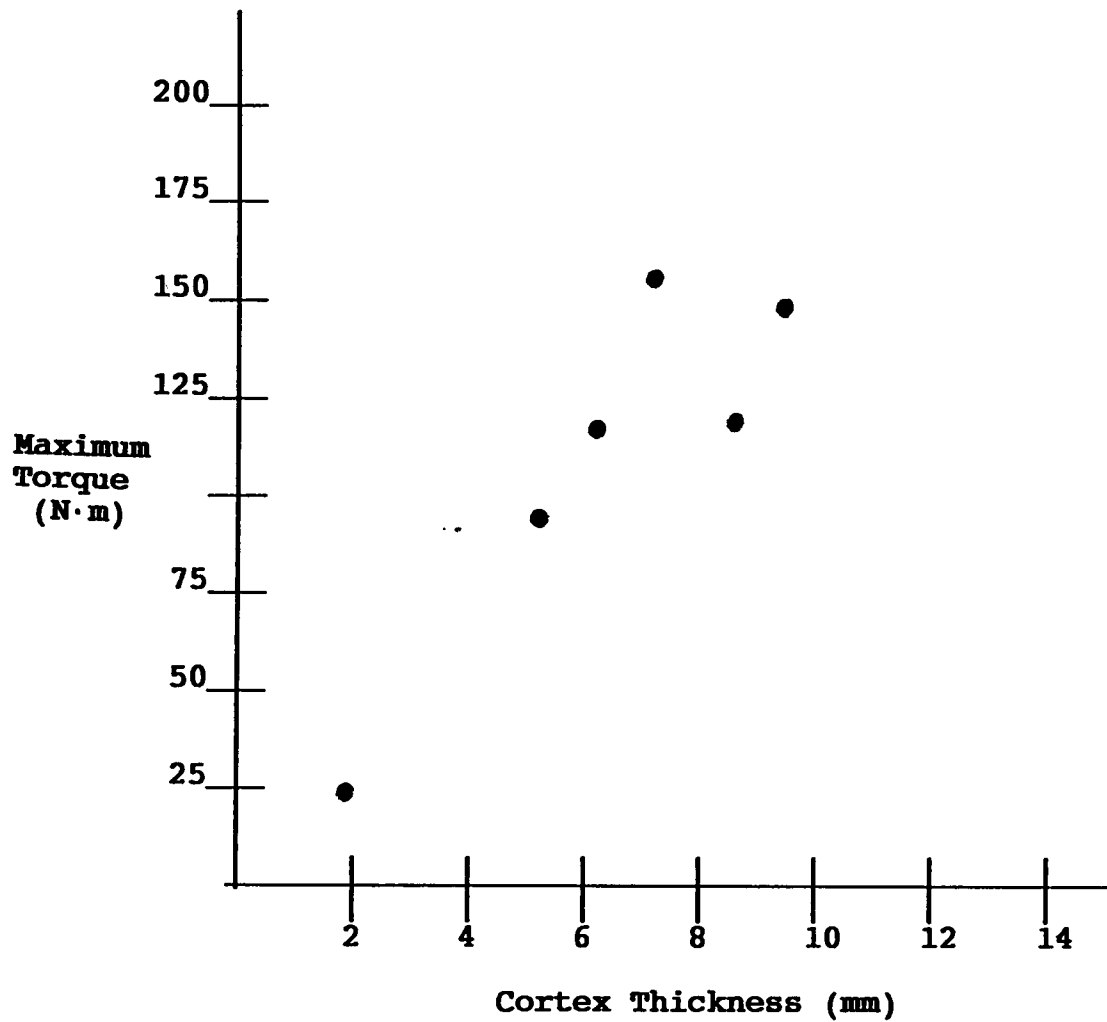


Figure 38. Plot of breaking torque vs. cortex thickness for femur loaded statically in torsion.

$$\text{Maximum Torque (N}\cdot\text{m)} = 6 + 15 \times \text{cortex (mm)}$$

Using a polynomial curve fit improves the relation to 0.92.
The least squares fit is

$$\text{Maximum Torque (N}\cdot\text{m)} = 49 + 1.2 \times \text{cortex}^2 \text{ (mm)}$$

Ultimate torsional stress was calculated and is shown in Table 19. The average breaking torque of 108 N·m is slightly lower than Yamada's (1971) (see Table 20). If, however, bone six is removed from the average, the breaking torque comes up to 125 N·m. Bone six was highly osteoporotic.

The torsional strength of 28 MPa is also lower than Yamada's value of 45 MPa. This can be accounted for by the use of embalmed bones rather than fresh, wet bones.

As Table 21 shows, five of the six femur fractures were spiral. The osteoporotic bone six shattered.

Table 19

Breaking Torque and Ultimate Torsional Stress
for Femurs Loaded at Low Strain Rates

Bone	Breaking Torque (N·m)	Ultimate Torsional Stress (Mpa)
1	96.0	28
2	154.6	31
3	113.8	31
4	115.9	23
5	145.0	28
6	<u>24.4</u>	<u>15</u>
AVERAGE	108	26

Table 20

Comparison of Results from Slowly Applied Torsion
with Those of Previous Researchers

Property	Value
Maximum Torque	
Tucker (without osteoporotic)	125 N·m
Yamada, 1971 (fresh)	137 N·m
Ultimate Torsional Strength	
Tucker	28 MPa
Yamada, 1971 (fresh)	45 MPa

Table 21

**Types of Fractures Occurring in
Torsionally Loaded Femurs**

Fracture Type	Number	Percentage
Spiral	5	83.3%
Shatter	1	16.7

2. Torsional preload with l-m impact

The question of combined loads on the femur is investigated by loading six bones in torsion and in bending. A torsional preload was placed on the bones, then the crash simulator impacted the bones in the lateral-medial direction.

Cortex thickness, support distance, amount of preload and breaking force are shown in Table 22. Notice that two different torque levels were used.

The important result in this experiment is depicted in Table 23. All six bones used in this test were the matched pair to six bones tested in the lateral-medial direction without a torsional preload. This table shows the effect the preload had on each bone. The average breaking force

Table 22

**Independent Variables and Torque Data
for Femurs Loaded in L-M Bending
with Torsional Preload**

Bone	Cortex Thickness (meters)	Support Distance (meters)	Torsional Preload (N·m)	Breaking Force (N)
RPFTU1L	.00611	.41	20.2	3657
RPFTU2L	.00586	.33	10.1	1355
RPFTU3L	.00702	.33	10.1	3335
RPFTNO2L	.00598	.33	10.1	1684
RPFTU4L	.00810	.34	20.2	1234
RPFTU5L	.00743	.36	20.2	1806

with no torsional preload was 2549 Newtons, and the average with a preload was 2179 Newtons.

Table 24 is a little difficult to understand, but it is very important. The first column tells the bone number for a matched pair of tested bones. Knowing the cortex thickness of the bone and the torsional preload applied to it, the percentage of the maximum torsional load for that bone is determined from Figure 38. This percentage is in column two. Column three shows the decrease in the breaking force in going from the non-preloaded bone to the preloaded bone. Column four is the percent decrease. On the average,

Table 23

**Matched Pairs of Femurs in L-M Bending-
One with Torsional Preload, One without**

Bone	Torsional Preload (N·m)	Breaking Force (N)
RPFTU1L	20.2	3657
RPFU1R	0	3928
RPFTU2L	10.1	1355
RPFU2R	0	1535
RPFTU3L	10.1	3335
RPFU3R	0	4262
RPFTNO2L	10.1	1684
RPFNO2R	0	1696
RPFTU4L	20.2	1234
RPFU4R	0	1703
RPFTU5L	20.2	1806
RPFU5R	0	<u>2169</u>
AVERAGE WITH TORQUE		2179
AVERAGE WITH NO TORQUE		2549

Table 24

Effect of Torsional Preload on
Matched Femur Pairs

Bone Pair	Percentage of Maximum Torque (%)	Decrease in Breaking Force (N)	% Decrease in Breaking Force (%)
U1	20.1%	271	6.9%
U2	10.4	180	11.7
U3	8.8	927	21.8
NO2	10.2	12	.7
U4	15.5	469	27.5
U5	<u>16.8</u>	363	<u>16.7</u>
AVERAGE	13.6%		14.2%

a 14 percent torsional preload decreases the breaking force 14 percent. Interestingly, spiral fractures are present in 50 percent of these preloaded bones.

Figure 39 is the force-time plot for a matched pair. The solid line represents the bone with the preload. The dashed line represents the bone with no torsional preload. These curves lie nicely on top of each other.

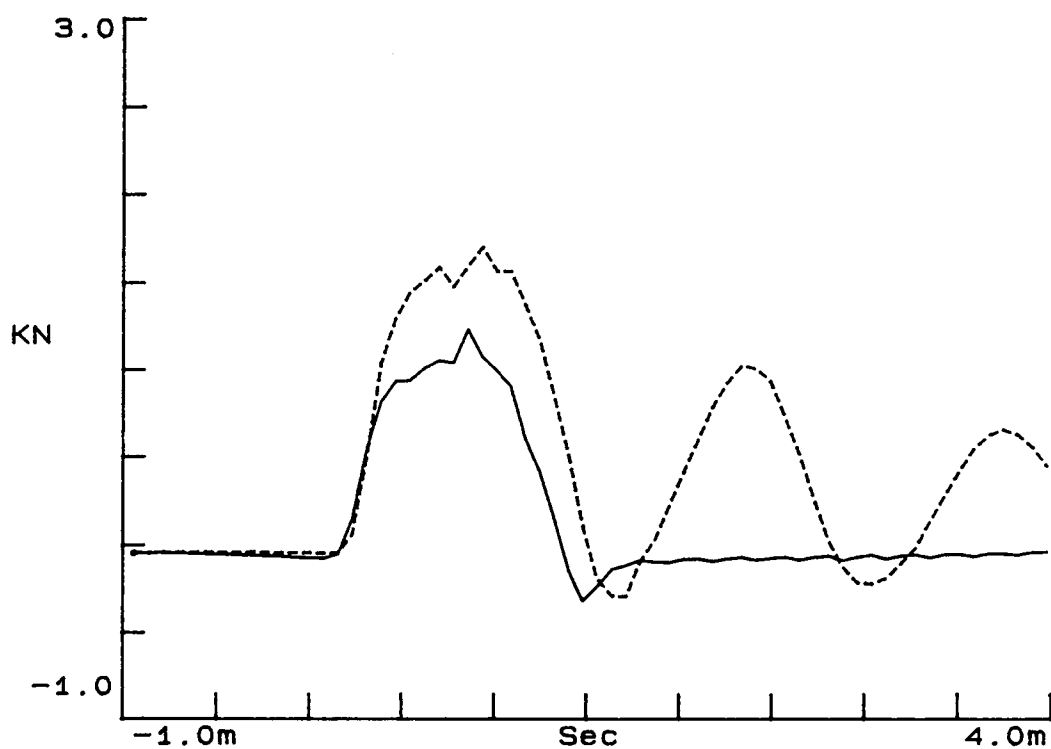


Figure 39. Plot of force-time curve for two femurs impacted in the l-m direction. The solid line represents the femur with a torsional preload. The dashed line represents a femur with no preload.

C. TEST III: COMPRESSIVE STRENGTH OF FEMUR

1. Axial loads at low strain rates

Nine femurs were loaded at low strain rates in the axial direction by a materials testing machine. The output from the machine is simply a force reading.

From the ultimate load and anatomical measurements, stresses can be calculated. Since fractures under these loading conditions occur most frequently at three locations (midshaft, sub-trochanter, and neck), stress measurements at all three of these cross-sections were made on each bone. This was discussed in Chapter IV. Figure 40 shows an example of how this calculation was made for bone 859R.

Figures 41-49 map out the stresses on each bone tested. Stresses on the lateral side of the bone are tensile, and those on the medial side are compressive. On each figure, based on the stresses, a prediction is made as to where the fracture will occur. Also on each figure is a drawing of the actual fracture location. On eight of the nine femurs the prediction is correct. Bone 881R is the only incorrect fracture. On this test the cup holding the head of the femur was impinging the neck leading to its fracture.

The results of the test are summarized in Table 25. The stress in bone 881R is omitted from the average calculation. Notice that in all cases the compressive stress is approximately 1.5 times greater than the tensile.

BONE: 859R

$$\sigma_x = \frac{P}{A} \pm \frac{MC}{I}$$

$$= \frac{P}{A} \pm \frac{Pdr}{I}$$

$$P = 10,578 \text{ N}$$

$$\begin{aligned} d_s &= 0.0270 \text{ m} \\ r_s &= 0.0160 \text{ m} \\ t_s &= 0.0076 \text{ m} \end{aligned}$$

$$\begin{aligned} d_T &= 0.0340 \\ r_T &= 0.0181 \\ t_T &= 0.0071 \end{aligned}$$

$$\begin{aligned} d_n &= 0.0210 \\ r_n &= 0.0176 \\ t_n &= 0.0039 \end{aligned}$$

$$\begin{aligned} M_s &= 285.6 \text{ N m} \\ I_s &= 4.76 \times 10^{-8} \\ A_s &= 0.00058 \end{aligned}$$

$$\begin{aligned} M_T &= 359.7 \\ I_T &= 7.28 \times 10^{-8} \\ A_T &= 0.00065 \end{aligned}$$

$$\begin{aligned} M_n &= 222.1 \\ I_n &= 4.77 \times 10^{-8} \\ A_n &= 0.00038 \end{aligned}$$

$$\sigma_{xs(comp)} = \frac{10578}{0.00058} + \frac{285.6(0.16)}{4.76 \times 10^{-8}} = 18 + 96 = 114 \text{ MPa}$$

$$\sigma_{xs(tens)} = \frac{10578}{0.00058} - \frac{285.6(0.16)}{4.76 \times 10^{-8}} = 18 - 96 = -78 \text{ MPa}$$

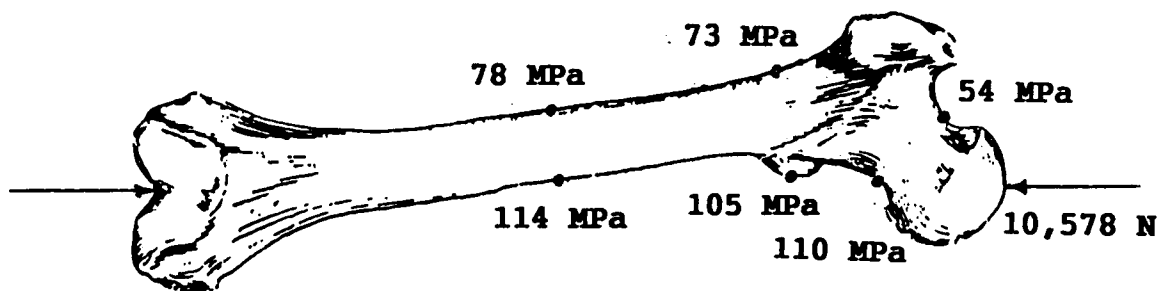
Figure 40. Example of stress calculations for axially loaded bone.

BONE: 859R

AGE: 52

SEX: MALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: SIMULTANEOUS NECK AND SHAFT FRACTURE

ACTUAL FRACTURE:

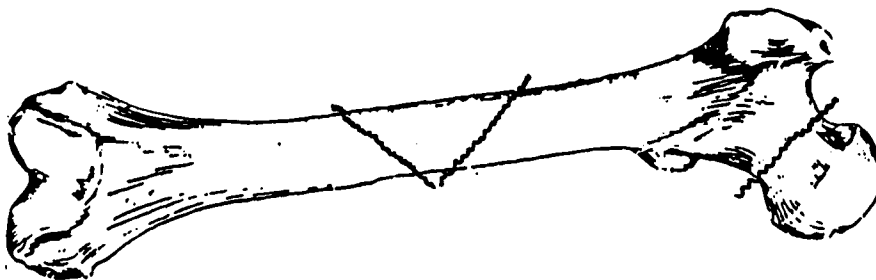


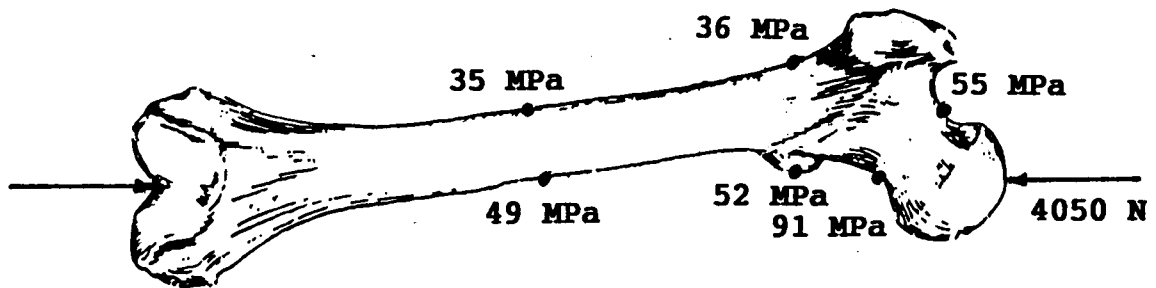
Figure 41. Analysis of low strain rate axial loading of femur 859R. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

BONE: 902R

AGE: 76

SEX: MALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: NECK FRACTURE

ACTUAL FRACTURE:

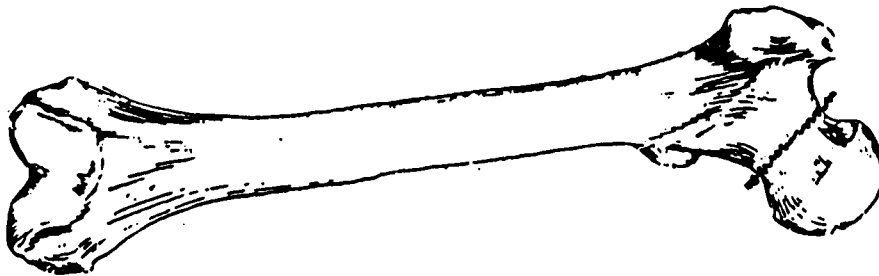


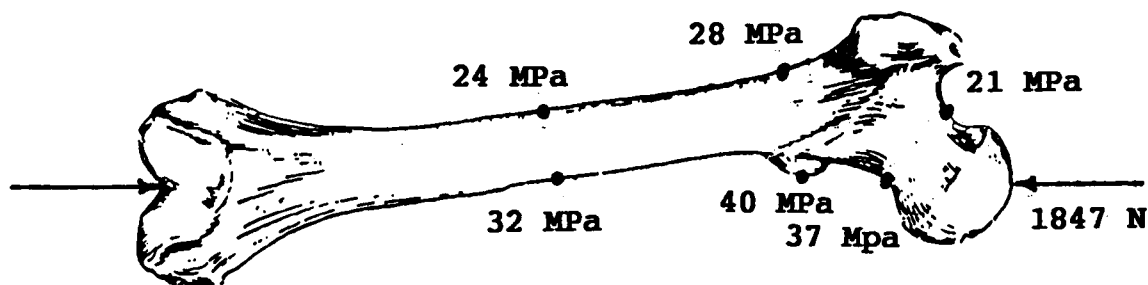
Figure 42. Analysis of low strain rate axial loading of femur 902R. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

BONE: 812R

AGE: 82

SEX: MALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: SUB-TROCHANTERIC FRACTURE

ACTUAL FRACTURE:

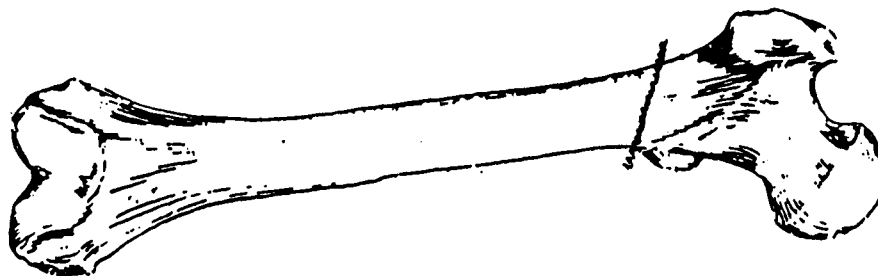


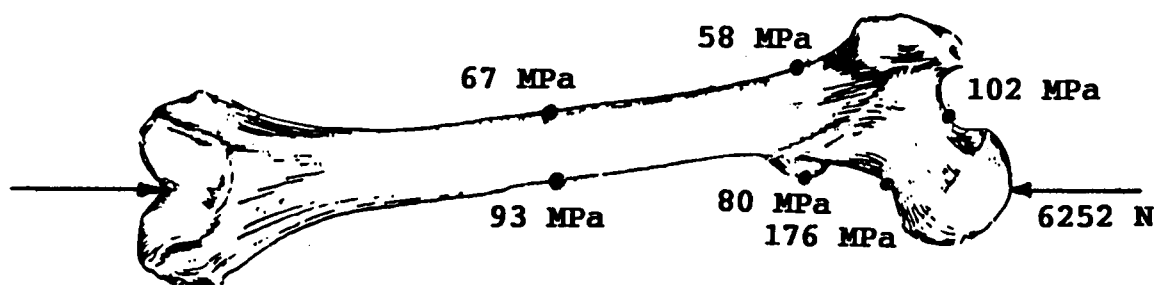
Figure 43. Analysis of low strain rate axial loading of femur 812R. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

BONE: 883R

AGE: 66

SEX: MALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: NECK FRACTURE

ACTUAL FRACTURE:

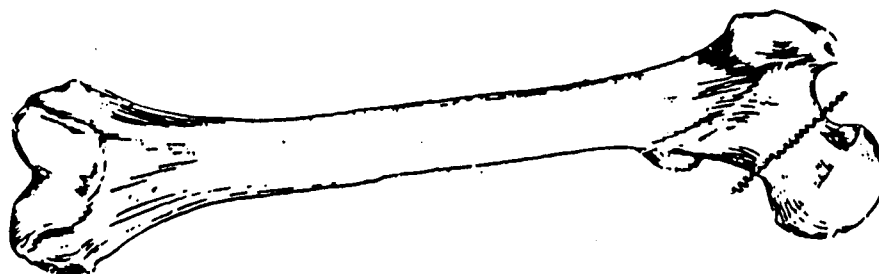


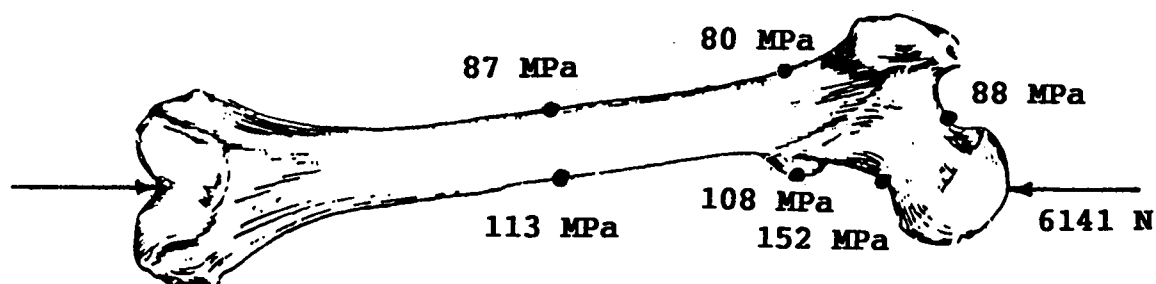
Figure 44. Analysis of low strain rate axial loading of femur 883R. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

BONE: 862R

AGE: 63

SEX: MALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: NECK FRACTURE

ACTUAL FRACTURE:

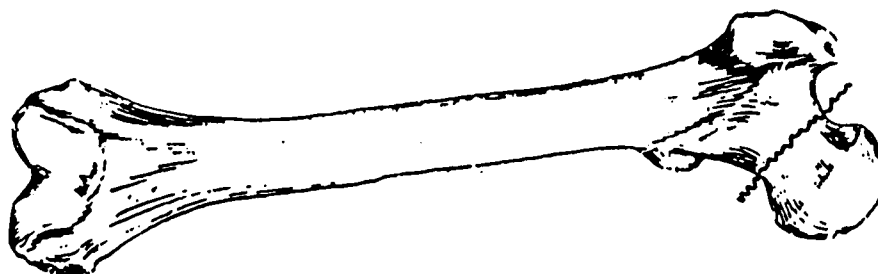


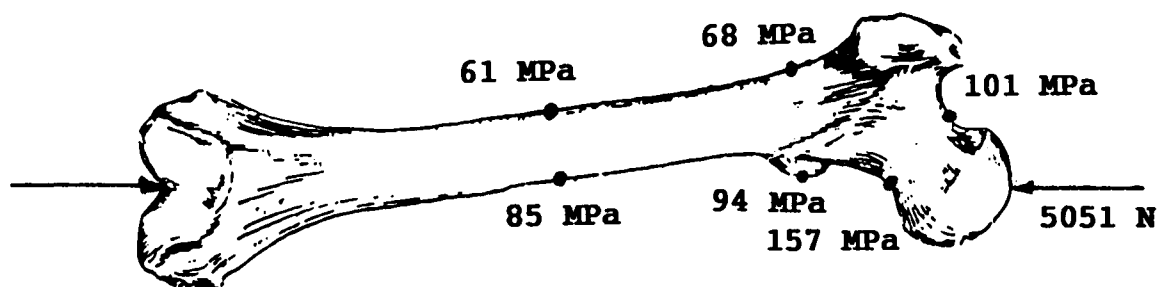
Figure 45. Analysis of low strain rate axial loading of femur 862R. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

BONE: 865R

AGE: 63

SEX: FEMALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: NECK FRACTURE

ACTUAL FRACTURE:

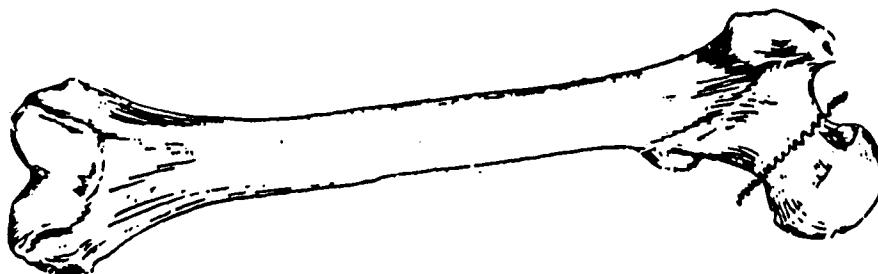


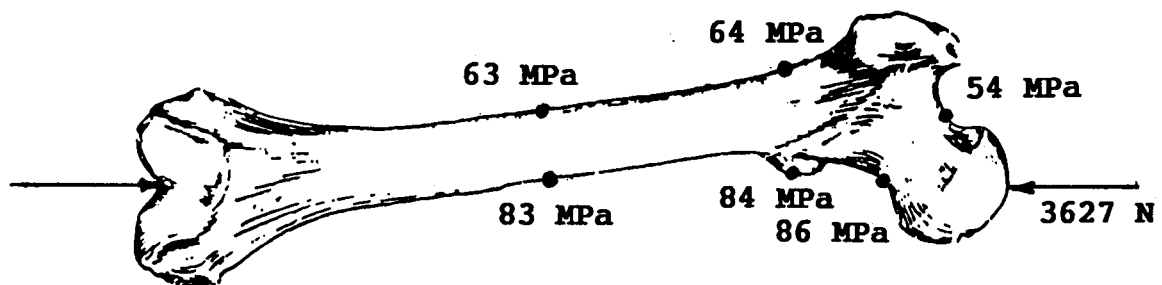
Figure 46. Analysis of low strain rate axial loading of femur 865R. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

BONE: 724R

AGE: 58

SEX: MALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: EQUAL DISTRIBUTION

ACTUAL FRACTURE:

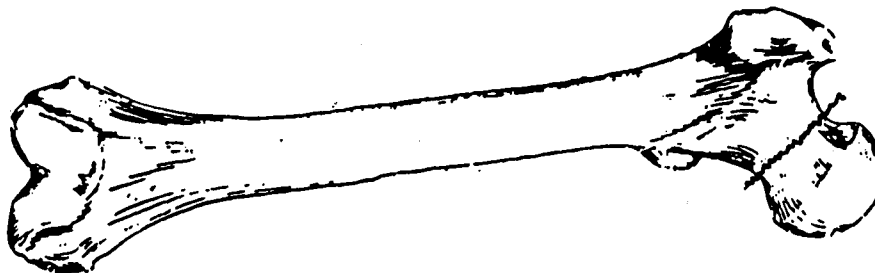


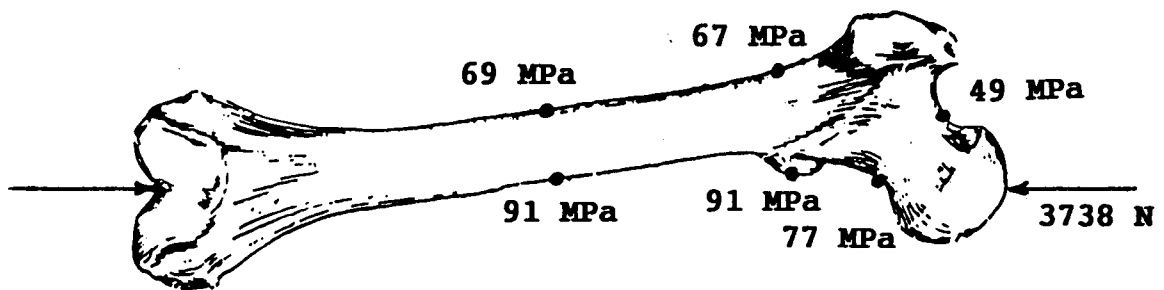
Figure 47. Analysis of low strain rate axial loading of femur 724R. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

BONE: 881R

AGE: 81

SEX: FEMALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: SHAFT OR TROCHANTER

ACTUAL FRACTURE:

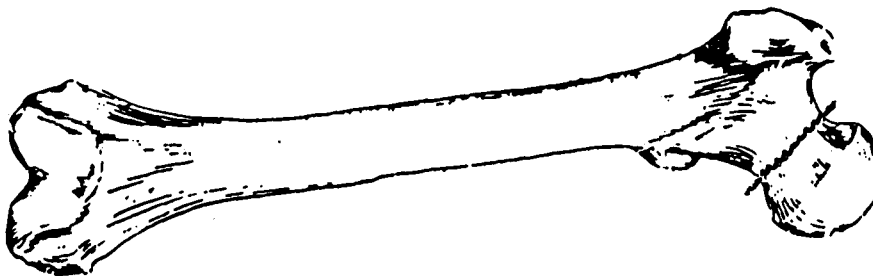


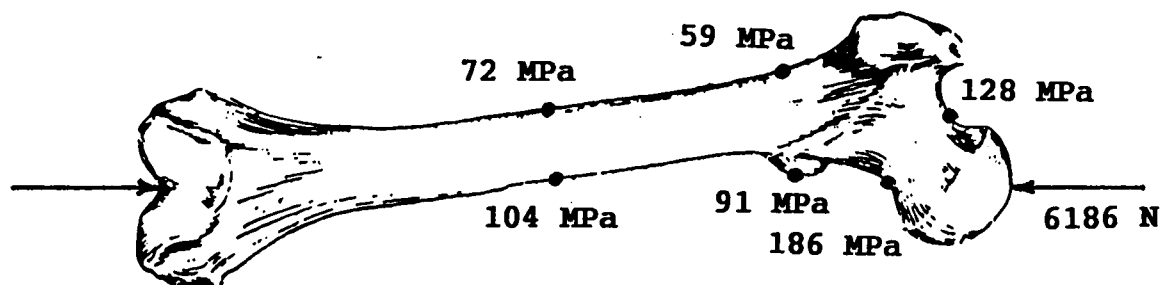
Figure 48. Analysis of low strain rate axial loading of femur 881R. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

BONE: 880R

AGE: 71

SEX: FEMALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: NECK FRACTURE

ACTUAL FRACTURE:

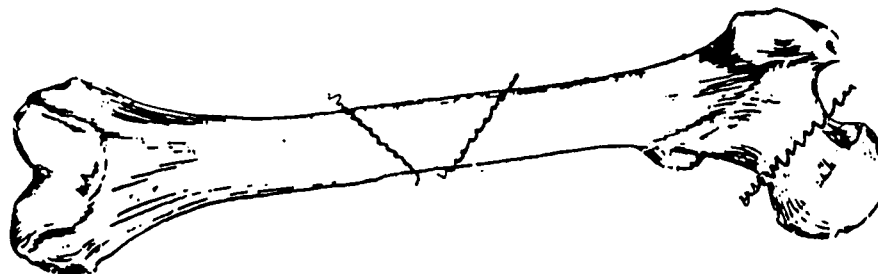


Figure 49. Analysis of low strain rate axial loading of femur 880R. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

Table 25

Breaking Load and Ultimate Stress
for Slowly Applied Axial Load Test

Bone	Breaking Force (Newtons)	Ult. Compr. Stress at Fract. Area (MPa)	Ult. Tens. Stress at Fract. Area (MPa)
859R	10578	114	78
902R	4050	91	55
812R	1847	40	28
883R	6252	176	102
862R	6141	152	88
865R	5051	157	101
724R	3627	86	54
881R	3738	---	---
880R	<u>6186</u>	<u>186</u>	<u>128</u>
AVERAGE	5274	125	79

2. High-speed axial loads (knee impact)

No other high speed axial impacts of the bare femur have been located in the literature. Nine bones have been struck in this matter; six provided good force data.

Figures 50-55 map out the stresses on these bones as in the previous section. In this case, all predictions were correct. The only minor exception is a concomitant shaft fracture in bone 862L. These concomitant fractures are common clinically (Chapman, 1984). Table 26 summarizes the results. Once again, the compressive stresses are 1.5 times greater than the tensile.

Young's modulus cannot be calculated from these tests. The deflection in the bone is unattainable since the impactor slows down considerably through impact.

Table 27 presents the results of both the static and impact tests and compares them to results found in the literature. In order to compare impact strength, strain rate must be determined. The strain rate is equal to

$\dot{\gamma} = \text{impactor velocity} / \text{specimen length}$
(McElhaney, 1966). The strain rate for these tests is

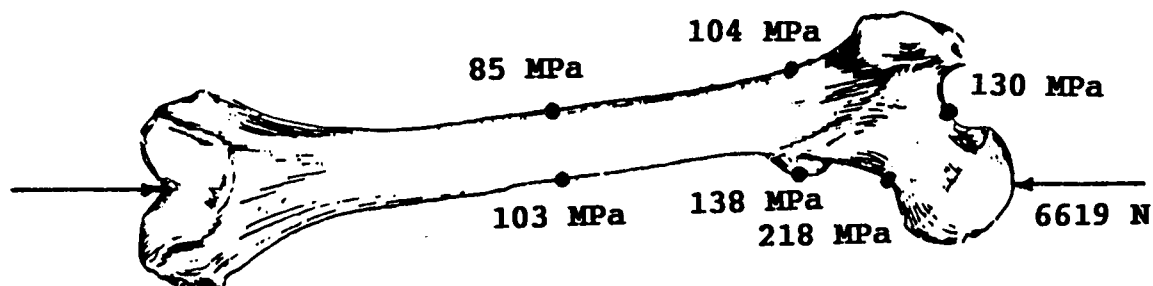
$$\dot{\gamma} = \frac{7 \text{ (m/s)}}{0.4 \text{ (m)}} = 17.5 \left(\frac{1}{\text{sec}} \right)$$

BONE: 862L

AGE: 63

SEX: MALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: NECK FRACTURE

ACTUAL FRACTURE:

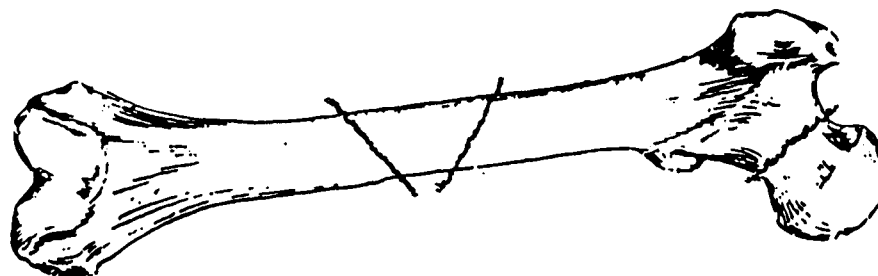


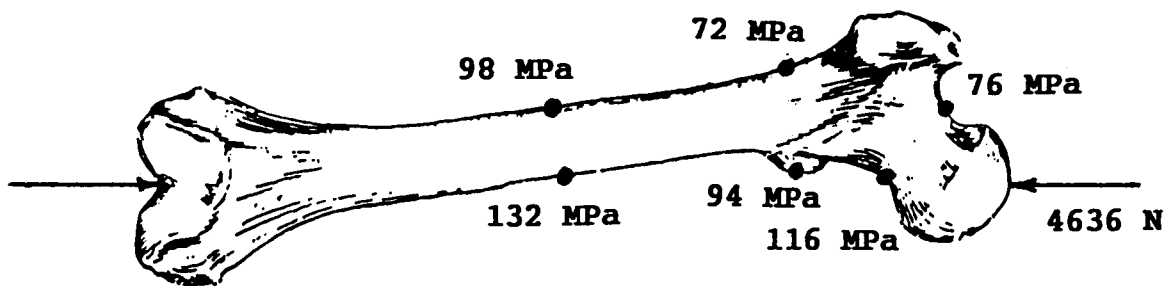
Figure 50. Analysis of impact loading of femur 862L. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

BONE: 867L

AGE: 32

SEX: FEMALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: SHAFT FRACTURE

ACTUAL FRACTURE:

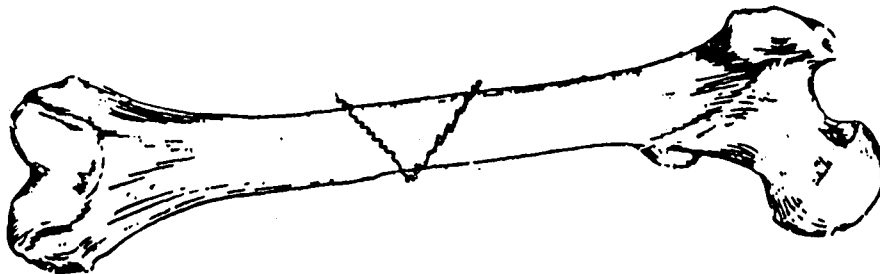


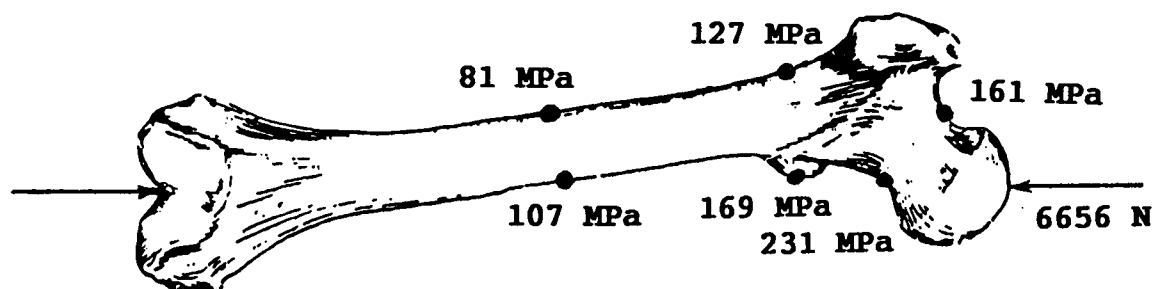
Figure 51. Analysis of impact loading of femur 867L. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

BONE: 865L

AGE: 63

SEX: FEMALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: NECK FRACTURE

ACTUAL FRACTURE:

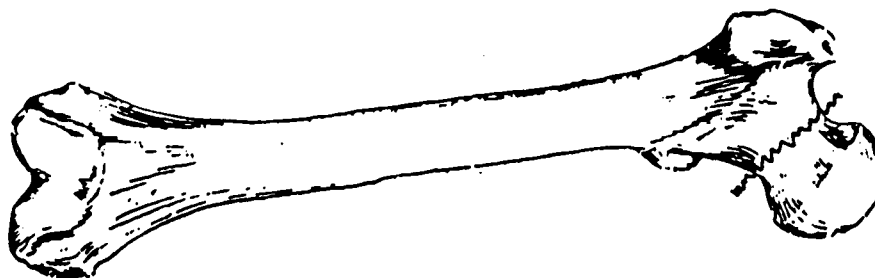


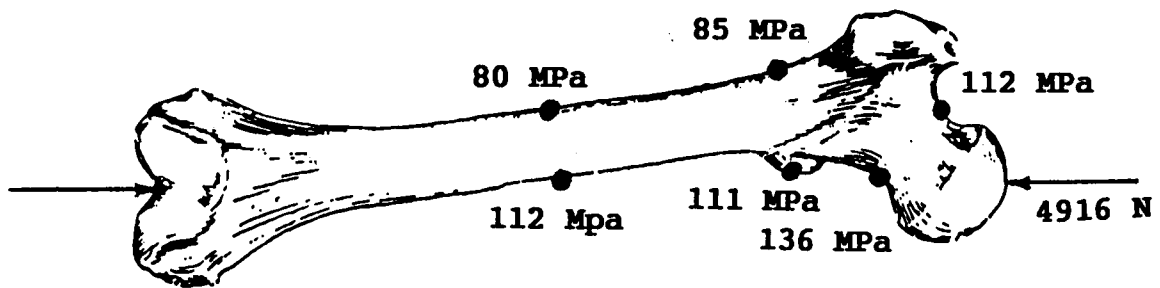
Figure 52. Analysis of impact loading of femur 865L. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

BONE: 881L

AGE: 81

SEX: FEMALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: NECK FRACTURE

ACTUAL FRACTURE:

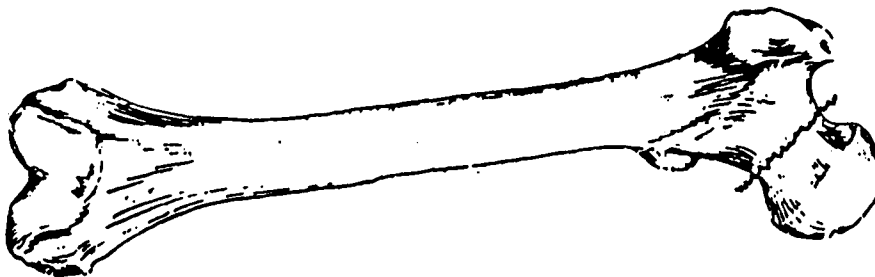


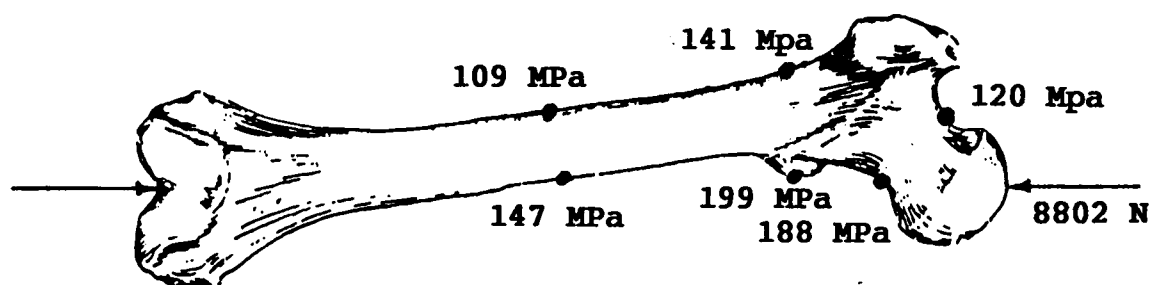
Figure 53. Analysis of impact loading of femur 881L. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

BONE: 859L

AGE: 52

SEX: MALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: SUB-TROCHANTERIC FRACTURE

ACTUAL FRACTURE:

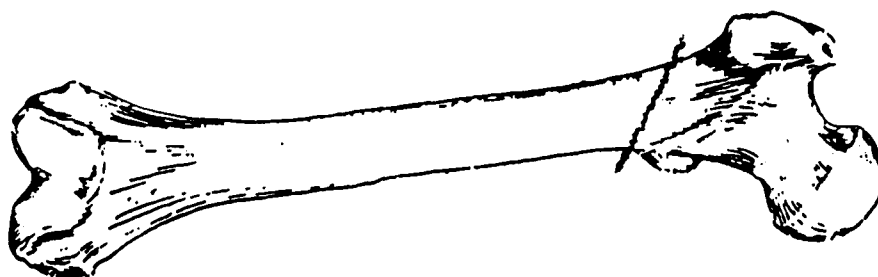


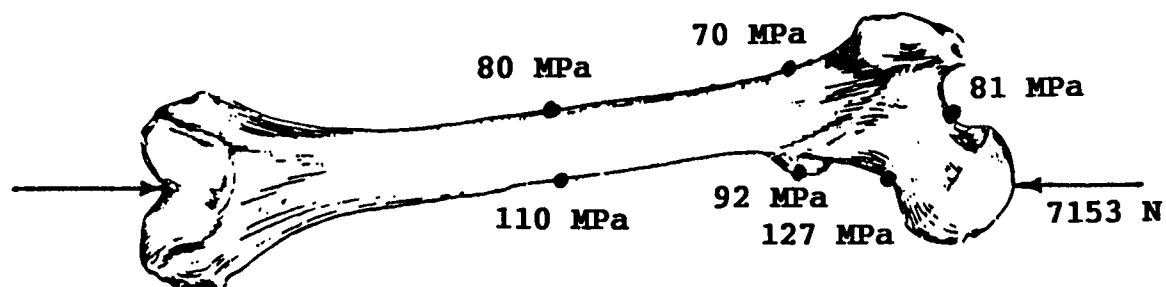
Figure 54. Analysis of impact loading of femur 859L. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

BONE: 888

AGE: 71

SEX: MALE

STRESS DISTRIBUTION:



PREDICTED FAILURE: NECK FRACTURE

ACTUAL FRACTURE:

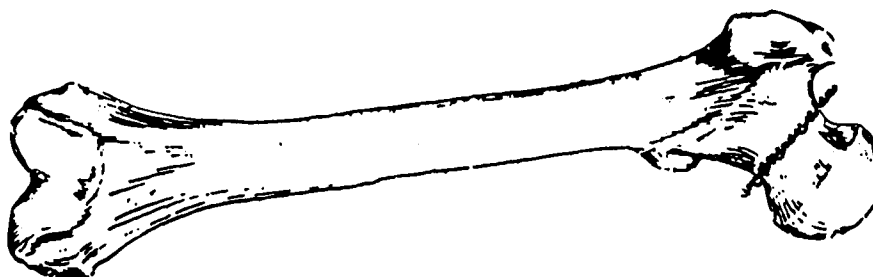


Figure 55. Analysis of impact loading of femur 888L. Top figure shows stresses at three cross-sections (lateral stresses are tensile and medial are compressive). Bottom figure shows actual fracture location.

Table 26

Breaking Load and Ultimate Stress
for Impact Axial Load Test

Bone	Breaking Force (Newtons)	Ult. Compr. Stress at Fract. Area (MPa)	Ult. Tens. Stress at Fract. Area (MPa)
AXF862L	6619	218	130
AXF867L	4636	132	98
AXF865L	6656	231	161
AXF881L	4916	136	112
AXF859L	8802	199	141
AXF888L	<u>7153</u>	<u>127</u>	<u>81</u>
AVERAGE	6464	174	121

Table 27

Comparison of Results from Axial
Test with Those of Previous Researchers

Property	Value
Low Strain Rate Breaking Strength	
Tucker	125 MPa
Yamada, 1971 (fresh)	130 MPa
Impact Breaking Strength	
Tucker	174 MPa
McElhaney, 1966 (fresh)	221 MPa

D. TEST IV: FRESH TISSUE TESTING

Fourteen cryogenically-frozen, fresh long bones were broken to help understand the effects of embalming on the properties of the bone.

Table 28 presents the results of these tests. Table 29 compares these results with embalmed data. Breaking force values appear to increase 44 percent for fresh femurs and 78 percent for fresh tibias. Energy absorption also increases for fresh bones. There is, however, too little data to make sound conclusions.

Table 28

Results from A-P Impact of Fresh Long
Bones of Two Individuals

Cadaver, bone		Breaking Force (N)
105		
	Femur (right)	9482
	Femur (left)	10017
	Tibia (right)	6620
	Tibia (left)	5542
	Fibula (right)	930
	Fibula (left)	772
	Humerus (right)	5285
	Humerus (left)	4469
98		
	Femur (right)	7228
	Femur (left)	6065
	Tibia (right)	4988
	Tibia (left)	4313
	Fibula (right)	1129
	Fibula (left)	895

Table 29

Comparison of Fresh Bone Results and
Embalmed Bone Results

Bone, property	Embalmed	Fresh	%
Femur			
Breaking Force (N)	5697	8198	43.9%
Area Under Curve (N ms)	2657.7	4778	79.8%
Tibia ¹			
Breaking Force (N)	3022	5366	77.6%
Area Under Curve (N ms)	--	--	

¹Embalmed tibia data from Kress, 1989

CHAPTER VI

CONCLUSIONS AND RECOMMENDATIONS

This thesis has presented data on the impact response of the human femur under several loading conditions. The data presented is important, because of the role it can play in the quantification of the damage to hard and soft tissue under loading conditions similar to those which occur when an automobile impacts a human leg. This provides insight into protection mechanisms and input for computer and physical models. Conclusions and recommendations for each test follow.

A: TEST 1: BREAKING STRENGTH OF FEMUR

From the data, it is apparent that the femur is stronger and stiffer when impacted in the anterior-posterior direction than it is in the lateral-medial direction. Bone is non-homogeneous, anisotropic and has properties that vary according to location on the bone (Evans, 1951). This directional change in properties, therefore, should not be surprising.

Bone remodels itself so that it is stronger in areas encountering greater stress. Since normal body functions (walking, running, etc.) put a moment on the femur similar

to three-point loading in the a-p direction, this strength increase in the a-p direction should actually be expected.

Viewing the plots of age vs. strength and age vs. stiffness, there appears to be very little relation between the parameters. While it is acknowledged that 20-year-old bones on the average would be stronger than 80-year-old bones, no such statement can be made for the age span shown in the plots.

Mather (1968) showed that the left and right femurs absorbed the same amounts of energy when impacted. It was further shown in this thesis that left and right matched pairs have essentially equal properties (with only a few exceptions). This finding adds validity to all past experiments involving matched pairs of bones and allows this technique to be used for future testing.

A study of the fractures resulting in this bending impact is of interest. The lateral-medial impacts produced wedges that occurred on the lateral and medial sides. The anterior-posterior impacts also produced wedges occurring on the lateral and medial sides. Kress (1989) stated that a vast majority of clinically seen femur impacts occur in the l-m direction. With these additional findings, however, it is possible that actual a-p impacts may appear clinically as l-m impacts and be misinterpreted.

Bending impact tests need to be performed on the intact thigh. Now that the bare femur response is understood, the role of soft tissue and the role of the pelvis can be examined.

B. TEST II: TORSIONAL STRENGTH OF FEMUR

The reason behind performing the low strain rate torsional test was to develop the ultimate torque-cortical thickness relationship. This relationship then allowed the "percentage of maximum torque" calculations to be made for the torsional preload.

The tests clearly showed that even a small torsional preload reduced the breaking strength of the femur significantly. It also showed that a small torsional load (when compared to the breaking load) can produce spinal fractures.

When combined load stress calculations are made, the additional stress added by the torsional preload is not enough to account for the decrease in the breaking load for the preloaded bone. Further investigation needs to be made into the exact mechanism behind this action.

C. TEST III: COMPRESSIVE STRENGTH OF FEMUR

This section title is a misnomer. When one thinks of axially loading a specimen, one certainly expects most of

the stresses to be compressive. The axial experiments on the femur, however, showed that tensile stresses could actually be the fracture initiators.

Viewing the 15 figures mapping out the stress on the bone and showing the fracture location, it is obvious that bone geometry determines fracture location. In 14 out of 15 tests, the cross-section under the greatest calculated stress was the fracture site.

A majority of the fractures (80 percent) occurred at the neck of the femur. This number appears high when compared to clinical studies where 50 percent of automobile accident victims (Daffner et al., 1988) and 17 percent of motorcycle victims (Deaner et al., 1975) with broken femurs have neck fractures. But when you consider age of the bones studied (the average age was 64.9 years), 80 percent neck fractures is not surprising. Osteoporosis attacks the femur in a disproportionate manner. The neck of the femur tends to lose bone at a higher rate than the rest of the femur (Hofeldt, 1987). As evidence of this, the three youngest bones tested broke at the shaft.

Fung (1984) reported a compressive strength in bone that is approximately 1.5 times greater than its tensile strength. Viewing the figures of mapped out stresses on the femur, it can be seen that the compressive stress on the medial side of the femur is always approximately 1.5 times

the tensile stress on the lateral side of the femur.

The impact breaking strength was 39 percent greater than the breaking strength at very slow rates of load application. This change is significant and is almost exactly the change predicted by McElhaney (1966). This indicates that high-speed impact tests are necessary in studying the automobile-motorcycle collision.

The average breaking load for the slowly loaded bone was 5274 Newtons and for the impacted bone was 6464 Newtons. Present automobile design regulations dictate that a force of 10,000 Newtons may not be exceeded when a knee impacts a dashboard at 6.6 meters per second (Krishnaswamy, 1991). Results from this research indicate this level might be high. Only one femur had a breaking tolerance higher than 10,000 Newtons.

A great deal more tests need to be performed in this area. Since a preponderance of neck fractures occur when using femurs of the elderly, young bones need to be tested. This is not as important in the other tests (l-m, a-p, torsion), since the bone geometry can be factored out using simple stress calculations. In these axial tests, however, the neck of old bones fracture before the shaft fracture threshold can be measured.

Also of interest would be whole leg testing. The role soft tissue plays would become apparent as would soft tissue

damage caused by the bone fracture. It would also determine if the pelvis alters the type of fracture that occurs, or if the pelvis itself sometimes fractures.

D. TEST IV: FRESH TISSUE TESTING

Kress (1989) concluded that there was essentially no difference in breaking strength of fresh and embalmed bones used in impact studies. Kress, however, used a small sample size of goat bones in forming this conclusion.

Findings from this research, while still based on results from a small sample size, indicate that there is a significant change in fresh and embalmed properties. The fresh human femurs were 43.9 percent stronger than the embalmed femurs and absorbed 79.8 percent more energy. The difference in breaking strength for the tibias was even greater.

This conclusion must be viewed carefully. Only 14 fresh bones were tested. Also, these fresh bones came from individuals younger than the average embalmed bone donor.

More fresh tissue testing needs to be performed, and anatomical measurements need to be taken on the fresh tissue tested. This will help determine the exact difference in fresh and embalmed tissue.

With all things considered, embalmed bones are the best choice for impact experiments. The embalmed bones are

easier to handle, and they are safer in that potential diseases are destroyed. It is also best to use embalmed bones, because the results in terms of the force required to break the bones are conservative.

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VITA

On February 27, 1965, an artist and naturalist was born into a golfer's body. He was named Guy Virgil Tucker like his father and grandfather. His parents and three older sisters spoiled Guy rotten, and he has lived a happy and reflective quarter century.

Guy has yodeled from the highest mountain tops, bathed in the quietest streams, hiked through the driest desert canyons, shushed down the steepest slopes and stared at the slickest birdie putts. Through all of these experiences, Guy has decided on his place on this planet. Kurt Vonnegut best sums up this role when speaking of Guy:

The world is full of people who are very clever at seeming much smarter than they really are. They dazzle us with facts and quotations and foreign words and so on, whereas the truth is that they know almost nothing of use in life as it is really lived. [Guy's] purpose is to detect such people - so that society can be protected from them, and so they can be protected from themselves.

Guy obtained a Bachelor of Science degree from the University of Tennessee in 1987. He has attended graduate schools at the University of Utah, in the beautiful state of Utah, and the University of Tennessee, where he is receiving this Master of Science degree. He is now heading to Washington, D.C. to offer the U.S. Patent Office his

services.

Guy considers the two greatest discoveries of man to be the jalapeno pepper and the pinot noir grape, and he considers the two worst to be the designated hitter and stadium golf.

When Guy reflects on this completed thesis with its 29 tables and nearly 500 calculations, he again agrees with Kurt Vonnegut, who says, "Just because some of us can read and write and do a little math, that doesn't mean we deserve to conquer the Universe."