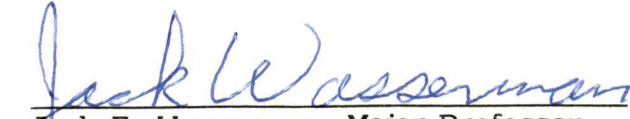


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**A THREE-DIMENSIONAL, NONLINEAR, TRANSIENT, DYNAMIC,
FINITE ELEMENT ANALYSIS OF THE
EMBALMED HUMAN TIBIA SUBJECTED TO
HIGH SPEED IMPACT LOADING**

A Thesis

Presented for the

Master of Science

Degree

The University of Tennessee, Knoxville

Michael B. Cooper

May 1993

DEDICATION

This thesis is dedicated to my parents, **Stanley** and **Susan Cooper**. Without their invaluable guidance, endless patience, and constant support and love throughout my undergraduate and graduate career, I would have never achieved this milestone.

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ABSTRACT

A relatively new topic in the biomedical field is the area of impact biomechanics, which involves the study of trauma to the human body due to impact conditions. The research in this area is primarily experimental, and little finite element analysis has been performed. The research presented in this thesis deals with the impact biomechanics area; the purpose of this thesis is to apply the finite element method to accurately model a 24.5 ft./sec. impact at the tibial midshaft in the anterior to posterior direction. It is through the finite element modeling that a better understanding of impact trauma to the human lower leg can be attained.

A three-dimensional, nonlinear, transient, dynamic, solid mechanics, finite element code--PRONTO 3D-- is used to analyze the impact condition. The model will investigate fracture propagation and the incurred fracture force in the tibia.

In an attempt to verify the finite element analysis, experimental testing matching the computational model conditions was performed on eleven dry, embalmed, human tibias. Fracture force data and fracture propagation trends were investigated in these tests.

Results show that the finite element model agrees with the general trends shown experimentally for both fracture force and fracture propagation in the tibia. However, comparison between the experimental and computational models were not entirely possible, and only general trends could be noticed. With the development of an accurate constitutive model of the tibia and better experimental verification, the finite element method can prove to be a valuable tool in injury prediction and the design of injury mitigating devices.

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Chapter I

Introduction

A. Project Overview

Over the past seven years at the University of Tennessee, the Japanese Automobile Manufacturers Association (JAMA) has provided funding for research in the area of high-speed impact of human, animal, and artificial leg bones. This research involves an in-depth study concerning the mechanisms of failure in leg bones. The study has been accomplished by quantifying the damage to hard and soft tissues that undergo loading conditions similar to an automobile impact upon a human leg. The study has given insight into the design and development of protective mitigating devices and will provide input data for computer and physical models.

During the first three years of the JAMA project the primary concerns were the development and subsequent testing of an impact facility. The consequent creation of a pneumatically operated machine, specifically developed for impacting human and animal intact leg and bare bones, allows the accurate simulation of automobile collisions with pedestrians or cyclists. This machine consists of a pressure tank that accelerates a cart down a guide-rail system at velocity ranges between 3.0 and 13.4 meters per second (9.8 to 44.0 ft/sec) with essentially constant velocity throughout the impact. The cart is instrumented at the impact bumper with a force transducer that directly measures the force of the impact. The cart can accommodate a variety of impact bumpers and other impacting surfaces as well.

After development of the test facility, cadavers were tested to

provide insight into the mechanisms of failure of the lower-leg bones during impact. Also, a large amount of time was spent developing force measurement capabilities; from these measurements, a force-versus-time history could be developed for bone fractures. These measurements provided insight into the design of breakable limbs for impact test dummies. To obtain these measurements, efforts were concentrated on the breaking strength of human tibias and intact cadaver legs, as well as on the differences in preserved and living tissues.

In addition to analyzing the breaking strength of the tibias and cadaver legs, efforts continued in the investigation on the characteristics of bone material properties and geometric considerations. The effect on the overall breaking strength of bone was also investigated. During the first three years of research, results indicated that the development of an artificial breakable bone was a real possibility.

The first Master's thesis on this topic, written during this third year of research, is entitled Mechanical Behavior of Lower Limbs in Response to Impact Loading: Facility Development and Initial Results by Tyler Kress (1989). Kress's thesis focuses on the response of the human tibia under impact conditions and the initial testing of the new impacting machine.

The fourth and fifth year research continued efforts in the characterization of the breaking strength and mechanisms of failure of the tibia and intact cadaver legs. The research also attempted to analyze the characteristics of the breaking and bending strength of the human femur as it underwent high speed impact conditions. Both torsional loads and axial loads in the anterior-to-posterior and lateral-to-medial directions were applied in this study. Fresh tissue testing was initiated to determine the

effects of embalming on the bone properties. At the end of this fifth year of research a second Master's thesis was written, Guy Tucker's The Mechanical Behavior of the Human Femur Subjected to Impact Loading (1991). Tucker's thesis deals primarily with the mechanisms of failure of the human femur undergoing different load orientations. Both Kress's and Tucker's documents are cited often in this thesis.

The sixth and seventh year research revolved primarily around the development of a computer simulation of the experimental testing; this computer simulation and subsequent analysis forms the subject matter of this thesis.

B. Research Objective

One of the long-term goals of the JAMA project is to develop impact/injury computer models that can be used to guide designs for a variety of personal protective equipment; these models would also aid in designs for automotive mitigating devices for injury reduction. This thesis analyzes the initial results of an application of a dynamic impact load to a three-dimensional, nonlinear, finite element model of a human tibia (PRONTO3D--developed by Sandia National Laboratory). This thesis also covers the experimental verification of the finite element model to ensure accuracy of results. The study follows a simple three step process:

- I. Experimental testing**
- II. Initial Finite Element Model development and analysis**
- III. Final Finite Element analysis and experimental testing comparisons.**

Chapter II

Literature Review

A. Background

Over the three year period between 1985 and 1988, 32.78 million musculoskeletal injuries occurred annually: 6.155 million annually were fractures of bones with 492,000 of these annual fractures occurring to the tibia, fibula, and ankle (Praemer et al., 1992). Praemer et al state that of these fracture injuries, 66.5% resulted in patient activity restriction and/or bed disability. According to the General Estimates System, in 1988 alone approximately 6.9 million police-reported motor vehicle accidents occurred; 187,000 of these accidents involved pedestrians or cyclists. Tibia fractures were involved in 22.6% of these events (Praemer et. al, 1991).

Research conducted by Cesari et al. (1988), Kress (1989), and Tucker (1991) points to the fact that the severity of leg injuries in pedestrian/auto accidents is directly proportional to impact speed and bumper height; by examining these two factors engineers can develop devices that diminish injury severity by absorption of the impact energy.

A device that adequately protects motorcycle occupants' or pedestrians' legs is not currently available (Kress, 1989). Although research conducted by Kress (1989) and Tucker (1991) analyzed the dynamic behavior of the leg in response to impact loading, further research is needed in this area--particularly in the area of computational prediction of injury based upon impact-loading scenarios. A few of the areas studied are the computational modeling of the human lower leg, impact biomechanics, and mechanics of materials (hard tissue).

B. Impact Biomechanics

The majority of previous research involving automobile accidents and biomechanics focused on injuries of the head and neck. Research has emphasized the use of anthropomorphic dummies as predictors for human body dynamics in these accidents, as well as computer code modeling of human body dynamics in impact simulations: i.e. the Articulated Total Body (ATB) code developed by Wright Patterson Air Force Base. The ATB code models airplane pilots during ejection from jet cockpits as well as the human body dynamics of a motor vehicle occupant during impact simulation.

There is a vast amount of literature on material properties of human and animal tissue. Yamada (1973) published much of the early material property determination; however, most of this previous work slow-loaded the tissues and therefore does not accurately define dynamic material properties (Kress, 1989). McElhaney (1966) is one of the few researchers to investigate dynamic material properties by impact loading animal tissues. Still this research shows some inadequacy: only small specimens of tissues were used instead of a whole bone/leg sample, and large changes occurred in velocity of the striking object at impact failure. Therefore McElhaney's results do not accurately model impact conditions of automotive/human accident interaction.

Table 1 provides a historical perspective of pioneering research on impact biomechanics (Kress, 1989).

C. Bone Anatomy

Bone is an anisotropic composite material that consequently exhibits different properties at different locations on the bone. As shown in figure 1 and figure 2, Fung (1981) and Park (1984) describe the typical long bone

Table 1**Historical Perspective of Some of the Pioneering
Research of Impact Biomechanics**

Date	Scientist	Contribution
1940	Hugh De Haven	Provided first insights into human tolerance of crash loads; established the Automotive Crash Injury Research Program at Cornell University
1941 1946	Sir Hugh Cairns	Helmet studies (Army and motorcycle)
1942	De Haven	Published a paper establishing the groundwork for whole-body tolerance
1942	John Lane (Australia)	Noted that aircraft should be certified in two ways -- airworthy and crash-worthy (this gave birth to the term "crash-worthiness")
1945 1951	Gurdjian Stapp	Cadaveric studies Volunteer studies
1954	Mathewson and Severy	Dummy testing

Table 1 (continued)

Date	Scientist	Contribution
By mid 1950's	Various contributors	A body of knowledge existed that gave insights into general frequencies of traffic collisions and injuries, some understanding of actual mechanisms which generate injuries, and a means whereby forces and accelerations applied to car occupants could be modified
1960	Sheldon	Paper entitled, "On the Natural History of Falls in Old Age"
1962	Aldman (Sweden)	Anatomical positioning for seatbelt analysis complimented with dynamic property data
1962	Snyder	Tibia testing (4448 N)
1962-1963	Ford Motor Company Garrett Huelk Gissane and Bull	Examined the real world of collisions
1966	William Haddon, Director	Establishment of the National Highway Safety Bureau by an Act of Congress
1966	Young	Cadaver leg testing (6543 N)

Table 1 (continued)

Date	Scientist	Contribution
1966	Goldsmith	Stress-strain behavior during impact
1967	Frank and Mather	Tibias of younger people and males were more resistant to fracture than tibias of older people and females; 75 N-m (using 4.22 m/s falling weight broke the tibia in 50% of tests)
1969	Mackey	Collision aspects of road accidents
1973	Kramer, Burow & Heger	Tested 209 cadavers (most > 60 years of age) for "bumper type" tibial impact velocity of 6.7 m/s produced fracture in 50% of tests
1960's 1970's 1980's	Various contributors	Researchers produced voluminous literature on various human kinematics under impact conditions; very important information regarding the composition and mechanical properties of bone was also provided

Source: Tyler A. Kress, "Mechanical Behavior of Lower Limbs in Response to Impact Loading: Facility Development and Initial Results." Master's Thesis in Engineering Science and Mechanics, The University of Tennessee, Knoxville, August 1989, pp.7-9.

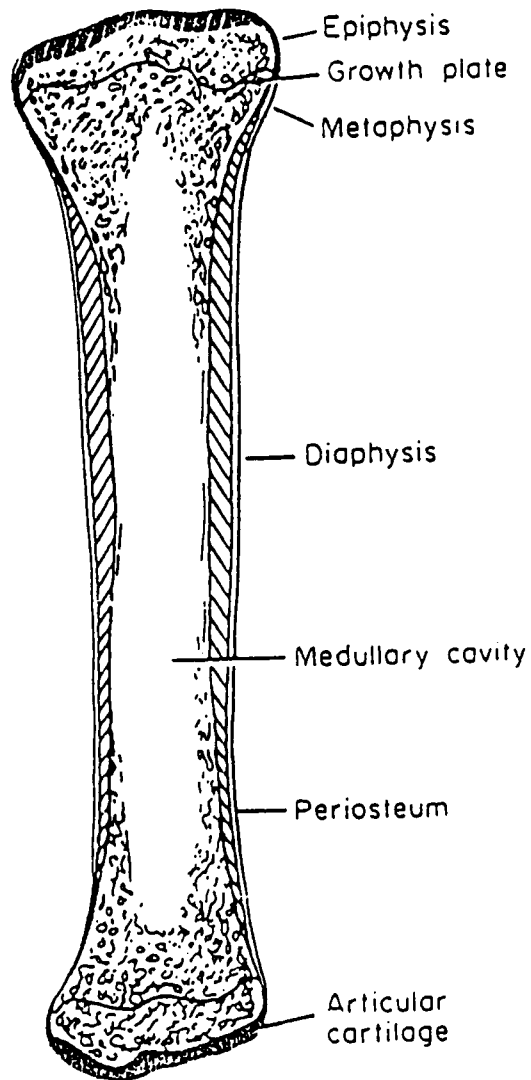


Figure 1. The parts of a long bone

Source: Y.C. Fung, Biomechanics Mechanical Properties of Living Tissues. Springer-Verlag New York Inc., 1981.

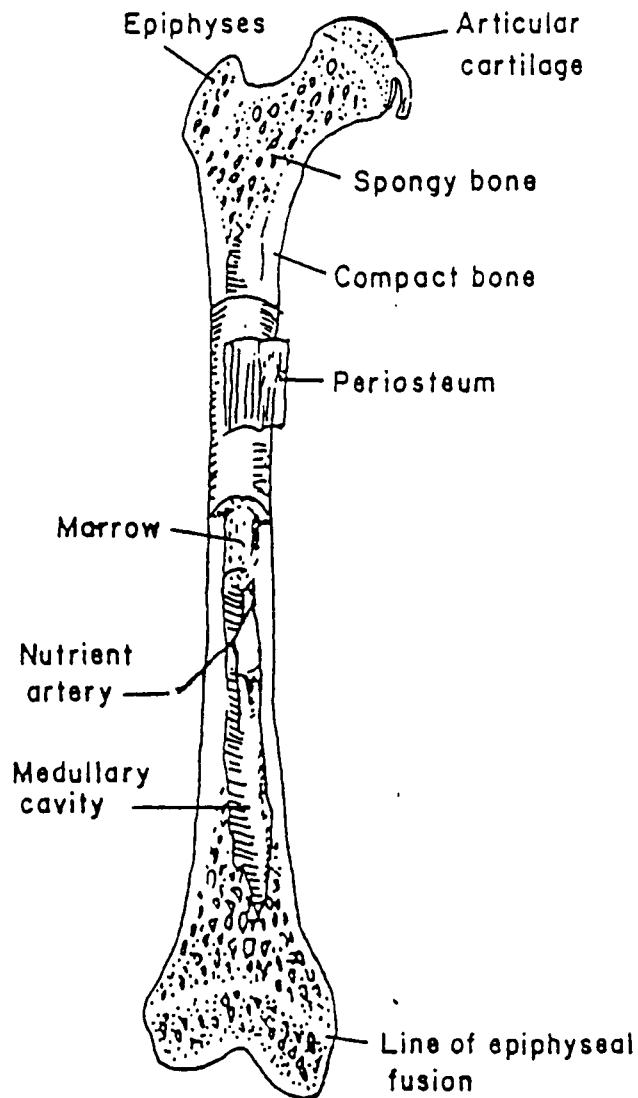


Figure 2. Organization of a typical bone

Source: J. B. Park, Biomaterials Science and Engineering. Plenum Press, New York, 1984.

as consisting of a shaft (diaphysis) with an expansion at each end (metaphysis). The diaphysis' geometric construction is of a hollow tube; the diaphysis' walls consist of a dense cortex (compactum). The compactum is thick throughout the length of the diaphysis, tapering off to become the thin shell of the metaphysis. The hollow area of the diaphysis, which contains the bone marrow, is the medullary cavity. Covering the entire external surface of the long bone, except the articulation, is the periosteum. Inside the inner layer of the periosteum dwell the highly active cells that produce circumferential enlargement and remodeling of the growing long bone; this inner layer is the osteogenic layer. The outer layer of the periosteum is fibrous and in mature long bones comprises most of the periosteum.

A simple sketched representation of compact (cortical) bone (figure 3) at the microscopic level shows that bone material is a composite. The representation of this basic microscopic unit of cortical bone is known as the Haversian system or osteon. Similarly Volkmann's canals are transverse channels connecting an artery or vein visible in the center of the osteon (Fung, 1981).

According to Fung (1981), about two-thirds of the weight of bone--half its volume--is inorganic material that appears to be composed of hydroxyapatite with traces of other ions. Organic material composes the rest of the volume of bone--mainly collagen. Hydroxyapatite crystals are arranged along the length of the collagen fibrils and groups of these collagen fibrils run parallel to each other to form fibers. The arrangement of fibers is different in different parts of bone. The hydroxyapatite, collagen and different fiber arrangements combine to make bone a composite

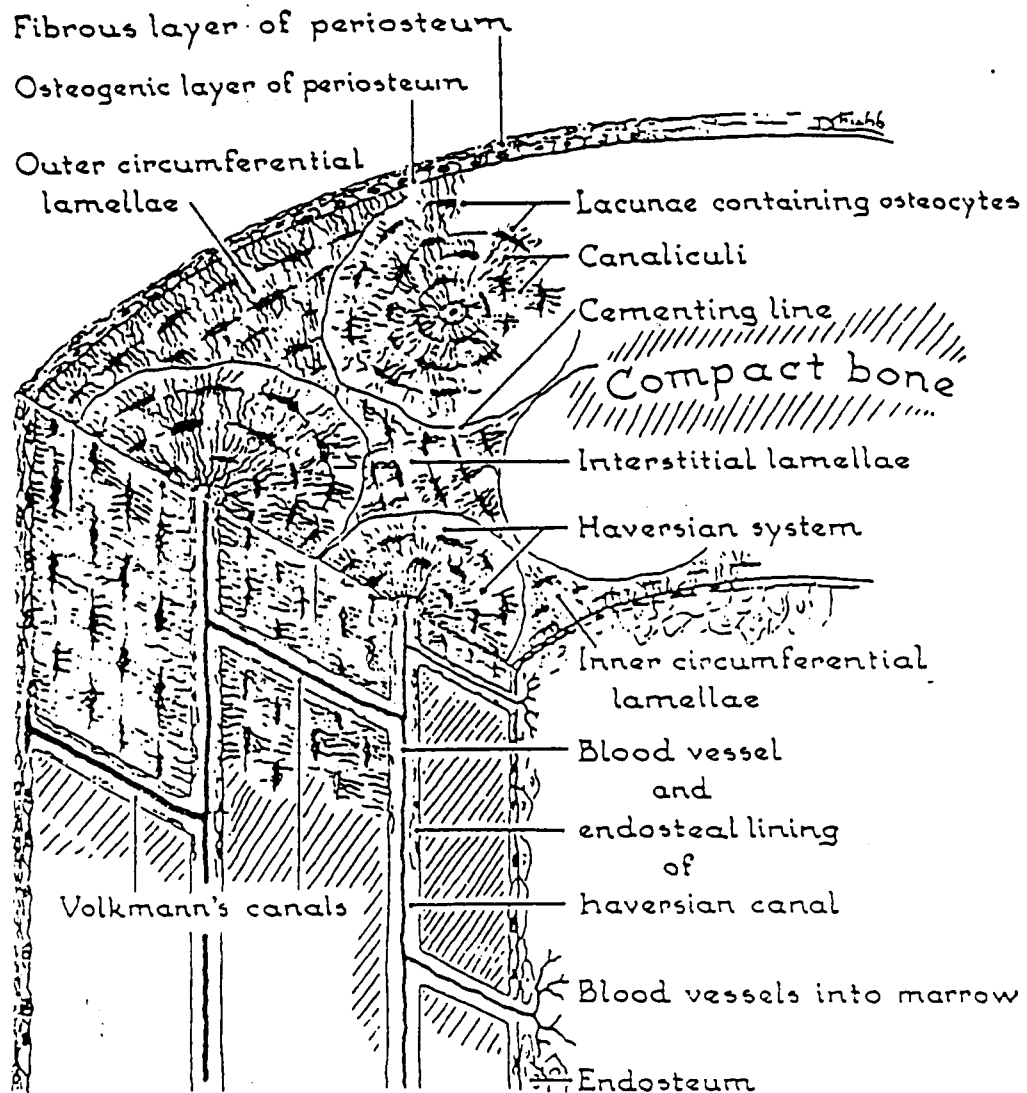


Figure 3. The basic structure of compact bone

Source: Fung, 1981.

material.

Due to the composite nature of bone, its strength is greater than that of either hydroxyapatite or collagen. The softer component (collagen) prevents the stiff component (hydroxyapatite) from brittle cracking, while the stiff component prevents the softer component from yielding. Young's modulus of bone is an intermediate value between that of hydroxyapatite and collagen.

Best and Taylor (1950) reported that water constitutes about 25% of bone's weight, thus increasing the energy absorbing capacity of living bone. Most traumatic fractures of bone arise from impacts that create energy that the bone must absorb. Therefore, this energy-absorbing characteristic is of great importance in traumatic fractures. Partly due to the large percentage of water in bone's weight, bones are able to absorb considerable amounts of energy and undergo marked deformations without breaking (Evans 1957). The distribution by weight of the constituents of whole cortical bone is shown in figure 4, which illustrates the proportion of cell protein in the organic material (Park, 1984).

D. Mechanical Properties of Bone

Because bone is an anisotropic material, its mechanical properties are directionally different. According to Fung (1981), factors affecting the mechanical properties of bone are

- * Age of bone
- * Sex of bone donor
- * Location of bone
- * Orientation of bone

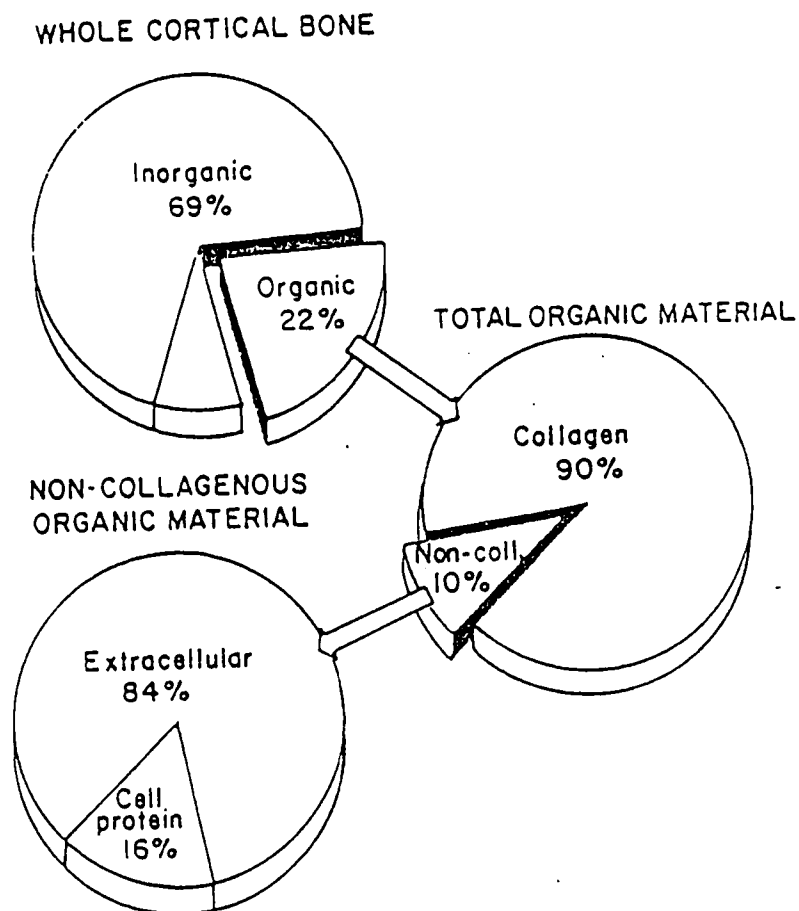


Figure 4. Distribution by weight of the constituents of whole cortical bone to illustrate the proportion of cell protein in the organic material

Source: Park, 1984.

* Strain rate of bone

* Test condition of bone

(Wet--freshly embalmed Dry--embalmed, air dried).

In the determination of the mechanical properties of bone, transverse isotropy (TI) or orthotropy (ORTH) are the two theoretical material symmetries used to describe bone (Cowin, 1989). TI is based on the assumption that bones have consistent properties throughout a cross-section; ORTH is based on the assumption that bones have differing properties throughout a cross-section. According to Cowin, the differences in the data are small, and the use of either model of material symmetries will have an insignificant effect on stress analysis of bone. However, the material symmetry will have an effect on the physiological behavior of bone tissue as a composite material.

The literature on the mechanical properties of bone is abundant and the experimental results among researchers are often contradictory; thus, basic preference determines a suitable set of mechanical properties for each researcher. The majority of the research involves mechanical properties of human and animal femurs. Rarely is a full intact bone used for testing purposes; instead, a small cube is normally used in mechanical property testing. Many researchers prefer using the cube test sample: i.e. Yamada (1970), Cowin (1989), Evans (1957, 1970), and Fung (1989), Reilly and Burstein, Yoon and Katz, Knets et al, McElhaney, Wong et al (1983), and Ashman. Of these researchers, only McElhaney determined mechanical properties under dynamic impact conditions. Unfortunately, only small square samples of femurs were used instead of the full intact femurs that

might have given more accurate results. The small cube samples are more widely used due to its ease of use in material testing; however, the mechanical properties of a full intact bone are much different than that found in a small cube section of bone. This is due to the anisotropic nature of bone. Wong is the only researcher to use full intact tibia specimens for dynamic impact loading to determine impact response; but unfortunately Wong did not attempt to find specific mechanical property values.

To accurately describe the tibia bone, the following mechanical properties are required for accurate bone modelling:

- * Elastic Modulus
- * Density
- * Poisson's Ratio
- * Yield Stress.

These mechanical property values taken from a variety of previous research will assume orthotropy material symmetry when possible. The properties are determined in the transverse or radial direction with respect to the long axis of the bone. The comparison study will focus on both dry and wet bone data.

Density is an important variable in the representation of a bone model. Density's importance is due to the fact that the behavior of a body subjected to a force is a direct result of the body's quantity of material, the body's mechanical properties, the body's size, and the body's distribution of mass. Therefore, a denser body will behave much differently under a given load than will a less dense body--even if the external geometries are identical (Evans, 1973). Evans (1973) found that strong correlations exist

between density and the following properties: tensile modulus, ultimate tensile strength, Rockwell hardness, and single (punching) shear strength. The density of human bone is the same as the density of cortical or compact bone and is found to be 0.065 lbm/in^3 (1.8 g/cm^3) for dry and wet bone (Skalak and Chien, 1987).

The stress-strain curve (figure 5) for dry bone in tension is essentially a straight line to failure; however, the curve for wet bone shows a deviation from the straight line at the yield stress to the failure point (Park, 1984). The differences in the stress-strain curves suggest that the material property characteristics depend upon the condition of the bone; for instance, dry bone exhibits a linear-elastic material characteristic with an absence of any plastic deformation, while wet bone exhibits both elasticity characteristics as well as plastic deformation. Therefore, careful attention must be paid to material property characteristics in the modeling of human bones.

The stress-strain curve for compact bone is normally used instead of the full bone cross-section of compact and cancellous bone. Because the compact bone is much stronger than the cancellous bone, the tests of compact bone are thought to result in the most accurate data for the entire bone cross-section. In 1948, Evans and Lissner tested full intact bones to obtain stress-strain data; unfortunately the tests used femurs instead of tibias--this data is therefore not applicable (Evans, 1957). Evans (1957, 1973) stated that the average ultimate tensile stress of wet, human tibia is 15,954 psi (110 MPa) and dry tibia is 18,665 psi (128 MPa). Evans also stated that the average yield stress for wet, human tibia is 13,053 psi

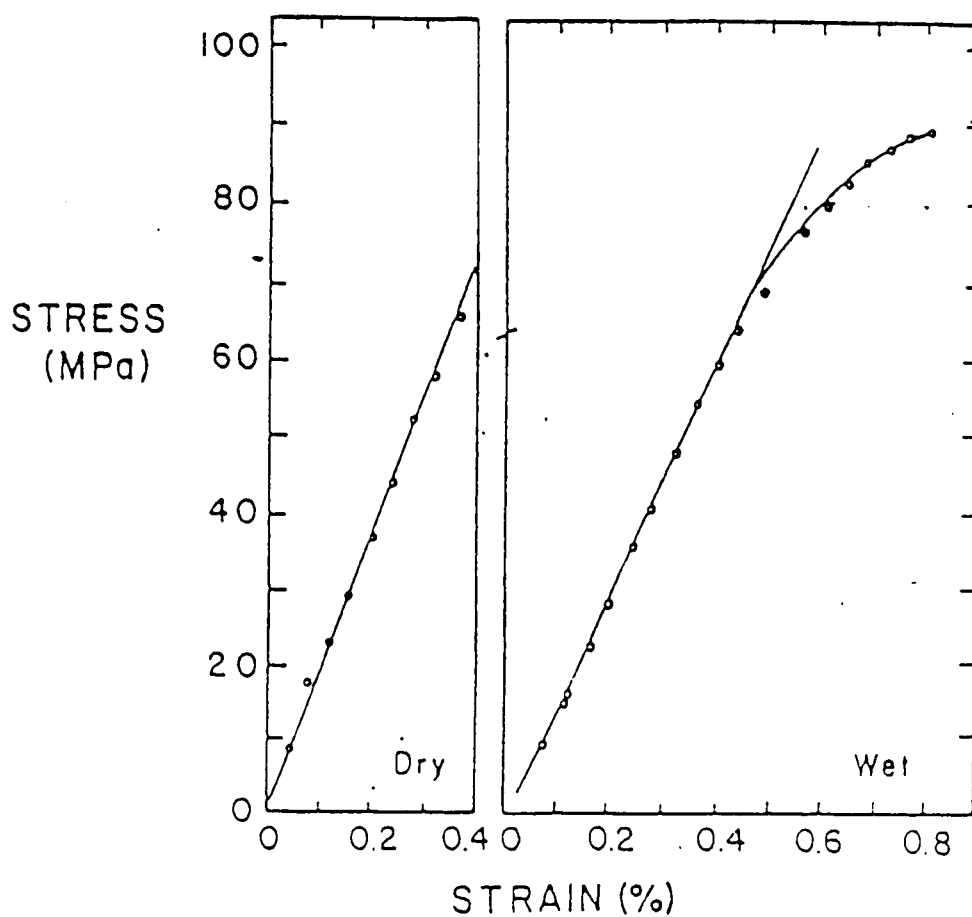


Figure 5. Effect of drying on the behavior of embalmed human compact bone

Source: Park, 1984.

(90 MPa) and for dry tibia is 18,665 psi (128 MPa) (Evans, 1957 and 1973).

According to laboratory data obtained from Kress (1989) and Tucker (1991), the fracture of human bone occurs on the tension side parallel to the long axis when undergoing anterior to posterior dynamic impact loading. Therefore, the value for Young's modulus in tension of a wet, human tibia is -2.67×10^6 psi (18.4 GPa)--instead of the value in the radial direction (Fung, 1981). Evans (1973) reports that Young's modulus increases by 18% for dry bone; thus, based on the value for wet bone, Young's modulus of a dry, human tibia is 3.15×10^6 psi (21.71 GPa).

A material loaded in one direction will undergo strains perpendicular to the direction of the load as well as parallel. The measurement of the lateral (or perpendicular) strain to the longitudinal (or axial) strain for a uniaxial state of stress is known as Poisson's ratio (Higdon et al., 1985). Human bone is no different in this respect than any other material. According to Cowin (1987), Poisson's ratio for wet or dry tibia bone in the radial direction is 0.3.

Further mechanical properties can be found for each component of the human bone (compact bone, cancellous bone, etc.), and extensive documentation for all of these component properties exists by such researchers as Fung, Evans, and Yamada. However, the preceding discussion summarizes the mechanical properties of whole, intact human tibias in sufficient detail for use in bone modeling. Table 2 summarizes the mechanical properties of both wet and dry embalmed human tibia bone.

Table 2
Mechanical Properties of Embalmed Tibia Bone

Mechanical Property	Wet Bone	Dry Bone
Young's Modulus	2.67E+6 Psi 18.4 GPa	3.15E+6 Psi 21.71 GPa
Ultimate Stress	15,954 Psi 110 MPa	18,665 Psi 128 MPa
Yield Stress	13,053 Psi 90 MPa	18,665 Psi 128 MPa
Poisson's Ratio	0.3	0.3
*Density	0.065 lbm/in ³ 1.8 g/cm ³	0.065 lbm/in ³ 1.8 g/cm ³

* Value for embalmed compact bone

E. Finite Element Analysis in Biomechanics

The finite element method (FEM) is a numerical method for solving problems of engineering and mathematical physics. Typical problem areas in engineering and mathematical physics that are solvable by the FEM include, stress analysis, heat transfer, fluid flow, mass transport, and electromagnetic potential (logan, 1986). The FEM utilizes computer technology to quickly solve problems that often involve complex geometries. In 1972, FEM was introduced for use in the biomedical community to evaluate stresses in human bones; since then it has been used almost exclusively for orthopaedic applications. These applications include such areas as stress-related architecture of bone and the bone remodeling process, artificial joint design and fracture fixation devices, and mechanical behavior of tissues such as artificial cartilage and intervertebral discs. (Huiskes and Chao, 1983). From 1972 to 1982, the greater part of the efforts in FEM of biomechanics were method-oriented rather than problem-oriented.

Most of the finite element work described in the early literature involves analysis of the femur due to its common involvement in orthopaedic applications; the majority of the FEM work today is still highly centralized in femur analysis. Huiskes and Chao report that the first three-dimensional FEM analysis was of the femur and was conducted by Scholten (1975). This analysis was followed by similar three-dimensional finite element models by Olofsson (1976), Valliappan et al (1977), Harris et al. (1978), and Rohlmann et al.

Research into bone failure mechanisms is extremely important

because these mechanisms govern orthopaedic problems. The FEM serves as a useful analytic tool to predict highly stressed regions. Due to problems associated with data manipulation and interpretation/representation of the three-dimensional FEM, the two-dimensional approach to FEM is a more popular choice in orthopaedic biomechanics (when possible). According to Huiskes and Chao, Vichnin et al. (1977) carried out the first FEM simulation of a bone fracture test. This test started a string of research projects using FEM for fracture analysis of the bone (Rybicki-1974, Rybicki and Simonen-1977, Simon et al-1977, Levine and Stoneking-1980, Claes et al-1982, and Carter et al-1981).

Artificial joint design and fixation is the most popular application of the FEM in orthopaedic biomechanics. The FEM in this area emphasizes mechanical failure design problems of implants in order to extend the implant's functional life; mechanical optimization plays an important role in this area. Some of the mechanical failure mechanisms analyzed are plastic deformation and fracture of metal and plastic components, breakdown of acrylic cement, loosening at boundaries between different materials, and stress-related bone resorption (Huiskes and Chao, 1983). The first and most common finite element analysis of artificial joint design and fixation is that of the total hip replacement (THR).

Although not as common, the bulk of the finite element analyses of soft tissues in the musculoskeletal system involves the intervertebral discs and the articular cartilage; Huiskes and Chao report that Belytschko et al. (1974) conducted the first finite element model of this type. Belytschko's model featured linear and isotropic elastic properties, although it was later

modified to include nonlinear characteristics.

There has been little finite element modeling of the human tibia. Most of the work in this area involves analysis of the tibia as a base for artificial knee implants; the whole tibia has not been reported in finite element analysis research (Huisdes and Chao, 1983). Table 3 represents a short synopsis of work completed in the finite element analysis of the human tibia.

Model validity is one of the many problems with use of the FEM in the orthopaedic community. While much attention is paid to the anatomical mesh configuration, the visual agreement between the finite element model and reality unfortunately does not guarantee the validity of the result produced by that model. During the first decade of FEM's use in biomechanics, the question of whether the results and conclusions made justifiable sense, given the model assumptions, was not examined thoroughly (Huiskes and Chao, 1983). The validity of the model can be tested experimentally, although close attention must be paid to insure that the experimental verification closely matches the finite element analysis simulation. Accuracy is an important consideration of the finite element model. The most important factor concerning accuracy determination is the element mesh density in relation to the element type chosen. Convergency tests allow the accuracy of the mesh density to be determined; however, convergency tests for three-dimensional, nonlinear analysis can be expensive and time consuming (Huiskes and Chao, 1983). Aspect ratios and boundary conditions are also important areas to consider in the validity of a finite element model.

Table 3
History of the
Finite Element Analysis of the Human Tibia

Date	Contributor	Description
1976	Chand, Haug, Rim	2-D model of problems between femur and tibia in artificial knee prosthesis
1977	Various contributors	3-D model of tibial component
1977-1981	Various contributors	2-D model of tibial plateau fixation of nonhinged surface replacement prosthesis
1980	Pissinou, Brook	Model of contact stresses in femoral and tibial knee prosthesis
	Shrivastava	Axisymmetric model of tibial plateau fixation without central post
1979	Campan	Axisymmetric model of tibial fixation with 3-D loading
1982	Croon	
1980 1982	Roehrl, Scholten, Sollbach	3-D model of tibial and femoral stem fixation of hinged prosthesis
1981	Valenta, Komared, Tomcik	Simple 2-D model of tibia-femoral and femur-patella complex

Table 3 (continued)

Date	Contributor	Description
1981	Hayes, Snyder, Levine, Ramaswamy	Axisymmetric FEM analysis of upper tibia (non-axisymmetric loading)
1982	Croon	2-D modeling of fatigue behavior at cement-bone interface (tibial component in knee prosthesis with experimental verification)
1983	Murase	3-D loading on axisymmetric geometry
1985	Munzinger	Determination of optimal stem length of cemented tibial component of knee prosthesis
1986	Little	Established a control group for future modeling of physiological load and stress distribution in the proximal tibia
1992	Tissokht	Analyze meniscal response when joint is subjected to complex 3-D loading at tibiofemoral joint

Throughout the past decade most of the finite element analysis has been conducted using two-dimensional models; however, the opportunities for three-dimensional, nonlinear and dynamic analysis will greatly increase as computer power expands. The greatest challenge over the next decade will be to generate enough dependable data to put into the computer models. Precise formulations of material properties, boundary conditions, and loading characteristics are but a few of the data points needed for these finite element models.

Chapter III

Description of Experimental Testing Devices

The experimental aspect of this research was performed in the Biomechanics laboratory at the University of Tennessee. This laboratory was built in 1986 and has been used almost exclusively for the investigation of lower leg trauma due to impact loading. The main component of the laboratory consists of a device that simulates an automobile impact; this component will be referred to as the crash simulator. The crash simulator was the most frequently used tool in this experimental research.

The test setup will consist of a dry embalmed tibia specimen potted in woods metal filled steel cups at the proximal and distal ends. The specimen is hung by a rope from the proximal steel cup and impacted midshaft at 24.5 ft./sec. in the anterior to posterior direction by an impact cart. The impact cart is mounted with a flat plate bumper and instrumented with a force transducer. The tibia is instrumented with crack gages to determine crack propagation.

The crash simulator consists of a pneumatically-powered machine which simulates a car/motorcycle or car/pedestrian impact. A 300 pound impactor cart is propelled down a rail system to impact a test specimen (i.e. a human bone, a human leg, an animal bone, or an artificial bone). The impactor cart can accommodate a variety of different impact bumper configurations. Each bumper configuration is instrumented with a force measurement system that allows the user to obtain force data about the specimen.

The crash simulator comprises three principal parts: the accelerator and cart, the force measurement system, and the specimen holding device.

These components have been used for previous research conducted by Kress (1989) and Tucker (1991) and are explained in minute detail in their theses. Therefore, this thesis will only describe modifications made to the crash simulator or additional equipment used in the experimentation. The modifications made to the crash simulator involve the following: a new bumper design, and a new specimen holding device. In addition, a fracture investigation system was used for the first time in bone impact testing and will be discussed; this system consists of the crack gages and application materials, and the electronic fracture measurement circuit.

A. Impactor Cart Bumper Design

Instead of using the rigid pipe bumper that is normally used on the impactor cart, a steel flat-plate bumper was utilized in this research. The bumper consists of a 8 inch long piece of steel that is 1.5 inches tall with rounded edges; the bumper is 3/4 of an inch thick. Figure 6 shows the mechanical design of the bumper and figures 7 and 8 show the bumper mounted to the cart and instrumented with the customary force measurement system.

B. Specimen Holding Device

This test setup will be referred to as the hang test throughout the thesis. The hang test is used to simulate a pedestrian's leg undergoing impact as he/she lifts a foot off the ground to take a step (leg up phase); the setup can also simulate a pedestrian trying to jump up or out of the way of the oncoming impact. Both of these situations mimic real possibilities and the total specimen holding device accurately imitates the approximate weight of an average human leg (about 30 lbs.). A specimen holding device was developed to hold the test specimen in a constant position for impact

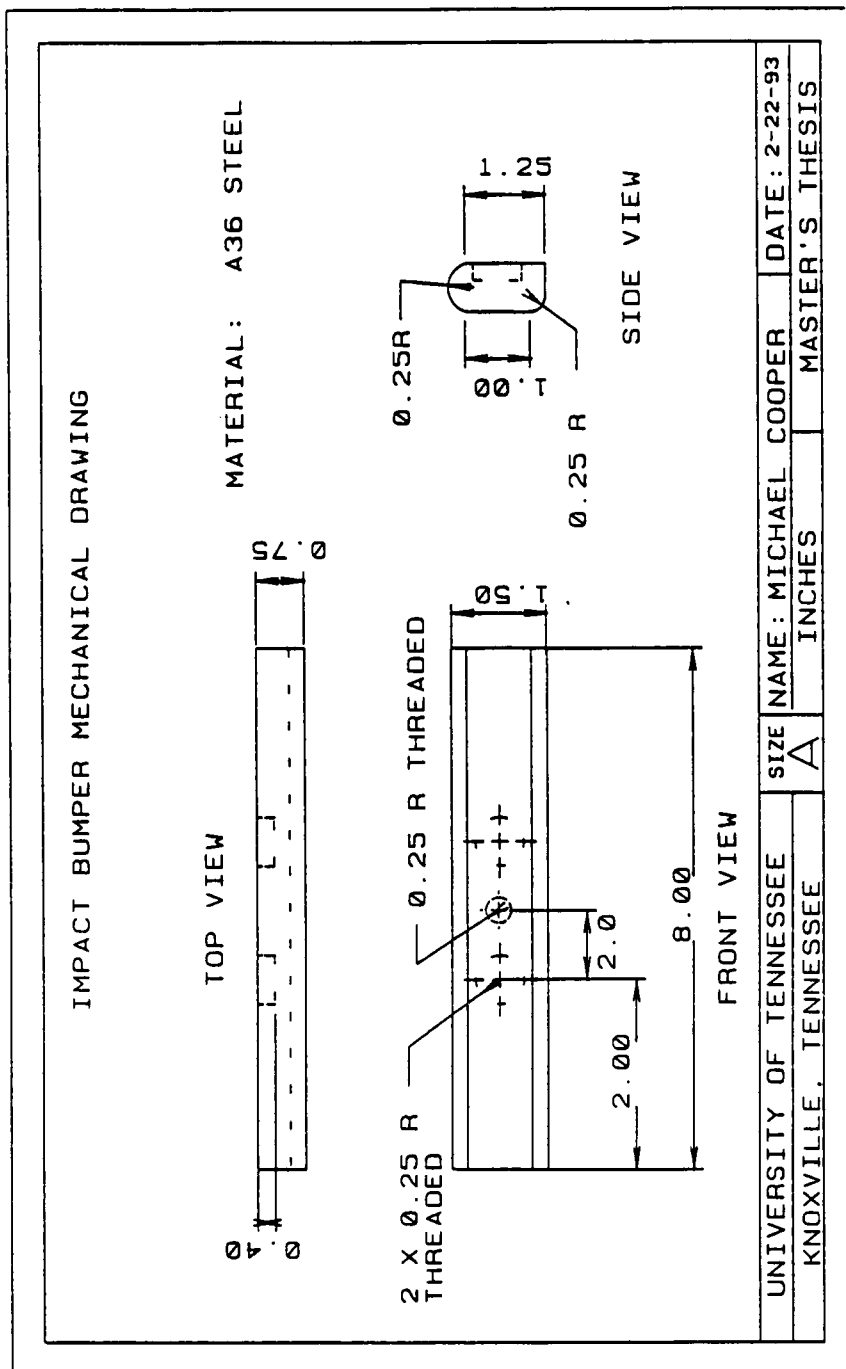


Figure 6. Mechanical design of the impact cart bumper

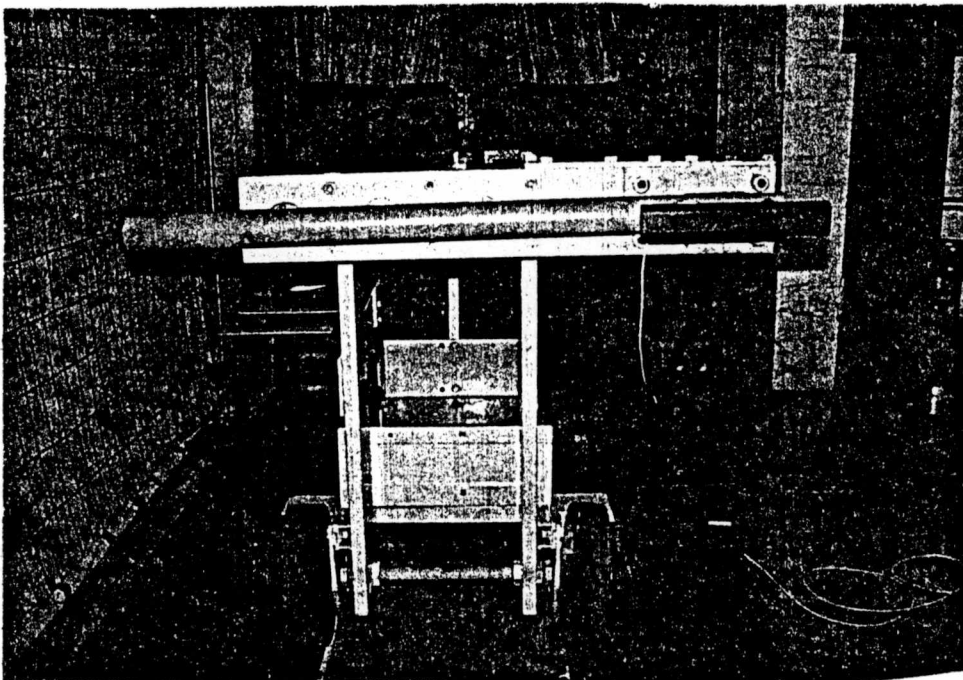
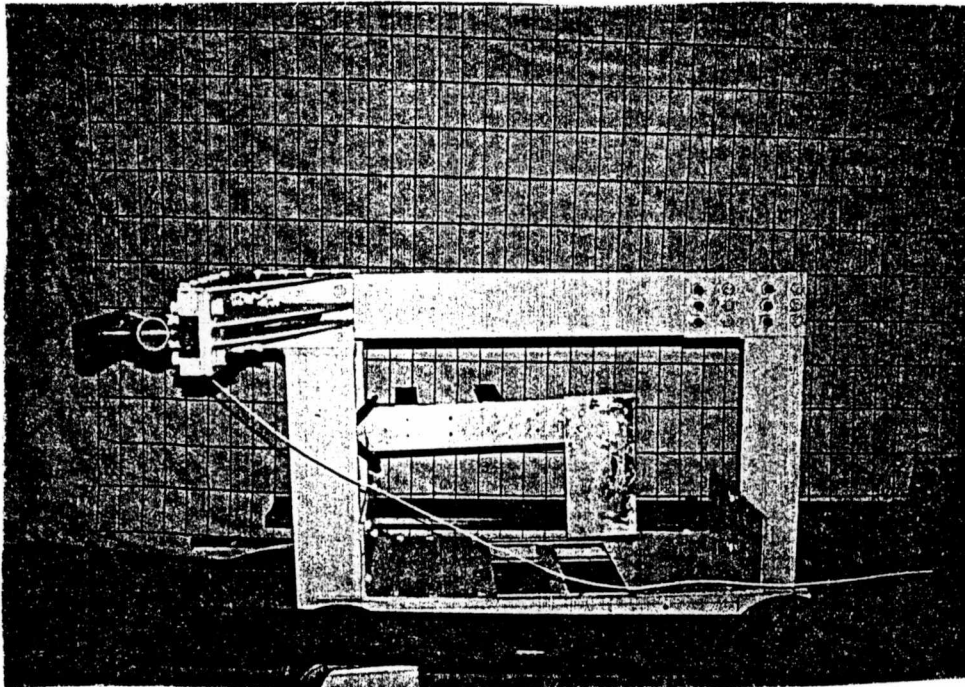


Figure 7. Overall view of impact cart mounted with impact bumper

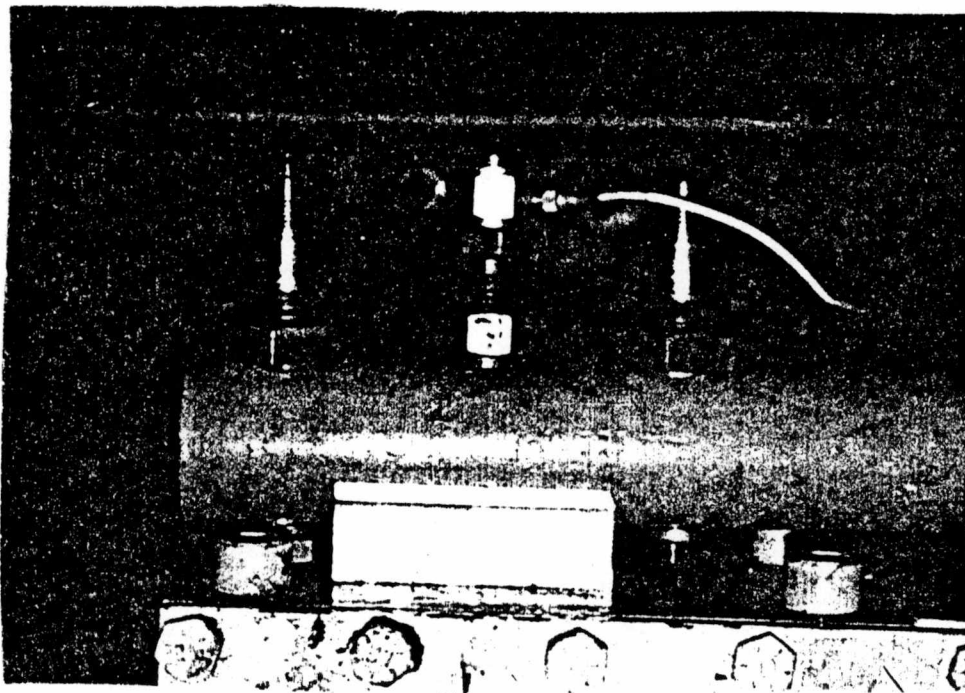
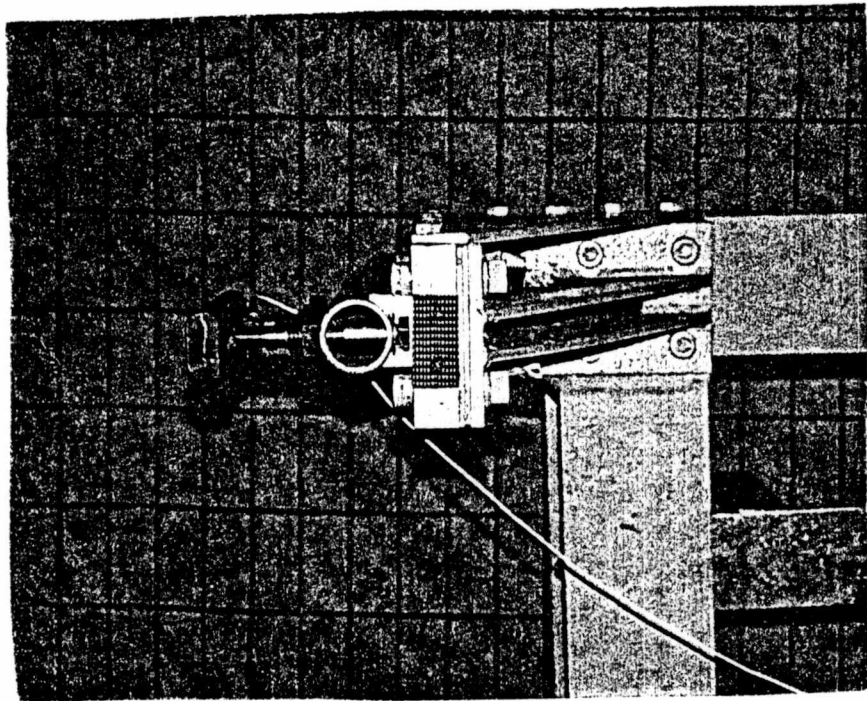


Figure 8. Detailed views of impact bumper with force transducer configuration mounted to the impact cart

testing. The specimen holding device was designed for its modeling simplicity in the finite element analysis while trying to simulate an actual impact situation. The holding device consists of two steel cups weighing 7 lbs. each. Figure 9 shows a mechanical drawing of these cups. The cups are filled with molten woods metal and the tibia is inserted into the woods metal at the proximal and distal ends of the bone. With woods metal, the proximal cup weighs 12 lbs and the distal cup weighs 15 lbs. Due to the nature of the woods metal, the metal will harden to a solid form and will hold the bone rigid in the cups. Woods metal was chosen as the fixating material because it was easy to melt for removal or insertion of the test specimen. Also, it returned to its solid state quickly therefore reducing the amount of "down time" between tests; in solid form, woods metal provided sufficient strength for tibia fixation. The proximal steel cup is fitted with an eye hook so that the bone can hang freely from a rope under the weight of the specimen holding device. The test specimen was hung from a rope in an attempt to simulate a human leg in the leg up phase of walking. Figure 10 shows the test setup of the bone hang test; notice the bone is allowed to move freely upon impact. By rotating the eye hook in the steel cup the bone can be oriented in a variety of positions. For this research the bone was oriented for an anterior to posterior midshaft impact. Midshaft of the bone is determined by measuring the bone and then dividing the length of the bone into thirds, positioning the bone with the impactor cart bumper at the middle third of the bone--midshaft. The position of the test rope allows the bone to impact the bumper at the center of the force transducer connection.

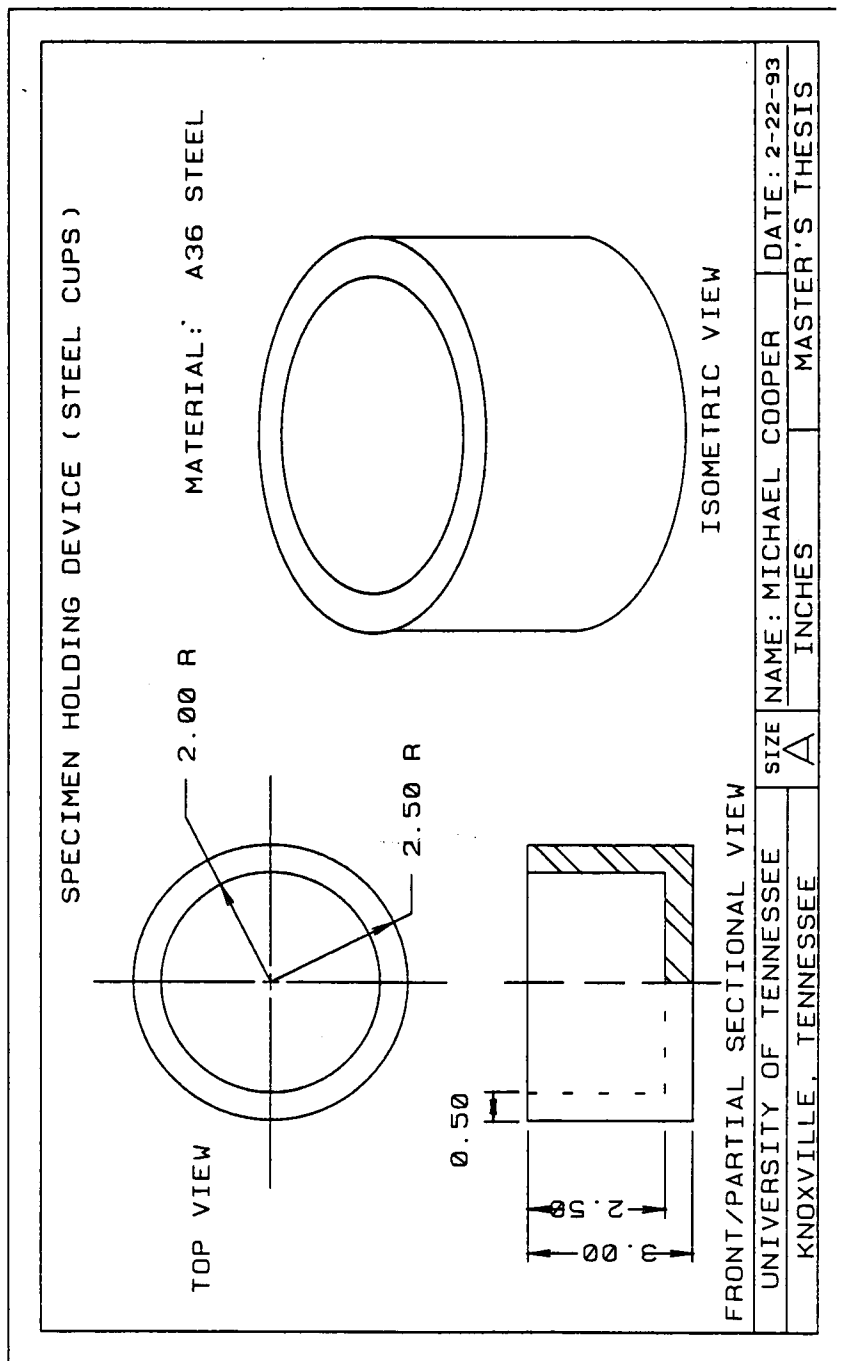


Figure 9. Mechanical design of steel cups--specimen holding device

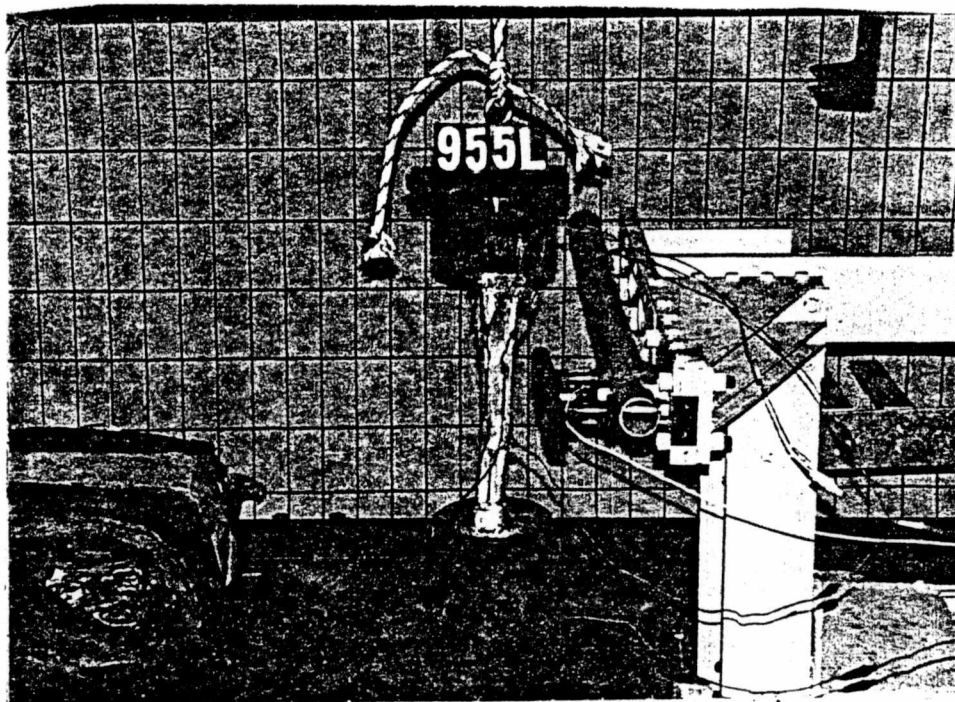
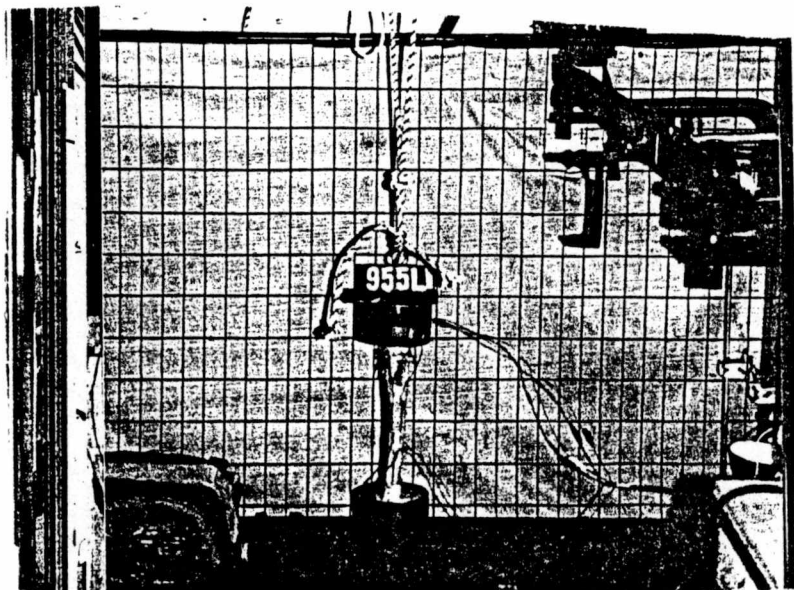


Figure 10. Views of the test setup for the tibia hang tests

C. Fracture Measurement System

1. Crack Gages

Crack gages act similar to an electric switch. When the switch is closed, a current runs through the gage; however, when the gage is open, the current through the gage is interrupted and a voltage drop occurs. The crack gages were used to detect fracture propagation in the tibia. Crack gages purchased from Measurements Group were mounted to the tibia with MBond 200 adhesive and catalyst. This was the suggested adhesive by Measurements Group for attachment of the gages to the tibia. The gages are 2 inches in length and consist of a beryllium-copper alloy on a tough polyamide backing. The gages carry a 2.3 ohm resistance and can be soldered into a electronic circuit to measure a voltage drop that indicates the gage is broken by the fracture. Figure 11 shows the crack gages.

2. Electronic Fracture Measurement Circuit

An electronic circuit was developed to measure the voltage drops in the crack gages as the fracture of the bone propagated through the gages. This circuit (figure 12) connects each of three resistors with a different resistance in parallel. Each resistor is wired to a different gage, and therefore when each gage breaks a different voltage drop will be observed. These three resistors in parallel are also connected in series with a shunt resistor. The voltage drop across this shunt resistor can be measured with the Hewlett Packard 3562A Dynamic Signal Analyzer by connecting the circuit to the analyzer through a BNC to alligator clip cable; the analyzer can then plot the voltage drops as a function of time to generate a voltage versus time plot. These voltage drops may differ between tests, therefore, it was necessary to measure the voltage drop readings before every test.

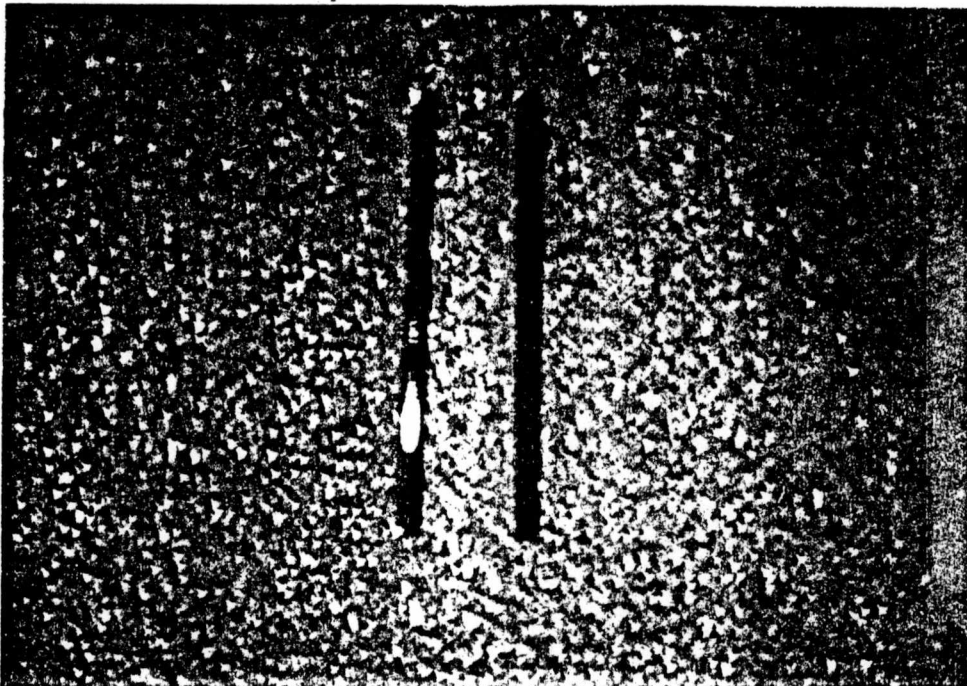


Figure 11. Crack gages used to determine crack propagation through the tibia

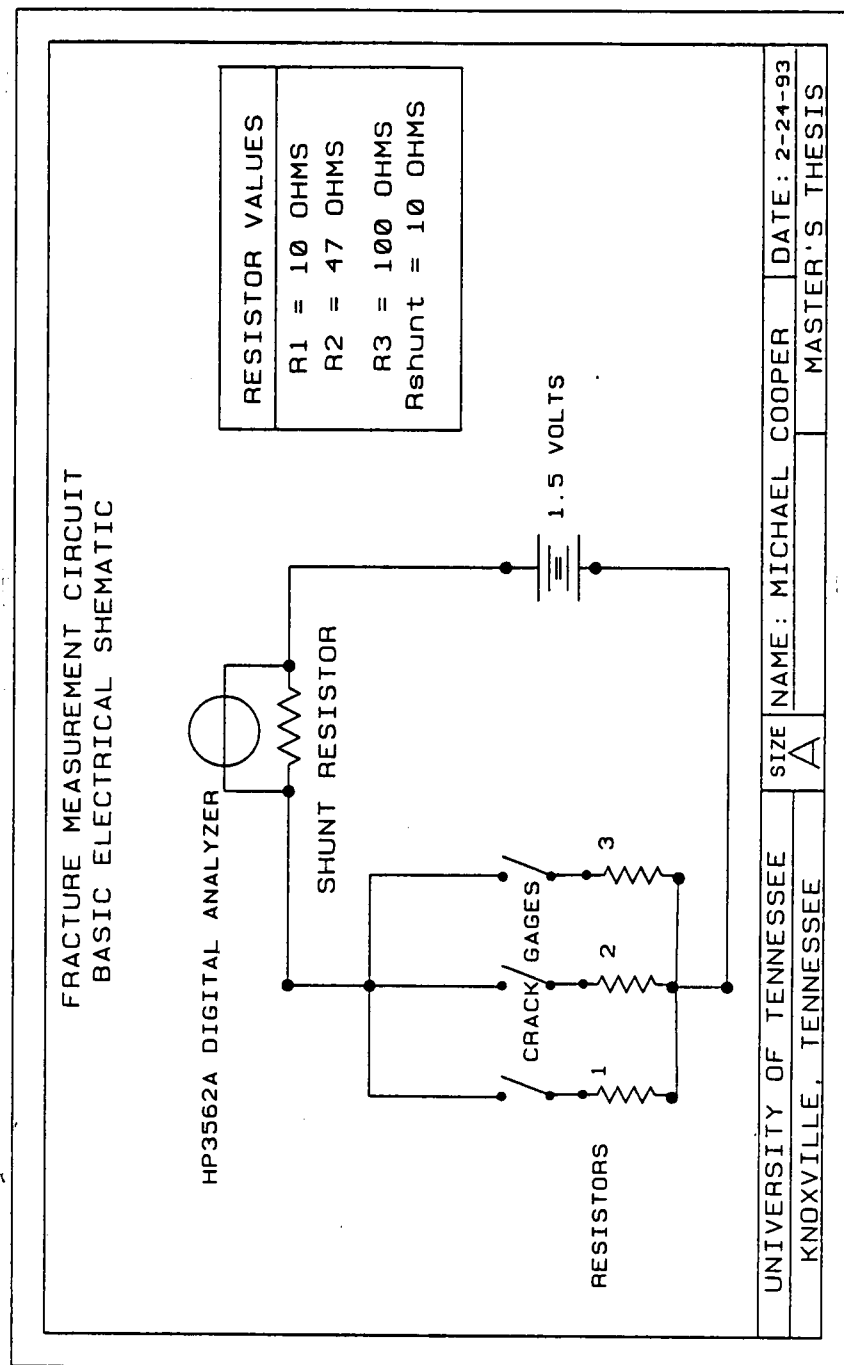


Figure 12 . Electrical schematic of the fracture measurement system (fmc)

This was accomplished by connecting and disconnecting the gages from the fmc and measuring the voltage drops with the signal analyzer. The analyzer starts measuring the voltage through the circuit as the analyzer is triggered by the impact bumper's force transducer impacting the test specimen. Figure 13 shows an example plot taken from the electronic circuit and figure 14 shows the circuit plot along with a corresponding force versus time plot. Notice that the two plots correspond in time, thus allowing the force data to be compared to the crack propagation through the gages.

D. Overall Experimental Setup

The analyzer allows two channels of data to be taken simultaneously. This experimental setup utilizes this capability. Figure 15 shows a block diagram of the test setup. The fracture measurement circuit is connected to channel two of the analyzer and is powered by a 1.5 Volt AA battery. Channel two is triggered to take data from channel one of the analyzer which is connected to the force transducer on the bumper and is powered by an eight channel power supply. The setup state of the digital analyzer is shown in figure 16.

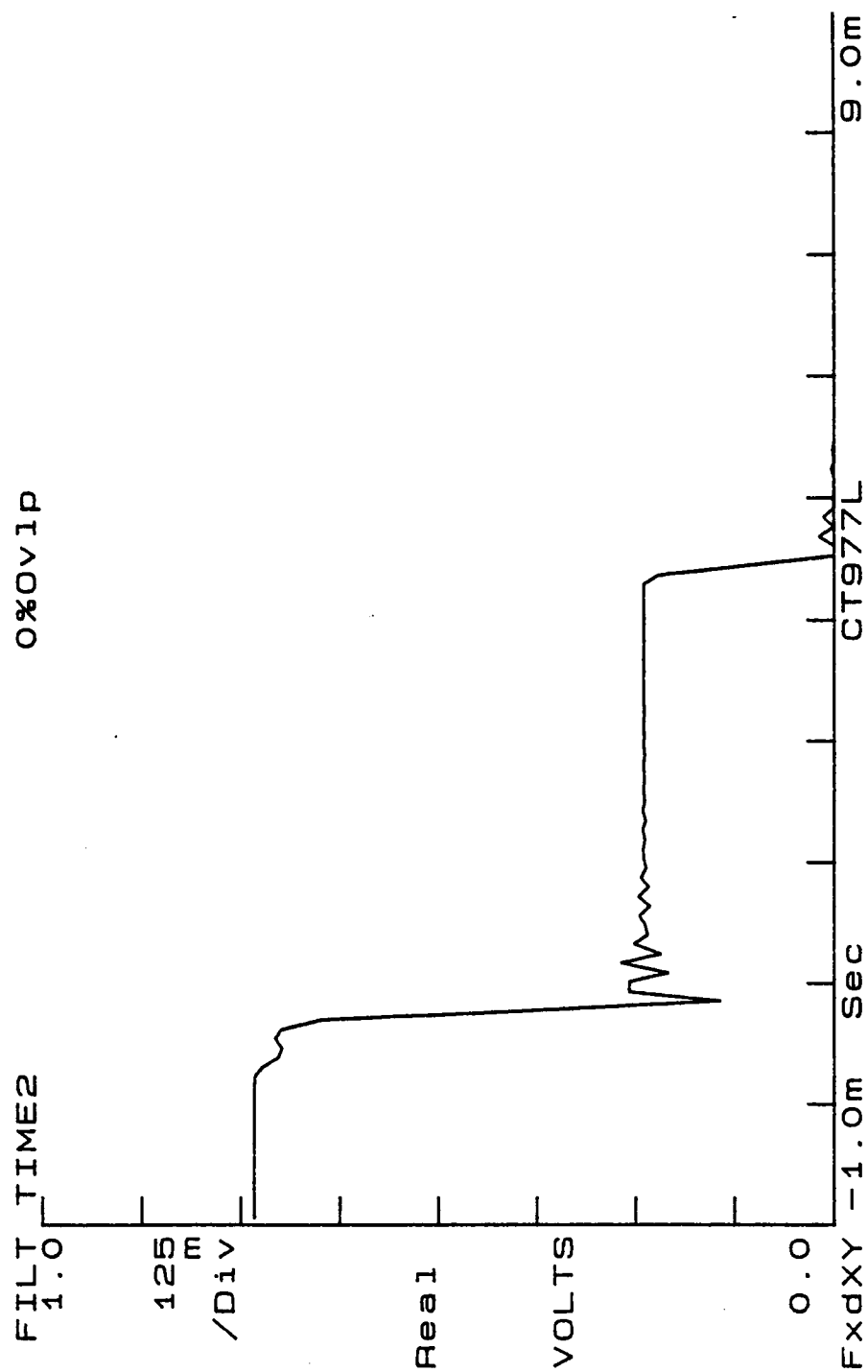


Figure 13. Example plot taken from the voltage drops measured by the fracture measurement system

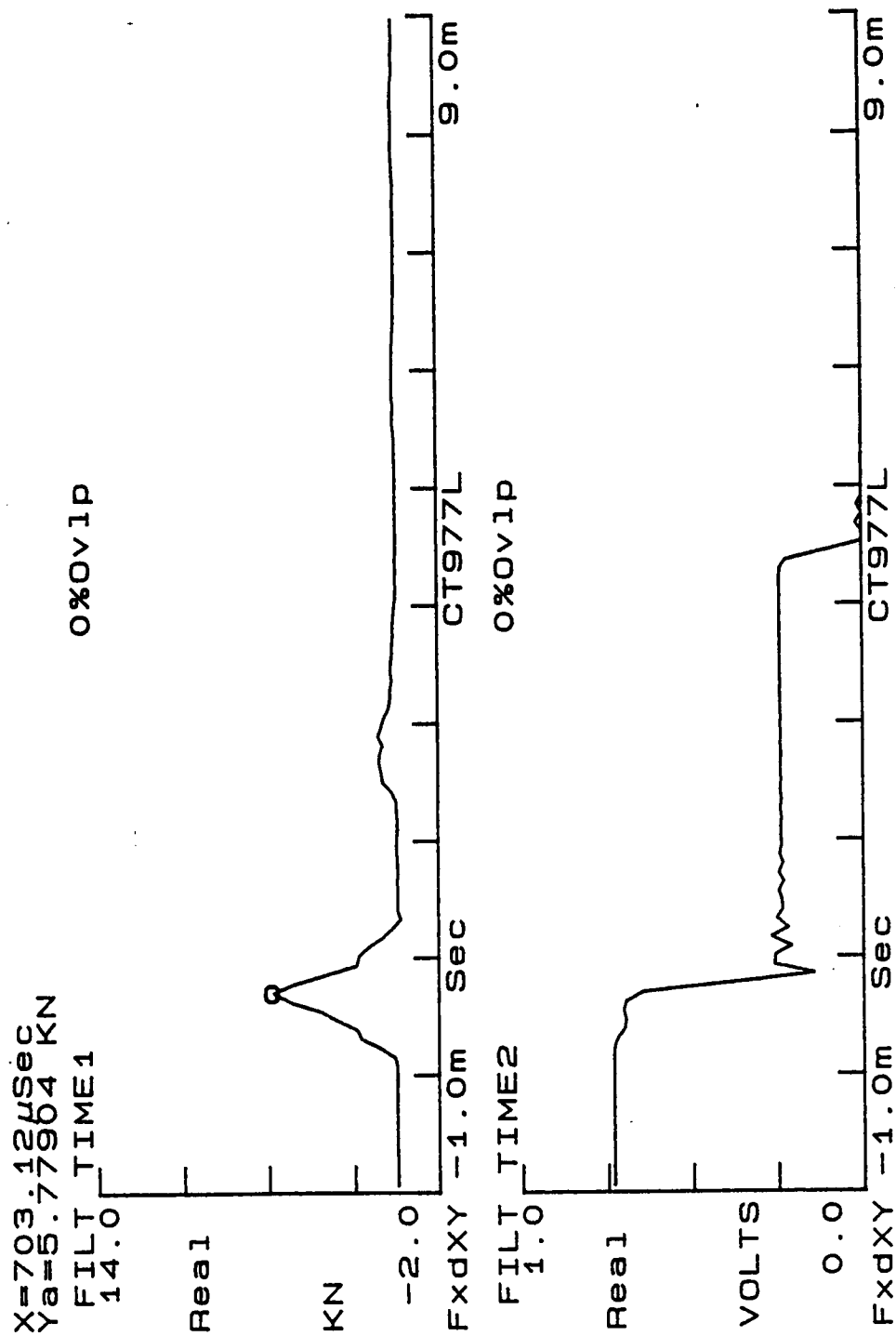


Figure 14. An example plot showing the fracture force measured by the force transducer (above) and voltage drops measured by the fracture measurement system (below)

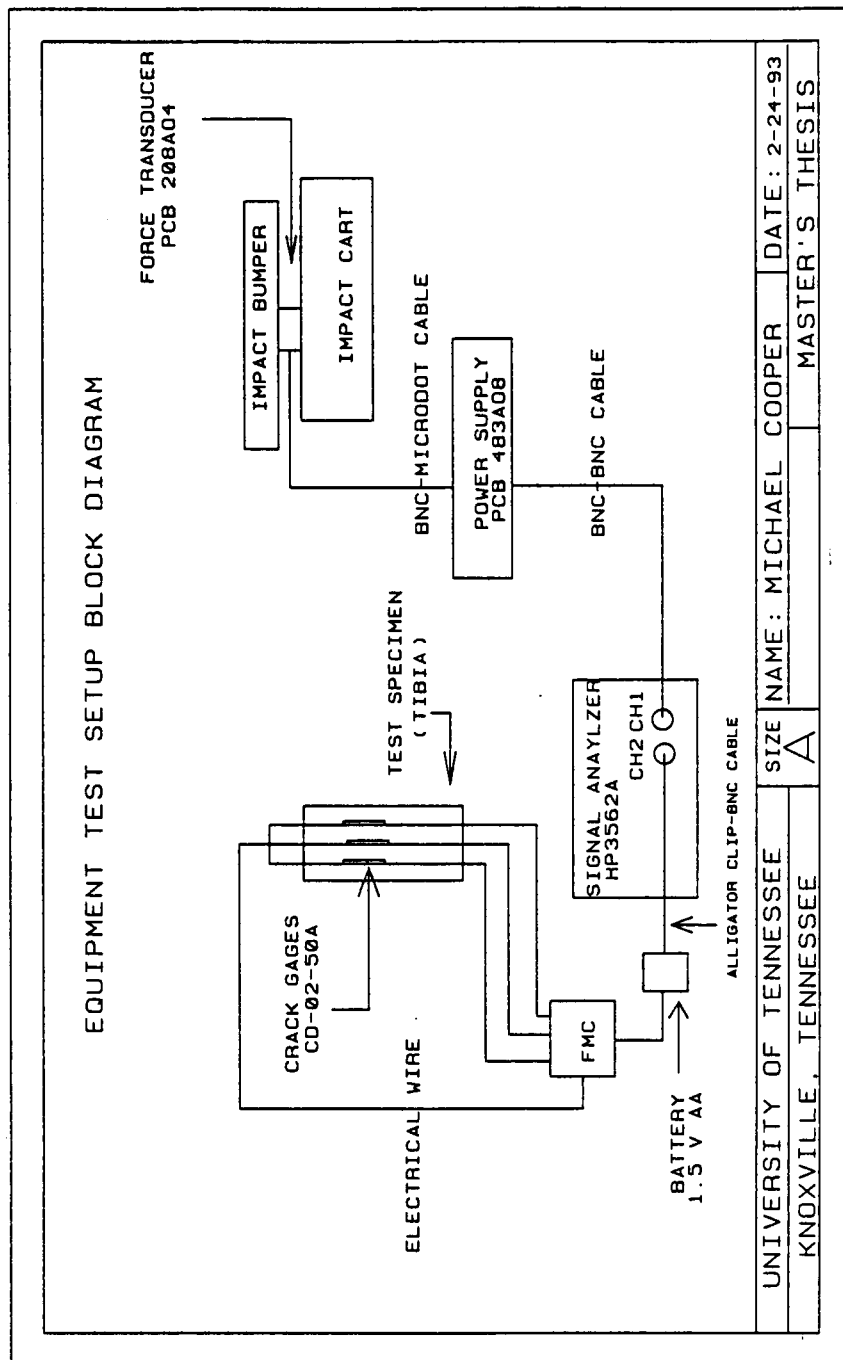


Figure 15. Block diagram of the equipment setup for the hang test.

Linear Resolution									
MEASURE:	CHAN 1 Freq Resp	CHAN 2 Freq Resp							
WINDOW:	CHAN 1 Uniform	CHAN 2 Uniform							
AVERAGE:	TYPE Stable	# AVGS 1	OVERLAP 0%	TIME AVG Off					
FREQ:	CENTER 2.5 KHz		SPAN 5.0KHz	BW 6.25 Hz					
	REC LGTH 160ms	Δt 78.1 μ S							
TRIGGER:	TYPE External	LEVEL 1.0 EU1	SLOPE Pos	PREVIEW Off					
INPUT:	RANGE	ENG UNITS	COUPLING	DELAY					
CH 1	10.0 Vpk	1.16 V/EU	DC (Flt)	-1.02ms					
CH 2	10.0 Vpk	1.0 V/EU	DC (Flt)	-1.02ms					
SOURCE:	TYPE		LEVEL	OFFSET					
	Rndm Noise		0.0 Vpk	0.0 Vpk					

Figure 16. Setup state of the digital analyzer for the hang test

Chapter IV

Experimental Approach

A. Test Parameters

For the tests, the following bone and motion parameters were measured and recorded: impactor velocity, force during impact, bone anatomical dimensions, and after-impact damage state. Following testing, cortical thickness measurements were taken at six points and averaged--figure 17 (Viano, Anukhalil, 1979; and Moore, 1985). All tibia cortical bone volumes were determined after removal of the bone marrow and after careful measurement of the displaced volumes when the bones were immersed in a beaker of water (Kress, 1989). All specimens were carefully photographed before and after impact testing. Fracture force, fracture time, and fracture propagation were determined from force versus time and volts versus time plots. The experimental data will serve to verify the computational model, as well as, predict bone fractures comparative to those present in persons who have been injured in the leg lift stage of running or walking.

B. Specimen Preparation

Since the objective of this thesis is to compare a computational computer model of a tibia midshaft impact to the experimental data, twelve embalmed human tibia-fibula bones with soft tissue were obtained from the University of Louisville. All soft tissue, periosteum, and the fibula were removed from the tibia with a scalpel and surgical scraper (1 & 2 -figure 18). After the tibia bone was cleaned, especially around the shaft, the bone was wiped with a isopropyl alcohol to help in the removal of the embalming fluid from the bone's surface. The medial side of the tibias were

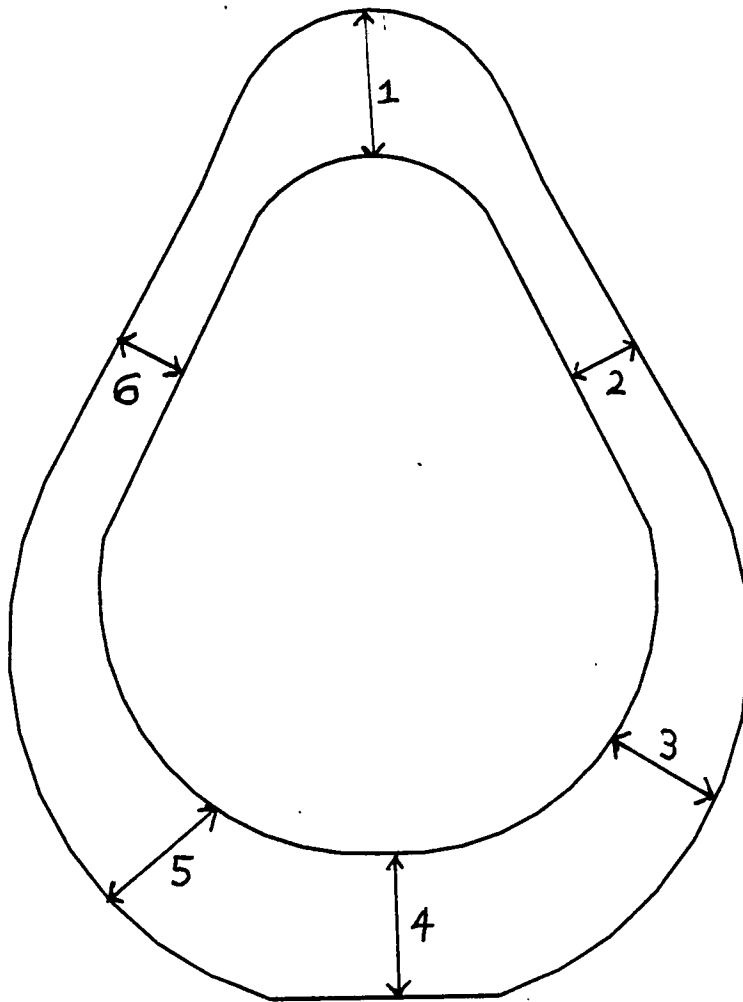


Figure 17. Idealized tibia cross-section showing the points of cortical thickness measurements

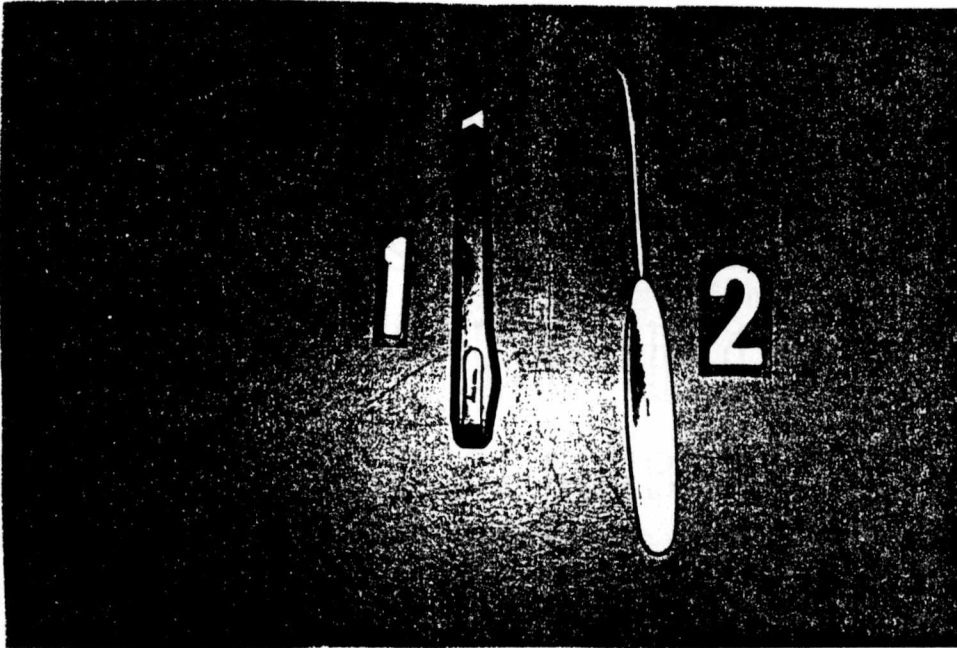


Figure 18. Instruments used for removal of soft tissue from the tibia
(scalpel (1), surgical scraper (2))

then sanded with 32 grade sandpaper to prepare the surface of the bone for crack gage application--as specified by Measurements Group. The sandpaper also helped to remove any periosteum or soft tissue that was inadvertently left on the tibia shaft.

C. Crack Gage Application

Three crack gages were applied to the medial side of each of the twelve tibia bones. The crack gages were positioned as shown in figure 19. One gage was positioned on the distal end of the shaft between the anterior and posterior direction (medial-distal-middle), while the remaining two gages were positioned midshaft anterior (medial-midshaft-anterior) and midshaft posterior (medial-midshaft-posterior). The gages were then bonded to the tibia with MBOND 200 adhesive and catalyst and by one minute of thumb pressure to aid in the glue curing process; complete curing for the bone occurs in three to five hours depending on temperature and humidity in the room. Gages should be used within twenty four hours of application or a silicone covering must be applied to protect the gages from corrosion.

D. Bone Potting Process

The specimens were placed in a specimen holding device, as discussed in chapter 3, the steel cups filled previous to testing with woods metal. The steel cup that was applied to the distal end of the tibia contained approximately three pounds more of the woods metal to account for the reduced volume of the bone on that end. The steel cups filled with woods metal were then placed in a boiling water bath until the woods metal was liquified. One steel cup was removed from the hot water bath and placed in a steel cup fixation device with one end of the bone submerged

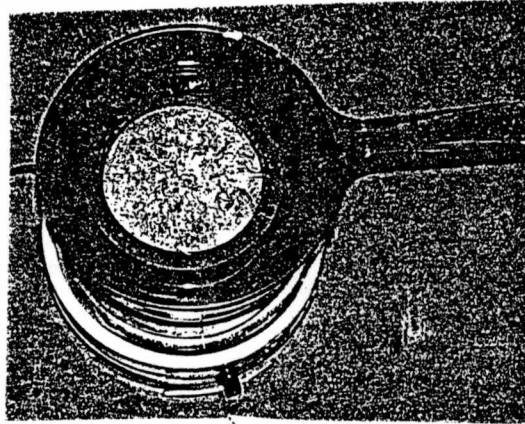


Figure 19. Mounted positions of the crack gages on the tibia

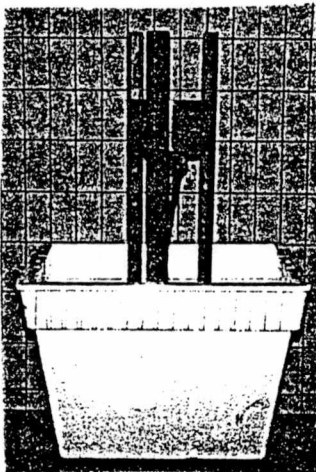
into the center of the woods metal filled cup. The tibia was held submerged if necessary by a seven to fifteen pound weight. The fixation device containing the bone and steel cups were placed in a cold water bath to allow the wood metal to solidify, thus fixing that end of the tibia in the steel cup; the cold water acted as a heat sink to quicken the hardening process. Once the woods metal was solid, the bone was removed and the unpotted end of the bone was submerged in another steel cup filled with woods metal; the same process of woods metal hardening was followed as before. Figure 20 shows a pictorial representation of the specimen potting process. Upon completion of potting both ends of the bone, the potted specimen was removed from the fixation device and the wires to the electronic fracture measurement circuit (fmc) were soldered to the crack gages. After specimen impact the steel cups were again put in a hot water bath for specimen extraction from the woods metal.

F. Soldering Fracture Measurement Circuit Wires

The copper tabs of the crack gages were covered with a hard substance that required removal by knife scraping. Once the covering was removed, the wires that connected the gages to the fmc were soldered into place. The wires were taped to the bone and steel cups: cellophane tape for the first five tibias, and duct tape for the last seven tibias. This tape kept the wires from moving around during the soldering process, and thus ensured good solder connections. The soldering was done using a rosin core solder. Although special care was taken to avoid applying the tip of the soldering iron directly to the copper tabs of the crack gages for long periods of time, overheating of the tabs did occasionally occur and cause the tabs to pop loose from the crack gage. The wires were color coded



1) Boiling water bath



2) Cold water bath and specimen mounting



3) Final mounted specimen

Figure 20. Pictorial representation of the specimen potting process

blue, green, and orange, with each color representing a different resistor value. Blue represented the 10 ohm resistor, orange represented the 47 ohm resistor, and green represented the 100 ohm resistor. Throughout the rest of this thesis the crack gages are referred to by color. The placement of the crack gages on the bone was noted for each specimen.

G. Specimen Mounting

When the specimens were potted and the wires soldered to the crack gages, the bone was then mounted for impact testing. The top of the proximal end steel cup was equipped with a removable plug that could be replaced by a steel eyehook; this steel eyehook was inserted and the specimen holding device was hung from a rope by this eyehook. The rope was positioned in such a way that the impact bumper of the cart would impact the bone at the midshaft. If necessary, the bone could be raised and lowered by tightening or loosening the rope or eyehook; the eyehook could also be rotated to bring the bone in a position for an anterior to posterior directional impact. The hanging of the bone allowed free motion upon impact. The fmc wires were then attached by alligator clips to the fmc.

H. Equipment Preparation

Once the fmc was connected to the HP3562A digital analyzer by alligator to BNC wire and the 1.5 volt battery was installed, the fmc was calibrated for each bone through measurement of the voltage drops as each wire was unclipped from the circuit. These values would then be used to decide which crack gage broke and when. After the fmc was calibrated, the test setup was ready for testing.

Chapter V

Finite Element Model Overview

The finite element method was applied to accurately model and analyze the laboratory experiments of a human tibia subjected to dynamic impact loading of 24.5 ft./sec. in the anterior to posterior direction of the tibia midshaft region. The finite element method will yield results that can be compared to the experimental data. This comparison will examine the general trend of the fracture propagation through the tibia as well as examining the fracture force data. Chapter V will discuss the overall concepts of a nonlinear, three-dimensional, transient, dynamic, solid mechanics, finite element code--PRONTO 3D. This chapter will also discuss the specifics of the constitutive model used to describe the material behavior of a dry, embalmed, human tibia, and will examine the development of the finite element model.

A. Overall Concepts of the Finite Element Code

Based on several factors, a nonlinear, three-dimensional, transient, dynamic, solid mechanics, finite element code was chosen for this research. First, due to the tibia's many contours and localized loading, the problem does not lend itself to a two-dimensional representation; therefore, a three-dimensional code was required. Second, the change in geometry from pre-impact stage to post-impact stage is large and also nonlinear in nature. Thus, a nonlinear code is needed again. Third, the nonlinear code is required because of the moving boundary conditions between the bumper and the bone in the model. Fourth, the impact loading on the bone is dynamic and requires a dynamic code for accurate modeling. With all of these considerations, PRONTO 3D, written by Taylor and Flanagan (1989),

was chosen as the finite element code and was utilized on a SUN SPARC station 2.

The finite element method, used in such codes as PRONTO 3D, is described by Kevin Brown (1992) as follows:

"The finite element method integrates the equation of motion forward in time using an explicit time integration operator. Knowing the accelerations, the explicit time integrator integrates the accelerations forward in time to obtain the velocities and displacements at the next time step. The new velocities and displacements are then used to calculate new shape functions and a new gradient operator--analogous to the strain-displacement matrix in linear finite element methods. The strain-increment is found from the strain-displacement matrix and the displacements themselves. The constitutive model is then used to relate the strain increment to a stress increment for each gauss point in each element. Having the new stress, the gradient operator is again used to calculate the nodal loads caused by the internal stresses. The global load vector is obtained by using the nodal loads caused from internal stresses and adding any external loads. New accelerations are then calculated and the iterative process is repeated until the problem is finished" (Brown, 1992).

B. Constitutive Model Mechanics

Two constitutive models were used in the finite element analysis: linear elastic, and fracture dry tib. The woods metal, steel cups, impact bumper, and load cell all were modeled with the linear elastic constitutive model. The dry, embalmed, human bone was modeled using the basis of an elastic-perfectly plastic isotropic material constitutive model with some modifications implemented in this routine--renamed fracture dry tib. The following section will discuss the basic governing equation of the linear elastic routine as well as the mechanics of the fracture dry tib routine.

1. Linear Elastic Constitutive Model Mechanics

The specimen fixation devices, loadcell, and bumper were all modeled as a linear elastic material. The governing equation for this constitutive model describes the linear elastic material modes using Hooke's law in rate

form as follows:

$$\sigma = \lambda (tr d) * \delta + \mu d \quad (1)$$

where λ and μ are the elastic Lamé material constants (Taylor and Flanagan, 1989). A detailed explanation of this model is not given due to its classical common usage. No changes were made to this model, and figure 21 shows a typical stress-strain curve for this material.

2. Fracture Dry Tib Constitutive Model Mechanics

The dry, embalmed, human tibia is modeled as an elastic-perfectly plastic isotropic material; the presentation of the elastic-perfectly plastic constitutive model is given in this section. The original constitutive model has been modified to induce fracture characteristics in the tibia. The basic philosophy of these modifications will also be discussed. The constitutive model has been named Fracture Dry Tib and will be referred to as such.

The general equations and numerical implementation of the elastic-perfectly plastic isotropic model is as follows:

The total stress and strain rate are broken into volumetric and deviatoric parts where the volumetric part is considered to be elastic. The deviatoric strain rate is comprised of elastic and plastic parts as described by

$$\dot{e} = \dot{e}_e + \dot{e}_p \quad (2)$$

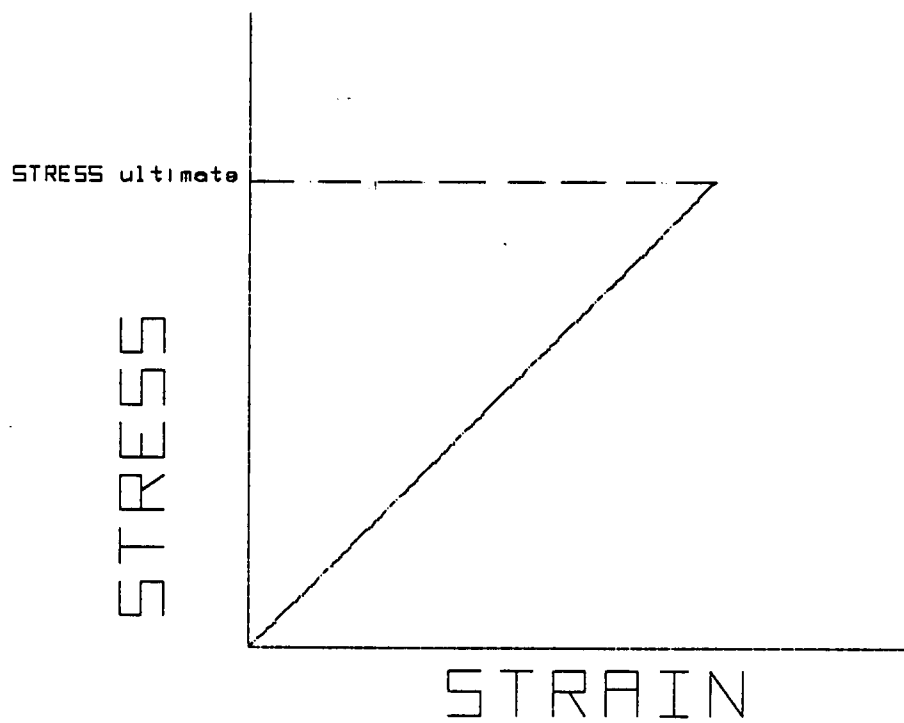


Figure 21. Stress-strain curve for a linear elastic material

where the elastic shear modulus, relates the elastic part to the stress rate by

$$\dot{\epsilon}_e = \frac{\dot{s}}{2G} \quad (3)$$

The plastic part, $\dot{\epsilon}_p$, is assumed to be normal to the yield surface,

$\phi = 0$. The von Mises yield surface is described in the deviatoric stress space form of a hypersphere of radius R

$$\phi = s \cdot s - R^2 \quad (4)$$

The stress state is required to remain on the yield surface during plastic loading. A vector differential equation for a plastic loading process is given for any general strain rate and for the absolute value of the stress state equal to the yield radius, R, by

$$\dot{s} = \frac{2G\dot{\epsilon} - 2sG(s \cdot \dot{\epsilon})}{R^2} \quad (5)$$

This produces a stress rate vector which is tangent to the yield surface at S.

The elastic-perfectly plastic process can be integrated by a variety of numerical algorithms. These algorithms can be used to directly calculate a final stress state from the initial state S_0 on the yield surface as

$S_0 + \dot{s} \Delta t$. However, the stress rate is tangent to the yield surface at S_0 ,

producing a stress state that is outside the yield surface. The numerical algorithm must produce a final state on the yield surface for any strain increment.

The most popular algorithm for integration purposes is the radial return method. For this method the trial stress and trial yield state are found by $\mathbf{s}_T = \mathbf{s}_e + \dot{\mathbf{s}} \Delta t$ and equation 4 respectively. If ϕ_T is found to be positive, then equation 5 is applied as

$$\bar{\mathbf{s}}_r = \mathbf{s}_T - 2GR^{-2} (\bar{\mathbf{s}}_r \cdot \dot{\mathbf{s}}) \bar{\mathbf{s}}_r \Delta t_p \quad (6)$$

However, the final stress state is seen to be colinear or parallel to the trial stress state. For computational ease, the unit vector of the final stress state divided by the yield radius is replaced by the trial stress state divided by the absolute value of the trial stress state. The final stress state is normally found to be inside the yield surface. A radial correction would be made to force the final stress state to lie on the yield surface. The radial return method allows the trial stress state and the final stress state to be calculated without solving for the contact stress state. Figure 22 shows a diagram of the radial return method for the elastic-perfectly plastic isotropic model in deviatoric space (Krieg and Krieg, 1977). **NOTE:** The above description of the elastic-perfectly plastic model is taken in parts from the referenced paper by Krieg and Krieg.

To simulate fracture in the constitutive model, a few modifications were made. The basic philosophy in modeling the fracture characteristic of the tibia is to allow for two modes of fracture. If fracture does not take

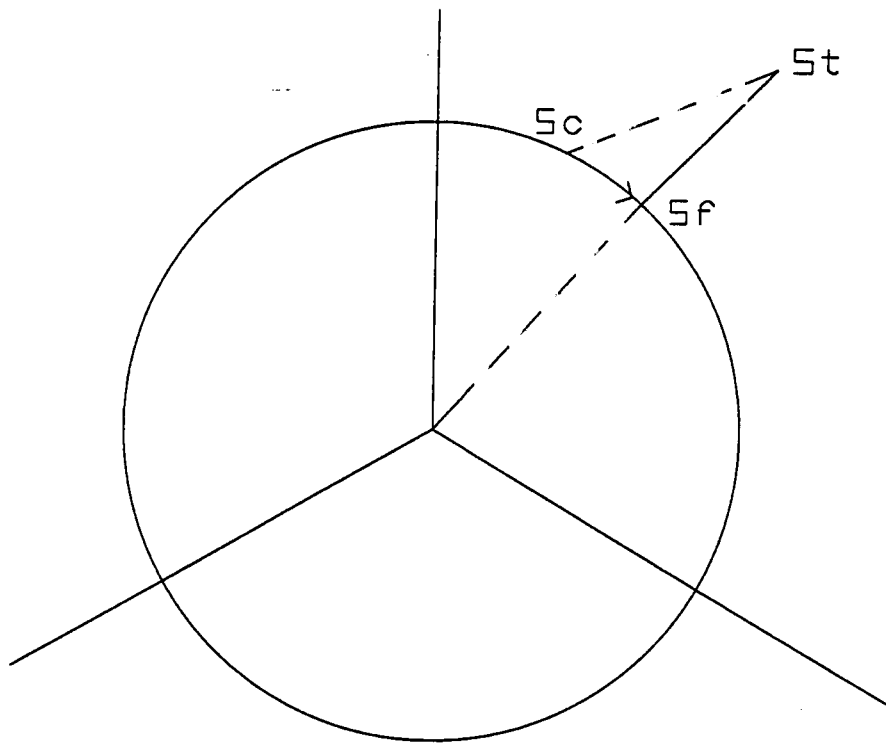


Figure 22. Geometric interpretation of the radial return method in deviatoric space

place then the element will maintain a linear elastic state. One mode will occur when the effective stress of a bone element is larger than some specified fracture stress of bone. Therefore, when

$$\sqrt{3 * \left(\frac{1}{6} * ((\sigma_1 - \sigma_2)^2 + (\sigma_2 - \sigma_3)^2 + (\sigma_3 - \sigma_1)^2) + (\sigma_{12}^2 + \sigma_{13}^2 + \sigma_{23}^2) \right)} \quad \text{becomes}$$

greater than or equal to the fracture stress, then the elements yield stress is dropped to some small value (200 psi). The element is allowed to continue in a perfectly plastic state; otherwise, the element will maintain a linear elastic state. By utilizing this method, the stress can still propagate through the element; but the distortional stress is greatly reduced while the element volume remains constant. An example of the stress-strain curve for the tibia is shown in figure 23.

The second mode of failure occurs when the mean stress in the element becomes too tensile. When this occurs, the element's elastic shear and bulk moduli are reduced to a small fraction of their original values. This allows the element to maintain constant volume in compression while allowing it to distend easily.

These two methods of failure are meant only to approximate the fracture characteristic of bone and in no way are exact models of fracture. The constitutive model does not track the fracture direction and only shows regions where failure has occurred. This is considered a weakness in the model. Most fracture characteristics in constitutive modeling use a maximum stress calculation to characterize fracture and track the fracture propagation direction. This method should be employed at a later date.

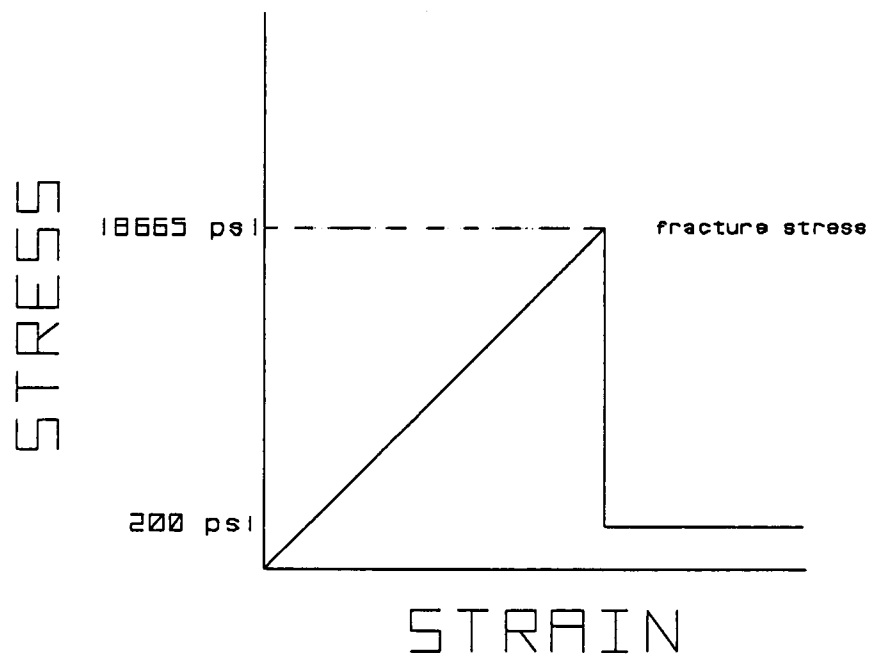


Figure 23. Stress-strain curve for the fracture dry tib constitutive model for the effective stress fracture mode

However, for a first time approximation, the method used to predict fracture failure is considered appropriate.

C. Finite Element Model Development

The finite element model used for this research was developed using H8 elements with one Gauss point integration and an antihourglassing stiffness algorithm (Taylor and Flanagan, 1989; Belytschko, 1981). The model is comprised of 13,347 elements (18,216 nodes) that are arranged in an idealized cross-section of a human tibia (figure 24). The cross-section geometry and average cortical thicknesses are approximately equal to that measured in the experimentally tested tibias. The average cortical thickness for the finite element model at midshaft is 0.263 inches. The length of the bone is approximately 13 inches which is also the average length of the experimental tibias. The tibia specimens ranged in age from 42 to 85 years and consisted of both male and female tibias. The proximal and distal ends of the bone are covered with the models for the woods metal filled steel cups with the appropriate volume of bone inside each cup. The bumper and load cell are also modeled as impacting the bone in the anterior to posterior direction midshaft. The bumper and load cell are given a prescribed velocity of 24.5 ft. per second to simulate actual velocity values used in the laboratory. Figure 25 shows the three dimensional finite element model. The load cell will measure the force required to push the bumper through the bone, thus measuring the fracture force. The mesh becomes more coarse in the tibia as the distance from the region of impact increases. There are no boundary conditions on the tibia, and the specimen holding device allowing it to move freely on impact. The bumper and load cell are allowed to move only in the Y direction and the velocity is

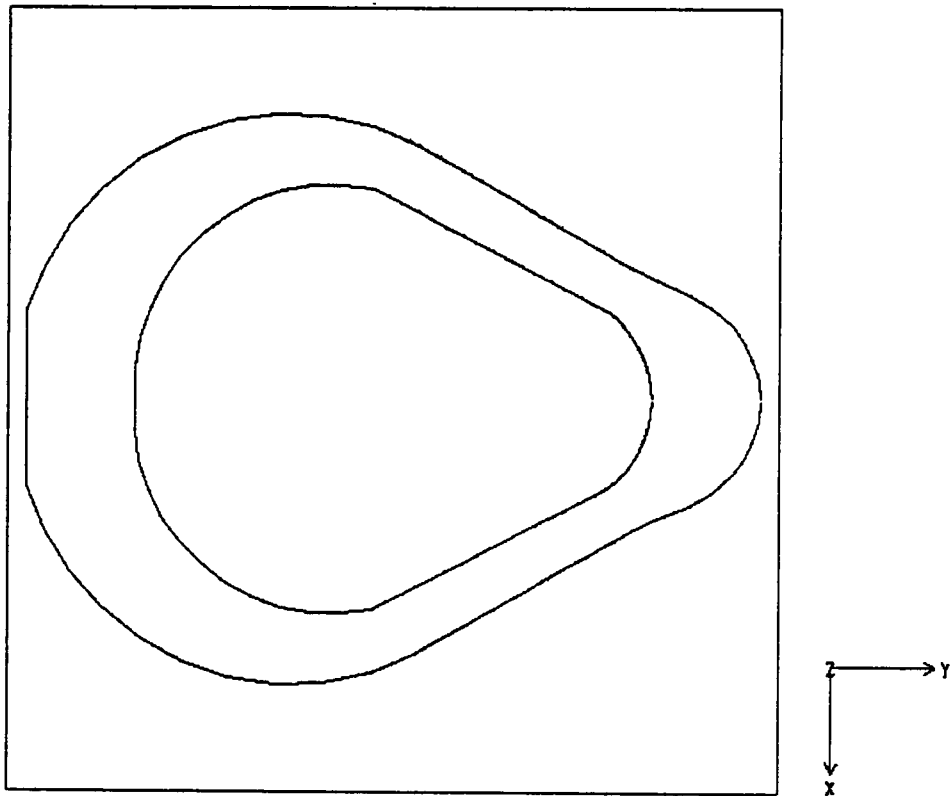


Figure 24. Idealized bone cross-section used for finite element modeling

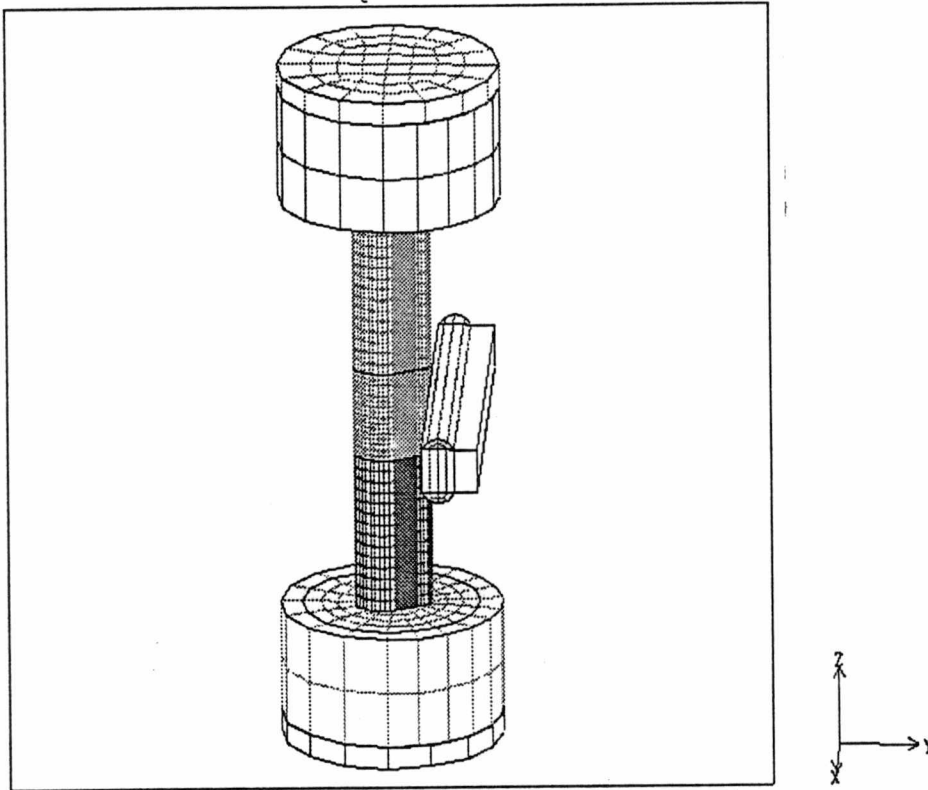


Figure 25. Wireframe mesh of the potted tibia used in the finite element model

prescribed solely in the negative Y direction. There is a contact surface between the bone and bumper which allows impact to occur. Also, each section of the tibia and specimen holding device (distal, midshaft, proximal tibia sections and woods metal and steel cups) was developed individually and zippered together in the finite element code. Gravitational effects are neglected. Time runs for this model start from the initial time of bumper and tibia contact.

D. Material Properties Used in the Finite Element Model

The material properties used in the finite element model are of utmost importance in insuring accurate results. The five types of materials that were modeled are the steel cups, woods metal (fixating material), impact bumper, load cell, and dry, embalmed, human tibia. The material properties used for each of these materials can be found in table 4. The material properties for the dry, embalmed tibia are taken from table 2 with the exception of the value for the density which was determined experimentally. The density value was determined from the average, experimental, cortical bone volume divided by the average experimental bone mass. The experimental process for determining the bone volume is described in chapter IV.

Table 4

Material Properties for Finite Element Modeling

Mechanical Properties	Dry Embalmmed Human Tibia	Steel Cups	Steel Impact Bumper	Load Cell	Woods Metal
Young's Modulus	3.15E+6 psi	29E+6 psi	30E+6 psi	13E+9 psi	4.6E+6 psi
Ultimate Stress	18665 psi	----	----	----	----
Yield Stress	18665 psi	----	----	----	----
Poisson's Ratio	0.3	0.3	0.3	0.3	0.3
Density lb-sec ² /in ⁴	0.000135	0.0007366	0.0005103	0.0005103	0.0007366

Chapter VI

Results and Discussion

This section is divided into two primary areas of concern: the experimental approach and the finite element approach. The experimental section will detail the results obtained from the testing of eleven, dry, embalmed, human tibias through impact loading at a velocity of 24.5 ft./sec. in the anterior to posterior (A-P) direction at the tibial midshaft. The general trend of crack propagation in the tibia and the fracture force data will be the main emphases. The finite element analysis section will detail the general trend of crack propagation and fracture force data for a generalized three-dimensional tibia model subjected to the same conditions as the experimentally tested tibias. The finite element analysis was performed on the nonlinear, three-dimensional, transient, dynamic, solid mechanics code--PRONTO 3D.

A. Experimental Results

1. Crack Propagation

To analyze the general trends of crack propagation in the tibia under impact conditions, crack gages were utilized and incorporated into the data acquisition system as described in chapter IV. Three crack gages were used to analyze the crack propagation; the gages were positioned as described in chapter IV. As the crack propagated across the gages, a voltage drop was measured. Out of the eleven tested tibias, only five of the tibia tests produced reliable crack propagation data. Figures 26, 27, 28, 29, and 30 show the voltage versus time plots detected by the crack gages. Each voltage drop corresponds to a calibrated value that is measured before each tibia is impacted. For these five tibias, the crack

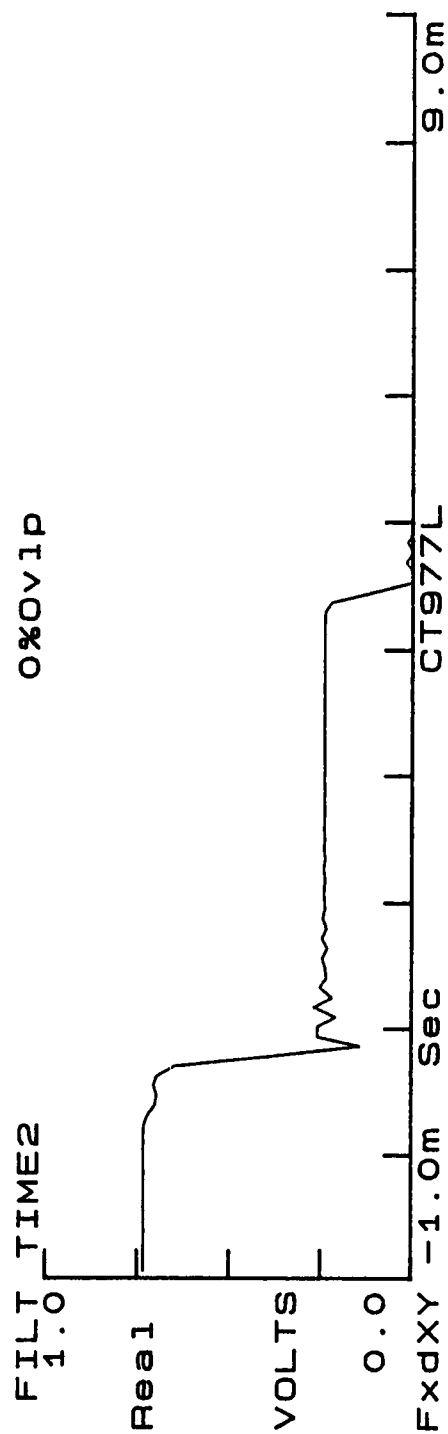


Figure 26. Crack gage data for tibia specimen 977L

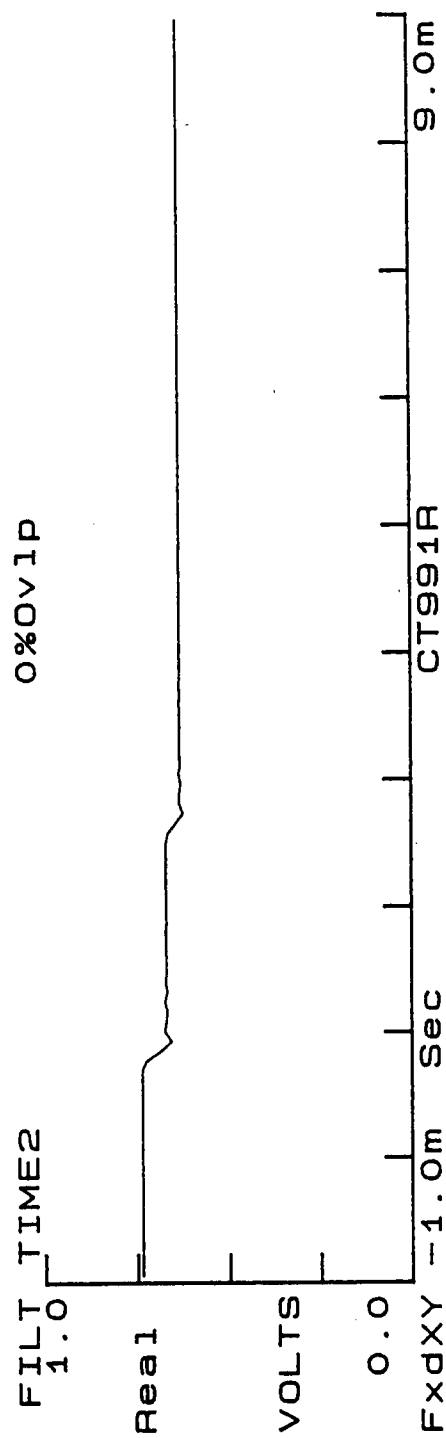


Figure 27. Crack gage data for tibia specimen 991R

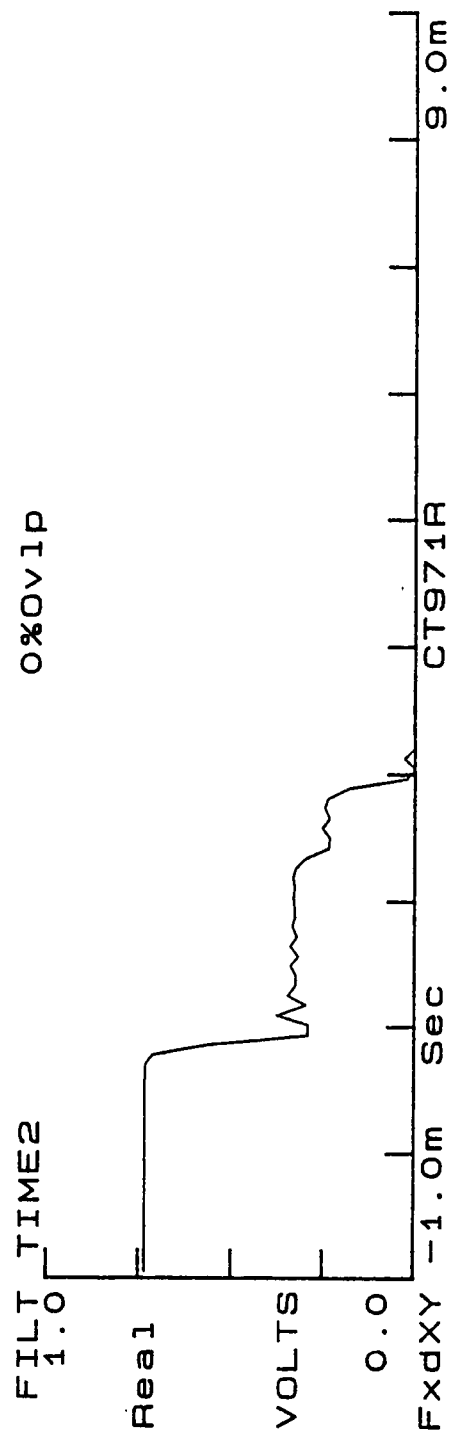


Figure 28. Crack gage data for tibia specimen 971R

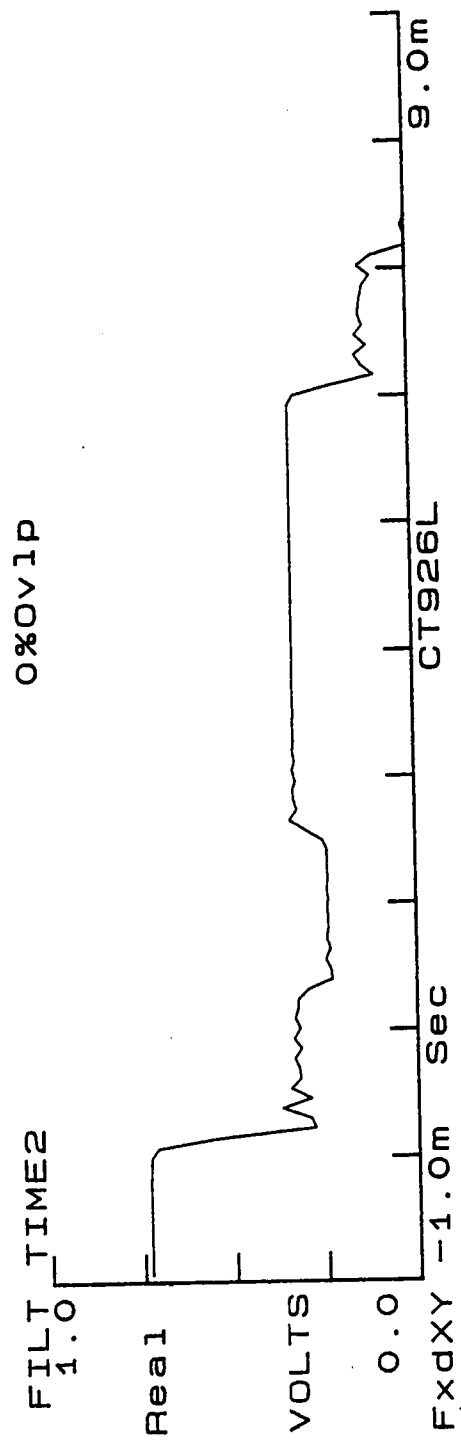


Figure 29. Crack gage data for tibia specimen 926L

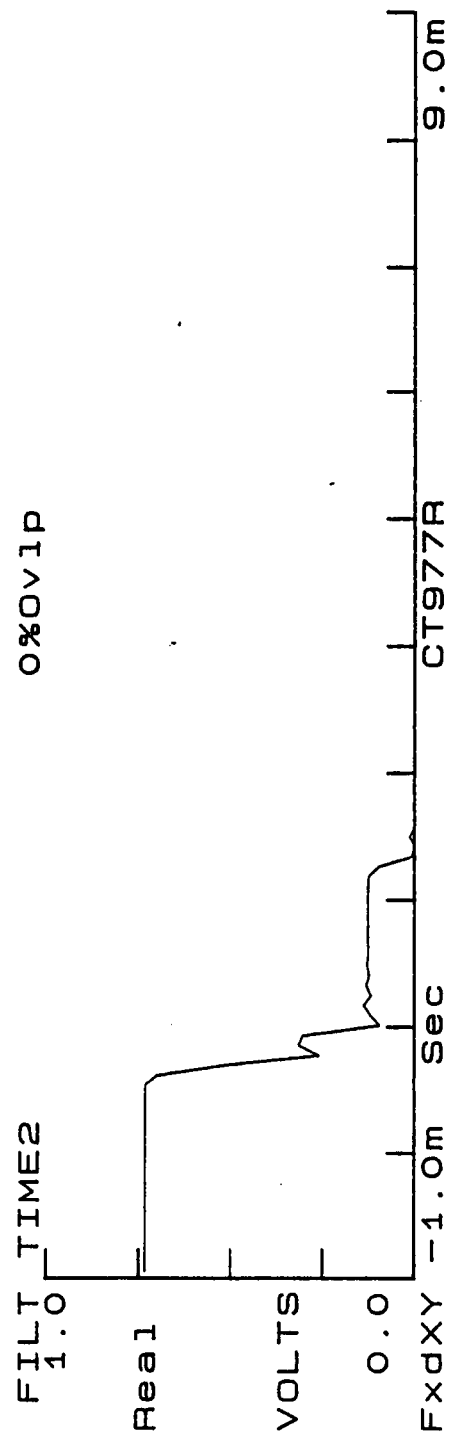


Figure 30. Crack gage data for tibia specimen 977R

gages were positioned with the blue gage in the medial-midshaft-posterior position, the orange gage in the medial-distal-middle position, and the green gage in the medial-midshaft-anterior position. Figure 26 shows the volts versus time plot for tibia 977L. Notice the four distinct voltage levels. The first voltage level, at the start of the time run (740 mV), shows that all the gages are unbroken and indicate no crack propagation. The next voltage level that occurs at approximately 712 mV indicates that the green gage is broken, and the crack has propagated through this point. The voltage drops to about 245 mV indicating that the blue gage is broken and then proceeds to drop to the next level of 0.0 mV which indicates that the remaining orange gage is also broken. Post-impact photos (figure 31) are examined to aid in determining the crack propagation. The photo shows that all the gages did indeed break and that the voltage plot provided reliable data. From this plot, it can be seen that the trend of the crack propagation for this bone is from the anterior to posterior direction with the distal area breaking after the midshaft has fractured. There are some noteworthy points observed about this particular bone. Before impact, a hairline fracture in tibia 977L, was noticed running longitudinally on the lateral side of the bone from the proximal to the distal ends. Tibia 977L was tested anyway due to the lack of test specimens available for this research. The determination of whether this crack played a part in the direction of the crack propagation will be discussed in the next chapter. Figure 27 shows the voltage plots for tibia 991R. This plot shows only three voltage levels with the voltage never dropping to 0.0 mV which indicates one of the gages remained intact throughout the entire experiment. The first voltage level occurs at



Figure 31. Post-impact photos for tibia specimen 977L

approximately 748 mV indicating all gages are intact; the next level at 681 mV indicates that the orange gage is broken; the third voltage level at 643 mV indicates that the green gage is broken. The blue gage appears to have never been broken. Therefore, the crack propagated from the medial-distal-middle position to the anterior position. The posterior region, however, remained intact. Post-impact photos for this tibia were not available. The voltage plot for tibia 971R is shown in figure 28. There are four distinct voltage levels with the first occurring at 735 mV which is indicative of all gages unbroken; voltage level two occurs at 332 mV indicating the blue gage is broken; voltage level three at 243 mV signifies that the green gage is broken; level four at 0.0 mV shows the orange gage to be broken. Figure 32 shows post-impact photo. It can be seen that the posterior blue gage never broke and did not adhere to the bone even though fracture did occur in this region. Instead, the blue gage pulled loose at the solder connection. It was, therefore, unable to detect the fracture. The orange gage also appears completely intact with no fracture occurring in its region. It is assumed that this gage pulled loose at the solder connection as well. This photo also shows that the fracture did not splinter the bone as was seen in the other photos. Thus, the trend of the crack is to propagate in the posterior to anterior direction (P-A). Figure 29 of tibia 926L shows six voltage levels which complicate the determination of the crack propagation. The first level at 736 mV signifies that all gages are intact; the second level at 332 mV shows that the blue gage is broken; the third level at 243 mV indicates that the green gage is broken; level four jumps back up to 332 mV indicating that only the blue gage is broken now; the fifth level drops down again to 130 mV signifying that the blue and



Figure 32. Post-impact photos for tibia specimen 971R

orange gage are broken but that the green gage is intact; finally, the sixth level at 0.0 indicates all gages are broken. The determination of the crack propagation is somewhat difficult with this plot. Looking at the post-impact photos (figure 33), it can be seen that the distal gage (orange) and the anterior gage (green) both remained intact. The green gage did not adhere to the bone, and even though there was fracture in that area, the gage was not able to detect it. It appears that the green gage broke loose at the solder connection. Based upon these observations it is assumed that the crack propagated in the P-A direction. The voltage jumps at levels 3, 4, and 5 can be explained by the green gage pulling loose at the solder joint and then briefly making contact again to allow the voltage to be measured across the gage before pulling loose again. From the post-impact photos it can also be seen that the orange gage pulled loose from its solder connection which indicates a broken gage even though this was not the case. Figure 30 shows the voltage plot for tibia 977R. There are three voltage levels. Level one occurs at 738 mV indicating that all gages are intact; level two at 130 mV signifies that the blue and orange gages are broken; level three occurs at 0.0 mV which is indicative of all the gages being broken. To help in the determination of the trend of the crack the post-impact photos (figures 34) are examined. The photo shows that the orange gage is fully intact and, once again, pulled loose from the solder connection. From the plot and photo it appears that the crack propagated in the P-A direction. Table 5 provides the determined direction of the crack propagation for each of the previously described tibias.

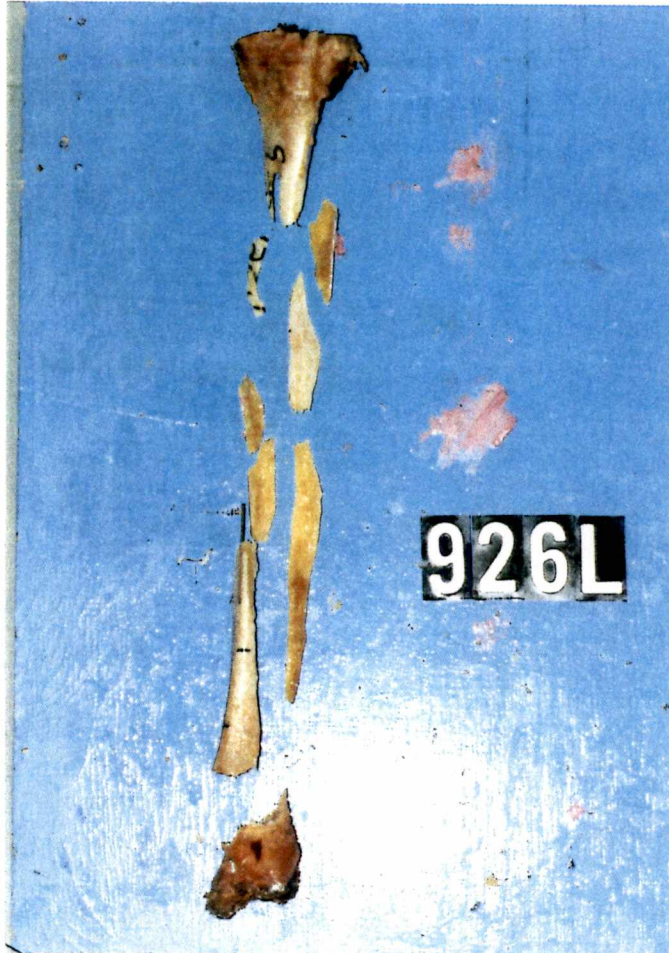


Figure 33. Post-impact photos for tibia specimen 926L



Figure 34. Post-impact photos for tibia specimen 977R

Table 5
Crack Propagation Trends

Tibia Specimen Numbers	Crack Propagation Direction
977L	Anterior to Posterior
991R	Medial-distal-middle to Anterior
971R	Posterior to Anterior
926L	Posterior to Anterior
977R	Posterior to Anterior

2. Fracture Force

Fracture force is another variable that was measured in the experimental testing. Force versus time plots were measured by the force transducer attached to the bumper configuration as described in chapter 4. All eleven tibias provided reliable data for the force determination. Fracture force was determined to be the peak level on the force versus time plots. The force values were determined in Kilonewtons and have been converted to pounds force. Figures 35 through 45 show the time plots for all eleven tibias, and table 6 provides the fracture force for all eleven tibias. Notice that figures 37, 41, 42 all show small humps before or after the fracture force. This can be caused by slightly off-center hits with the force transducer, and it is also possible that some vibration from the bumper configuration is also leaking into the force measurements as described by Kress (1989).

All force versus time plots and corresponding voltage versus time plots can be located in appendix A. Each plot is followed by the pre- and post-impact photos for each of the eleven tibias when possible. Appendix B also includes brief lab notes taken during the experiments. The lab notes contain the calibration table for the crack gages as well as the position of the gages, tibia number, sex, age, weight of the tibia, speed of impact, and general observed comments about the test. These notes are first impressions recorded by the researcher.

B. Finite Element Analysis

1. Crack Propagation

To analyze the crack propagation in the tibia model, several contour plots were developed of the diaphysis of the bone. These plots show two

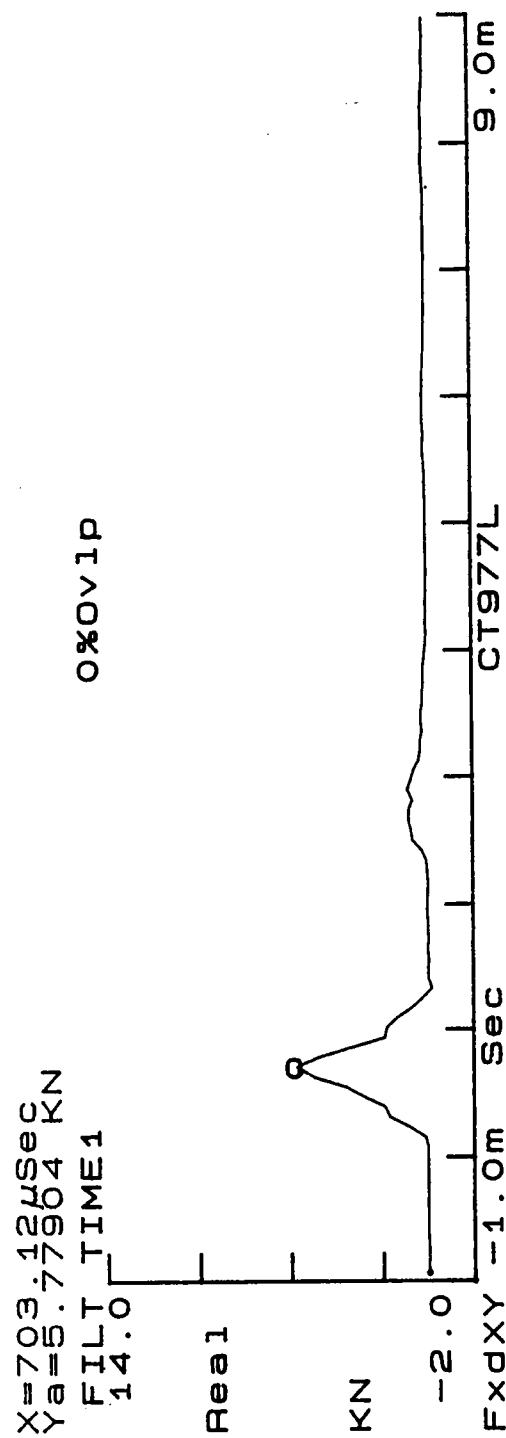


Figure 35. Fracture force data for tibia specimen 977L

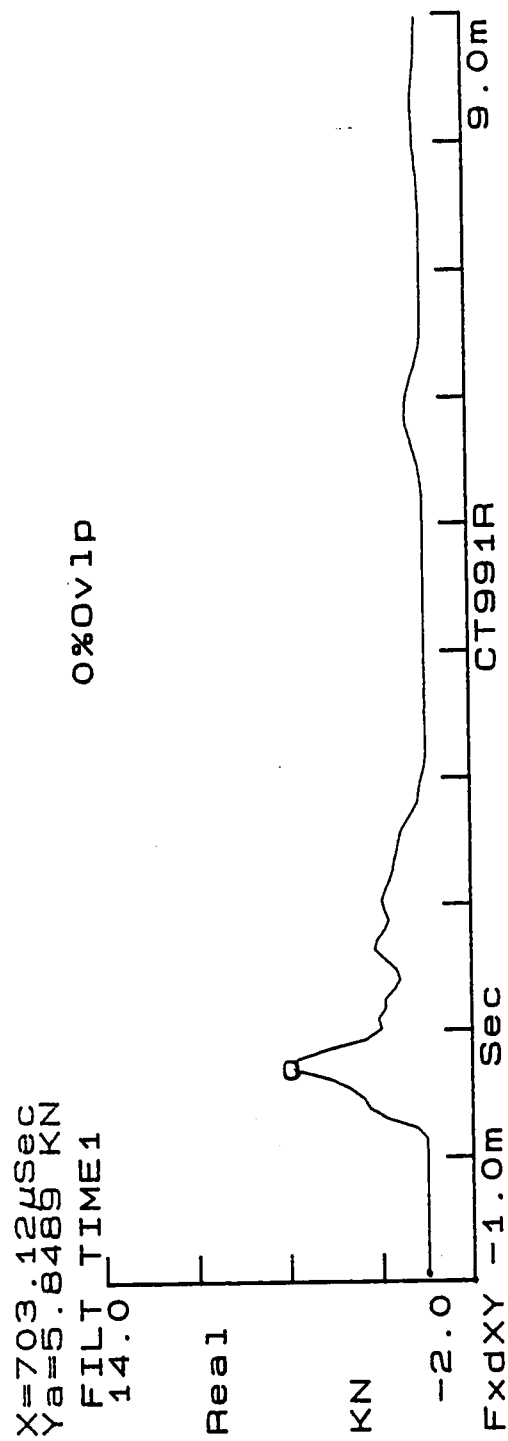


Figure 36. Fracture force data for tibia specimen 991R

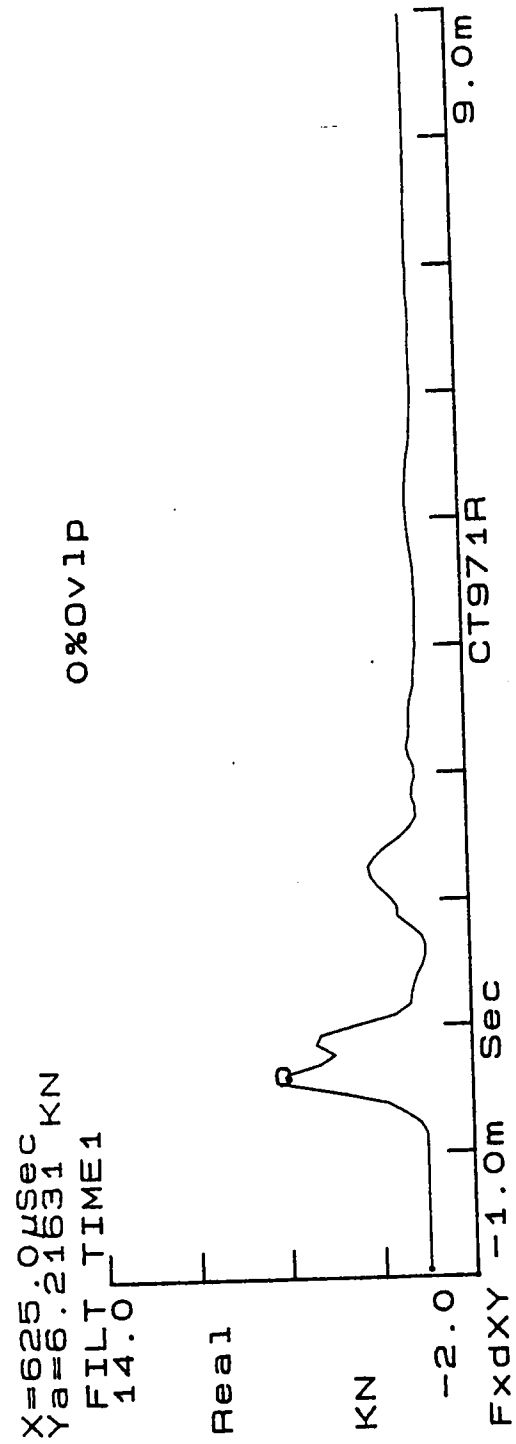


Figure 37. Fracture force data for tibia specimen 971R

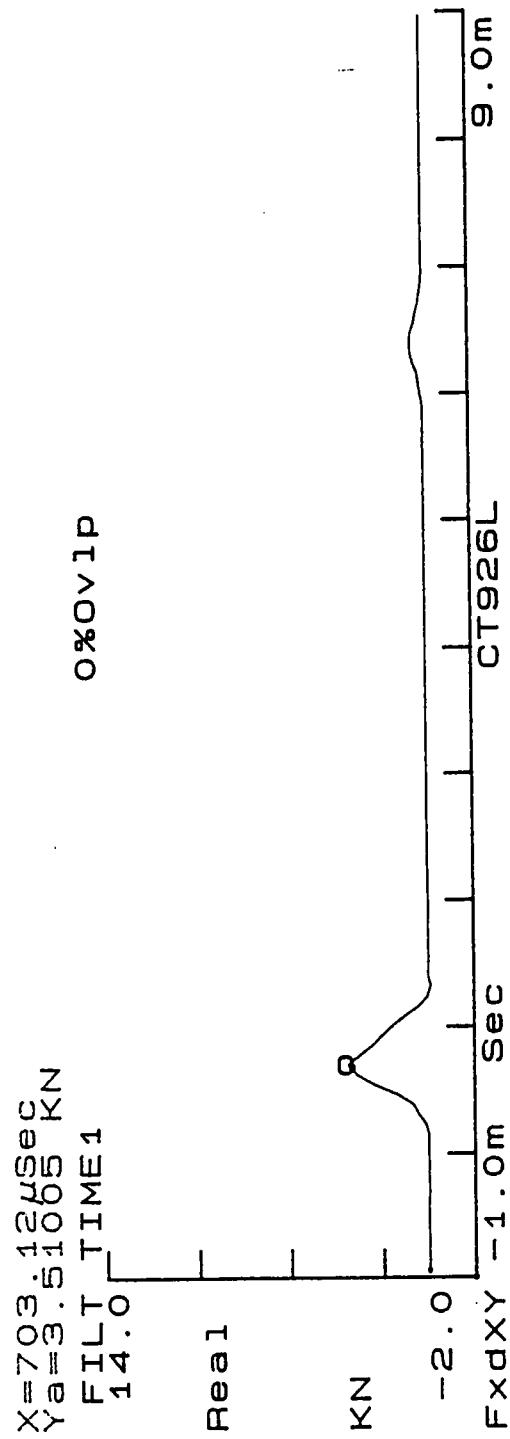


Figure 38. Fracture force data for tiawa specimen 926L

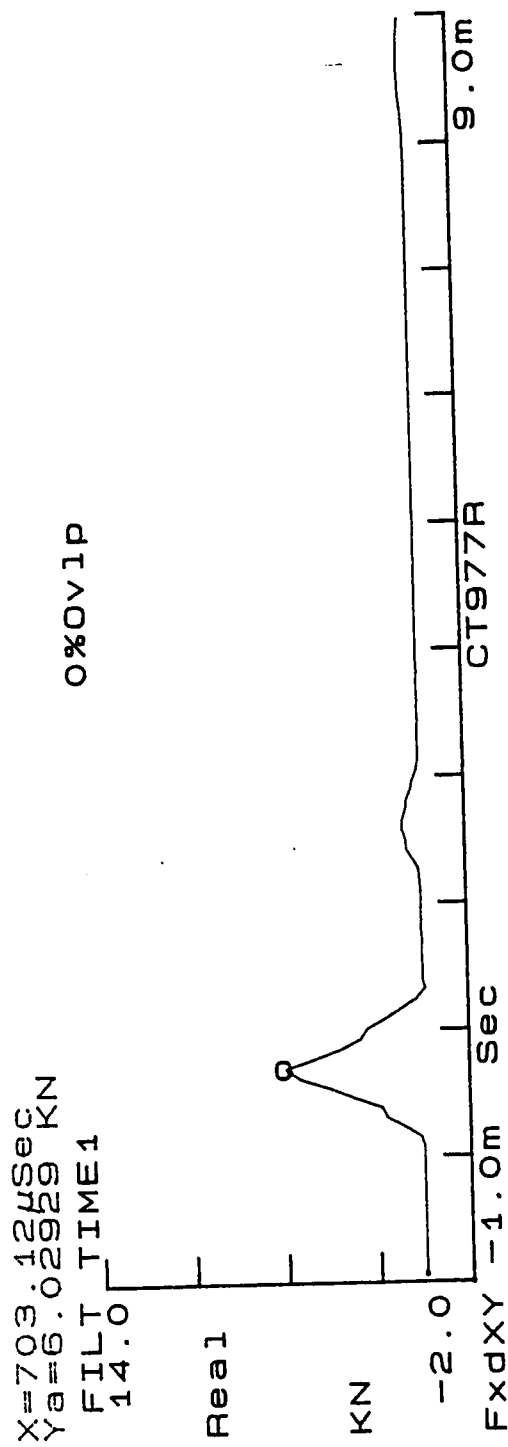


Figure 39. Fracture force data for tibia specimen 977R

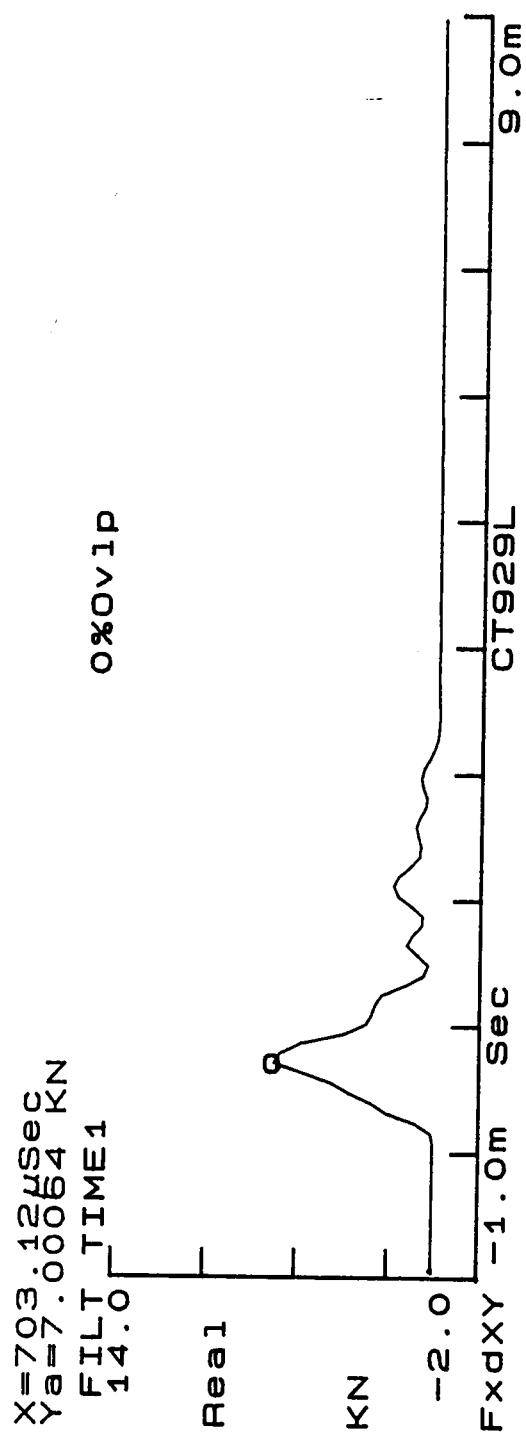


Figure 40. Fracture force data for tibia specimen 929L

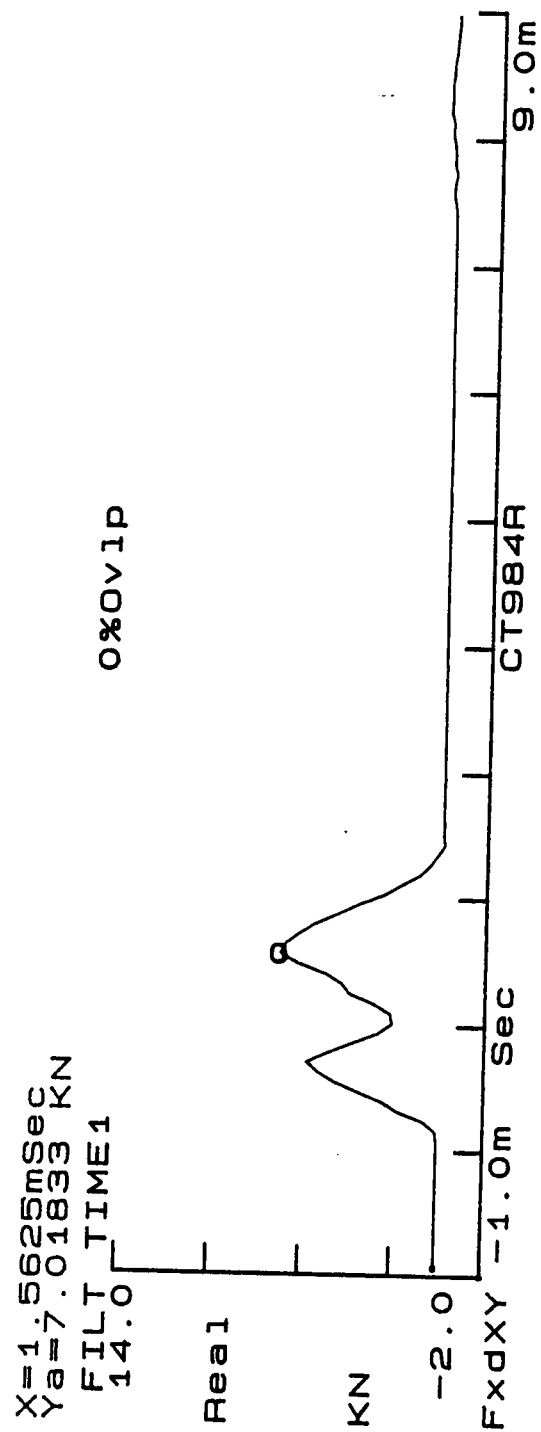


Figure 41. Fracture force data for tibia specimen 984R

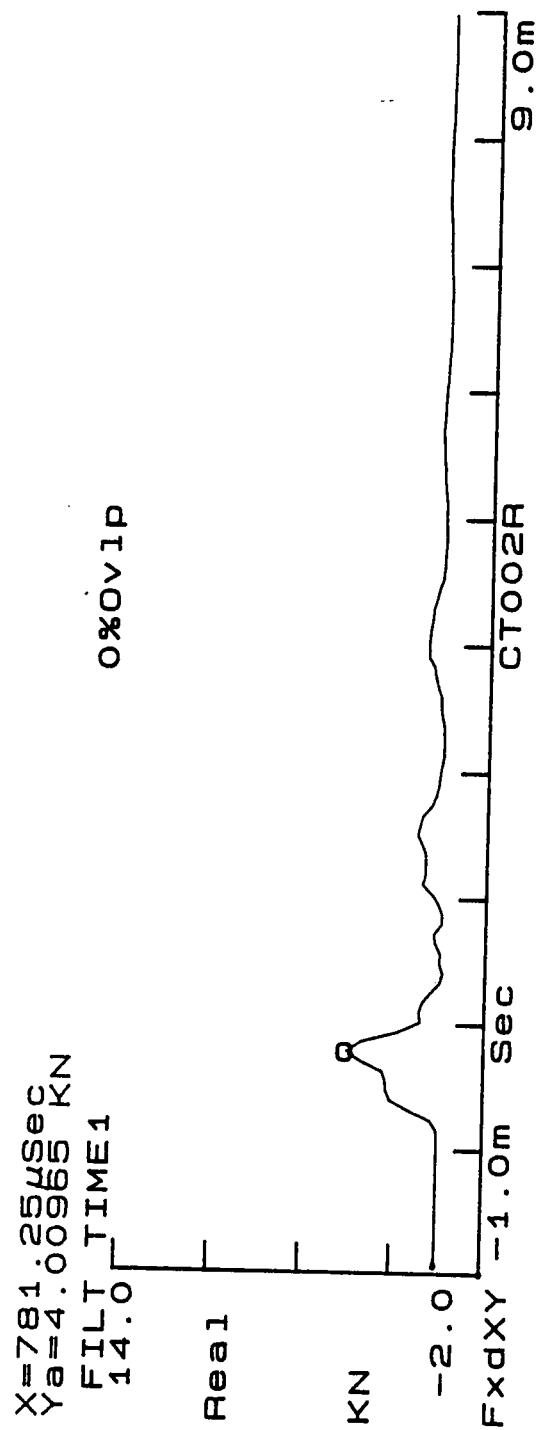


Figure 42. Fracture force data for tibia specimen 002R

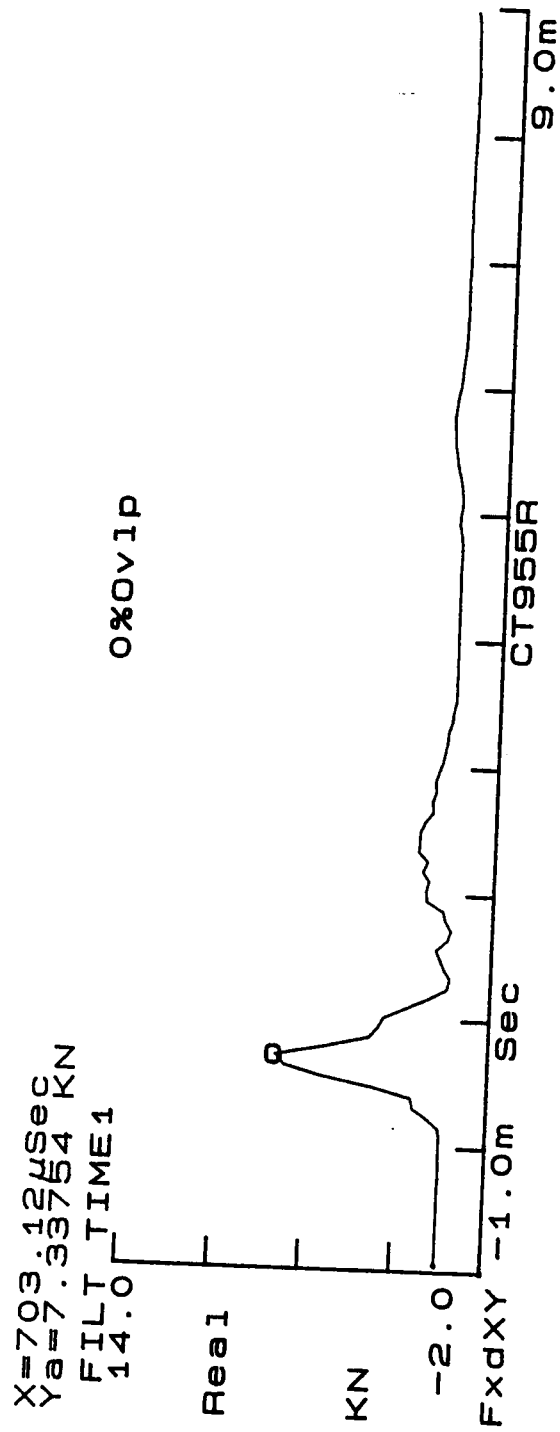


Figure 43. Fracture force data for tibia specimen 955R

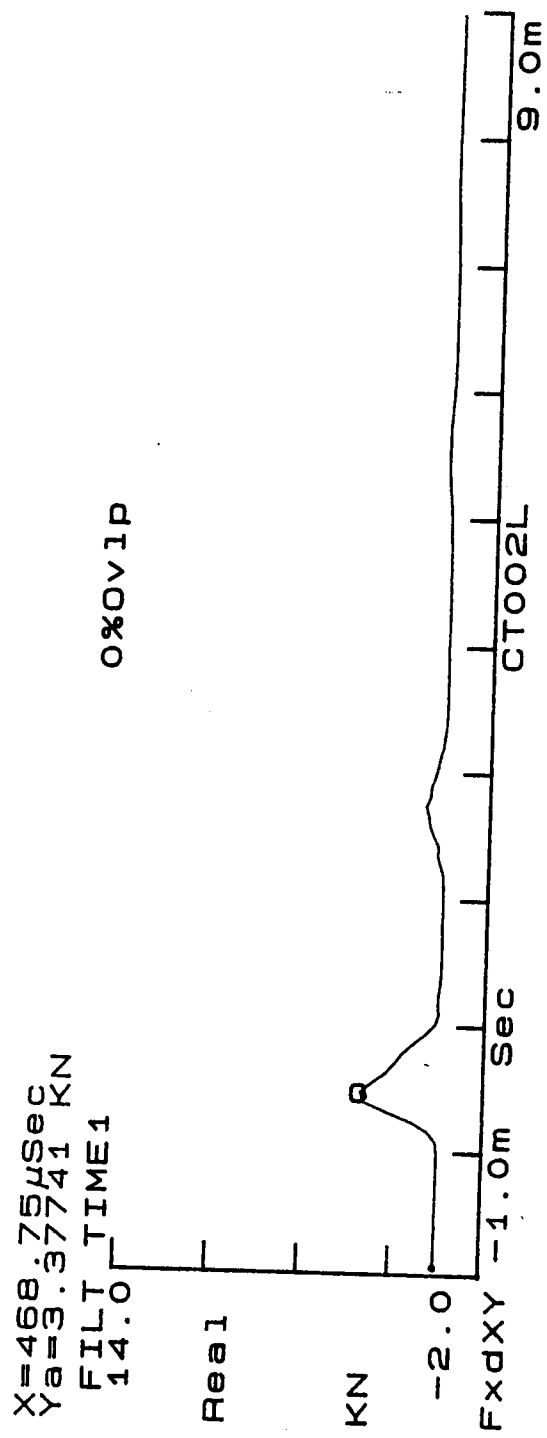


Figure 44. Fracture force data for tibia specimen 002L

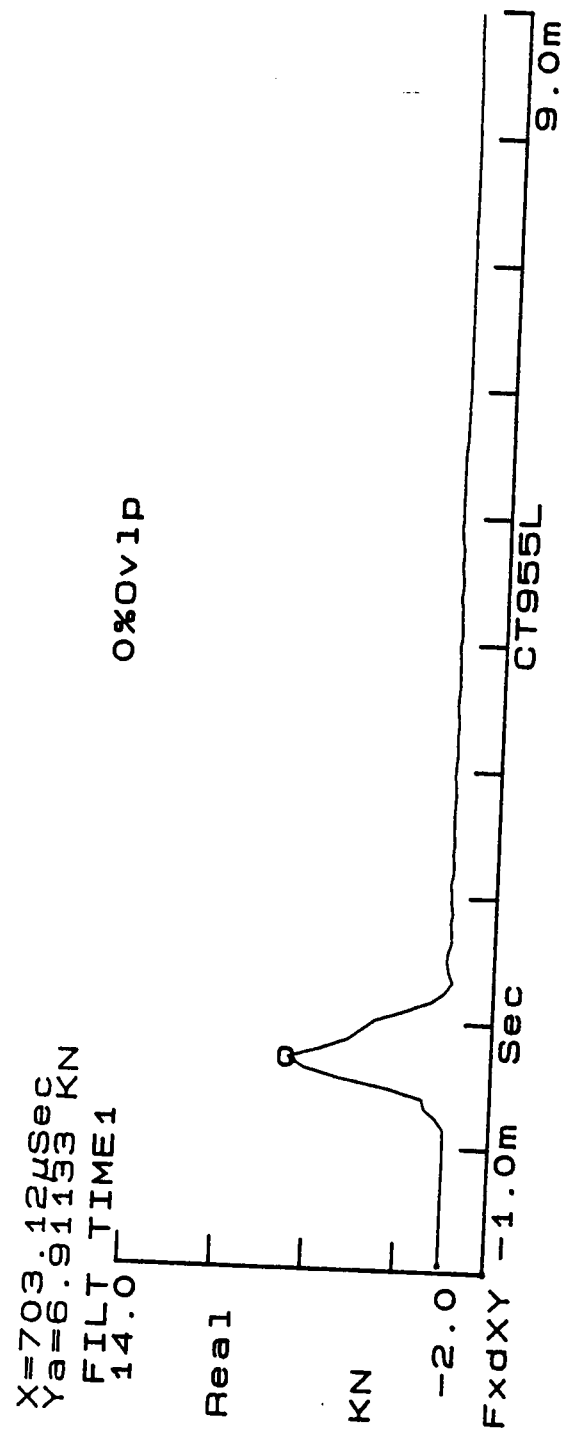


Figure 45. Fracture force data for tibia specimen 955L

Table 6
Fracture Force Data

Tibia Specimen	Sex	Age	Fracture Force
Number	Male / Female		lbf
	M / F		
977L	F	65	1299.17
991R	F	42	1314.89
971R	F	72	1397.48
926L	F	85	789.08
977R	F	65	1355.37
929L	M	72	1573.66
984R	M	80	1577.78
002R	F	53	899.00
955R	M	67	1649.54
002L	F	53	759.18
955L	M	67	1553.73
Average		Fracture Force	1288.08

modes of fracture in the tibia. One mode is denoted by contours having a value of one and show up as pink regions on the contour plots. This mode of fracture is caused by the effective stress of the element becoming larger than the fracture stress (yield and ultimate stress for dry, embalmed tibia). The second mode of failure is denoted by a value of two and is signified by light blue regions. These elements have failed due to the mean stress becoming too tensile. These two modes of failure, as indicated in the contour plots, denote elements in the model that are damaged and have failed structurally. Several views of the tibia are examined for fracture analysis; some of these views can not be examined experimentally. The speed at which the fracture propagates makes it difficult to film with high speed film processes unless the exact region of the fracture is known so as to concentrate the filming in this region.

Figure 46 and 47 show contour fracture plots of the tibia diaphysis from the medial view of the tibia. The point of impact occurs at the middle front edge of the tibia which corresponds to the anterior side of the bone. The bumper is moving in the negative "Y" direction. Each figure shows four contour plots at different time intervals. The progression of time for the contour plots are read from left to right and from top to bottom. Figures 46 and 47 comprise a full time run for the finite element model. All contour plots are presented in the same manner as figure 46 and 47. In figure 46 it can be seen that at the first time interval the crack initiates at the anterior side where impact has occurred. In the second time interval, the anterior fracture has started to propagate longitudinally as well as posteriorly toward the proximal and distal ends. In addition, a new crack has initiated in the middle of the midshaft and has formed a wedge shape

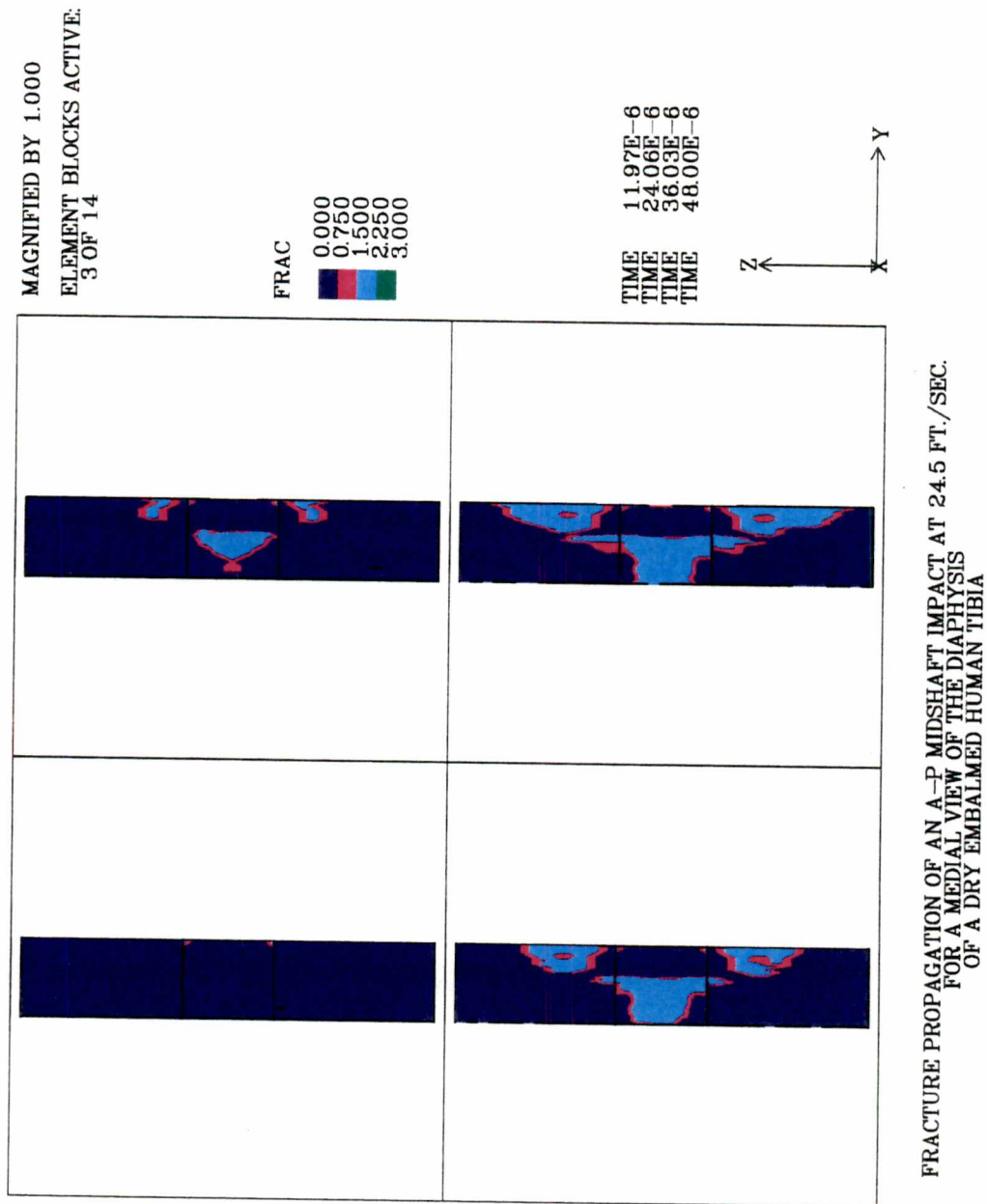


Figure 46. Fracture propagation contour plot from 11.97 to 48.0 microseconds of human tibia diaphysis viewed from the medial side

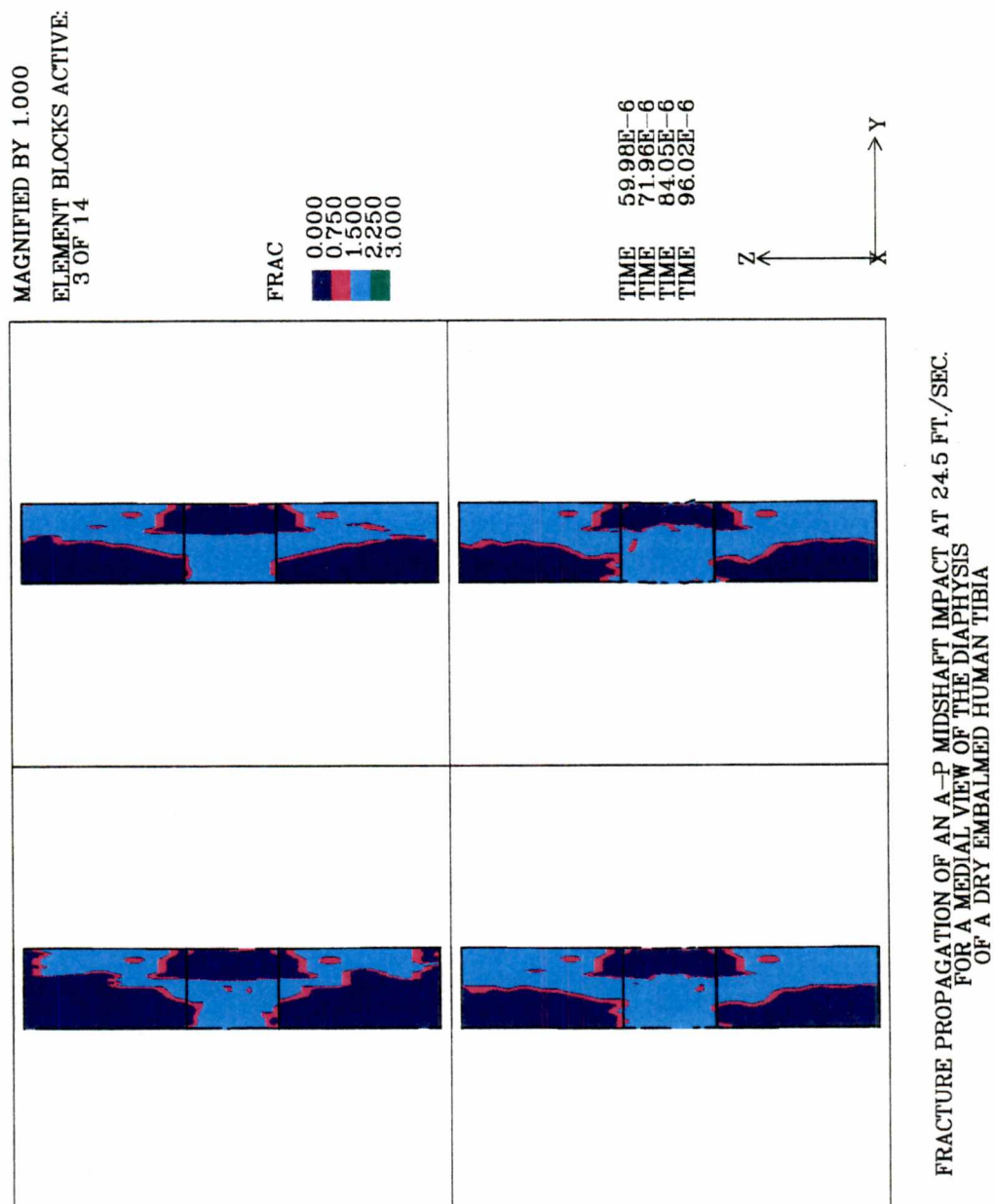


Figure 47. Fracture propagation contour plot from 59.98 to 96.02 micro-seconds of human tibia diaphysis viewed from the medial side

pointing in the posterior direction. As time progresses, the anterior fracture has continued propagating longitudinally but appears to have slowed in its progression posteriorly. Meanwhile, the crack that started in the middle of the midshaft has now propagated posteriorly, and the anterior edge of this fracture is propagating longitudinally. Near the end of the time run, the anterior fracture has completely involved the whole anterior side of the proximal and distal regions. The middle midshaft crack has completely involved the whole midshaft region from the middle of the midshaft to the posterior end. Also, this crack has propagated longitudinally to connect with the anterior crack. Several longitudinal intact areas of bone remain; thus, the trend of this crack from the medial view is symbolic of a longitudinal splintering effect. It is interesting to note that the majority of the failed regions occurred by the second mode of failure and is outlined by the first mode of failure.

Figures 48 and 49 show the fracture contour plots from an anterior view. In other words, this view is from the impact bumper's perspective. Notice that at the start of the time run, the fracture occurs in the first mode of failure and also occurs at the point of impact. As the time run progresses, this anterior fracture propagates longitudinally; and the second mode of failure becomes apparent. A new fracture appears on each side of the tibia as well. The fracture seen here is actually the middle wedge shaped fracture seen in the previous medial view figure 46. As the end of the time run approaches, the fracture has completely involved the anterior side of the tibia both proximally and distally as well as at the midshaft. It is interesting to note that at the point of impact, the first mode of failure is dominant throughout the time run. However, as the

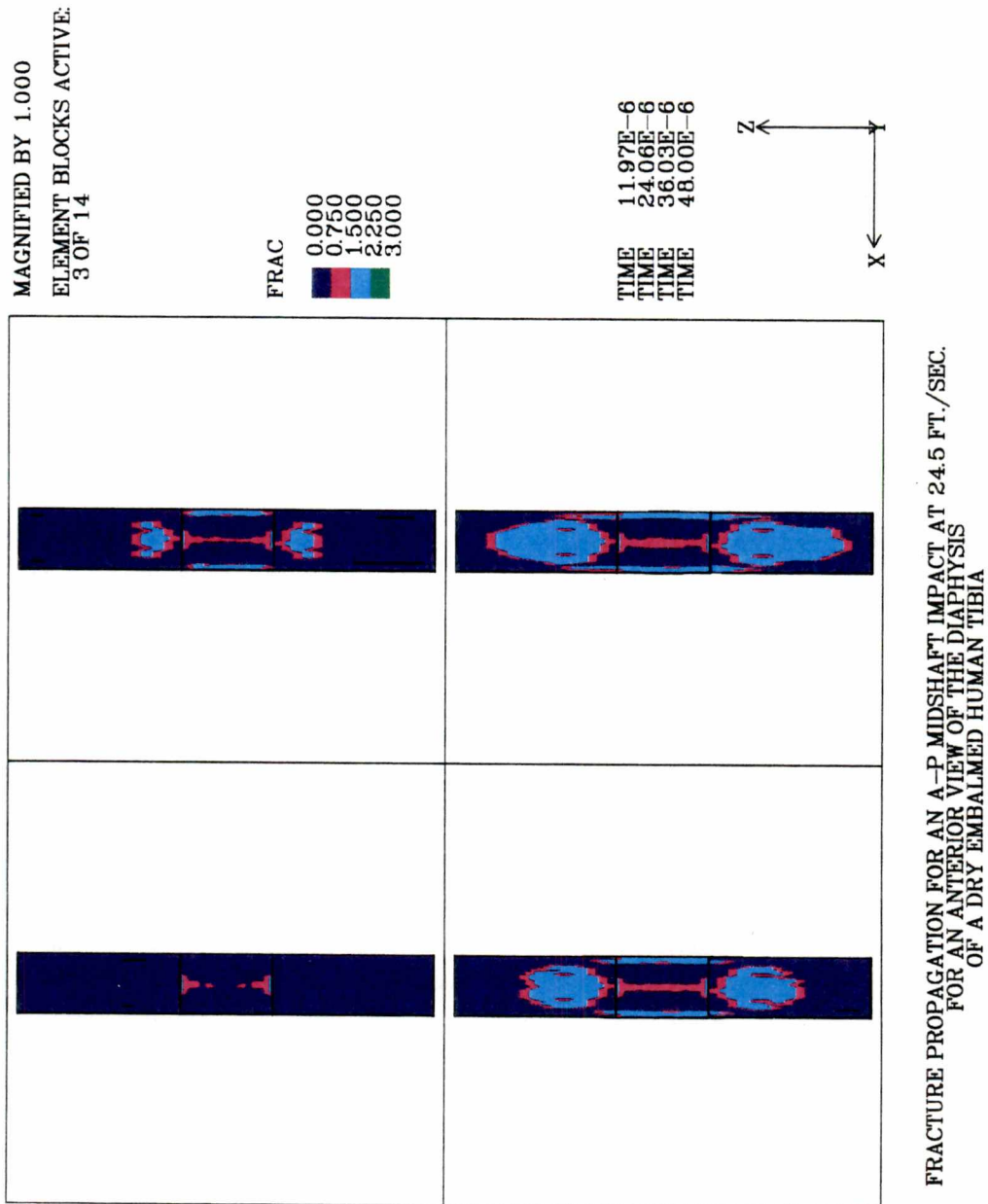


Figure 48. Fracture propagation contour plot from 11.97 to 48.0 micro-seconds of human tibia diaphysis viewed from the anterior side

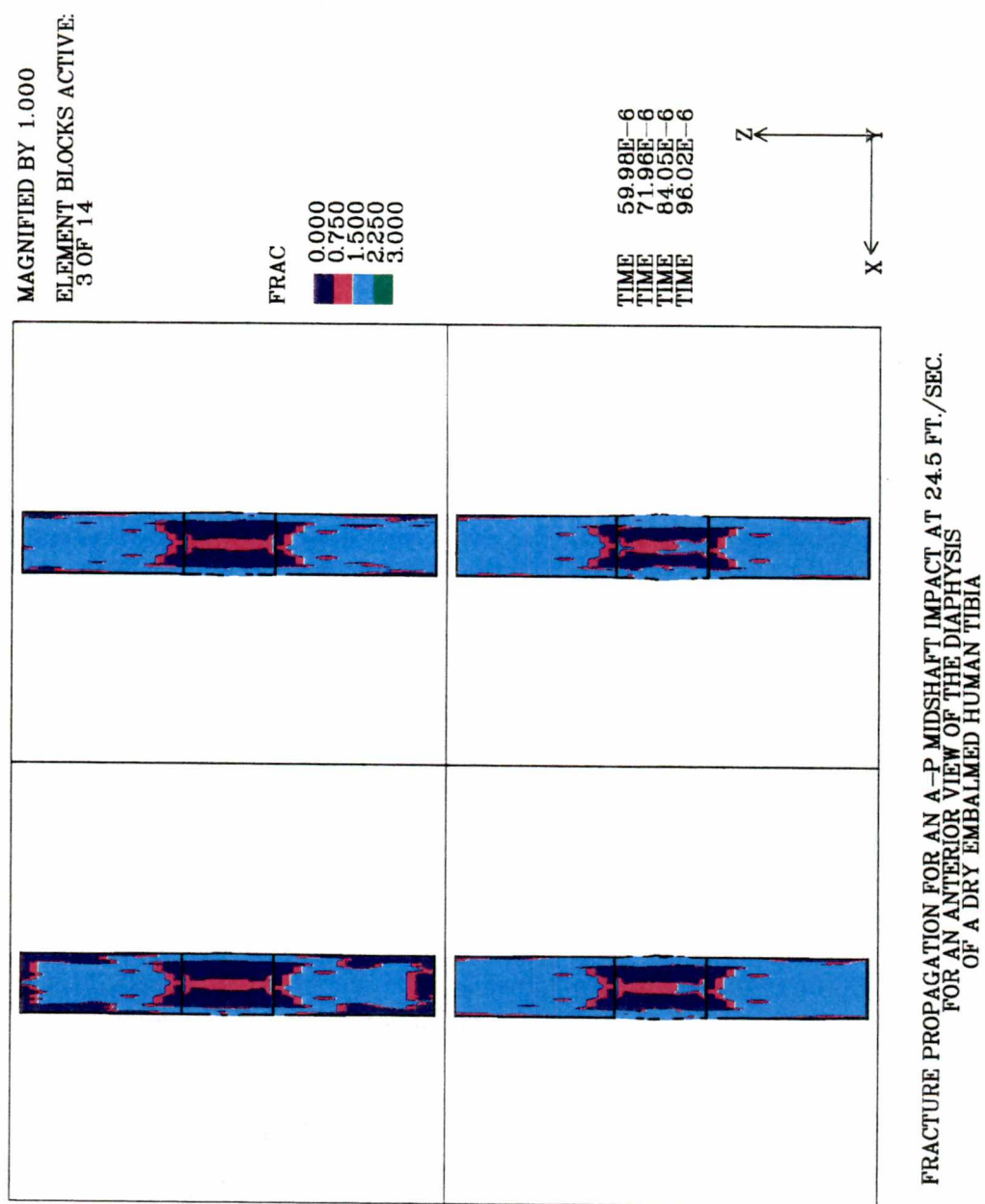


Figure 49. Fracture propagation contour plot from 59.98 to 96.02 micro-seconds of human tibia diaphysis viewed from the anterior side

crack propagated longitudinally the second mode of failure is dominant. These contour plots show a consistency between the medial and anterior views. Again, the longitudinal intact pieces appear indicating a splintering effect.

Figures 50 and 51 show the posterior (behind the bone) view of the crack propagation. From these plots it can be seen that the crack wraps around the bone until it has completely involved the posterior midshaft region. The crack does not propagate much in the longitudinal direction but stays mainly in the midshaft region. Again the primary mode of failure is a result of the mean stress becoming too tensile. These figures also remain consistent with the previous two sets of figures (figure 46, 47, 48, 49).

Experimental investigation of the crack propagation in the interior of the bone can not be accomplished without doing structural damage to the bone. However, the finite element method allows for analysis without structural damage. Figures 52 and 53 show a fracture contour plot of half of the diaphysis of the tibia. The bone has been sectioned longitudinally down the long axis of the bone, and the medial view is shown. It is possible, with this view, to examine the fracture as it propagates through the cortical bone thickness and to look at the interior wall of the tibia. The long, black lines running the length of the shaft both anteriorly and posteriorly signify the end of the cortical bone thickness in those regions. Again, the impact point is moving in the negative "Y" direction and occurs at the anterior midshaft edge. Notice that at the beginning of the time run, the fracture occurs at the point of impact along the front anterior edge as well as at the rear of the cortical bone section for the midshaft region.

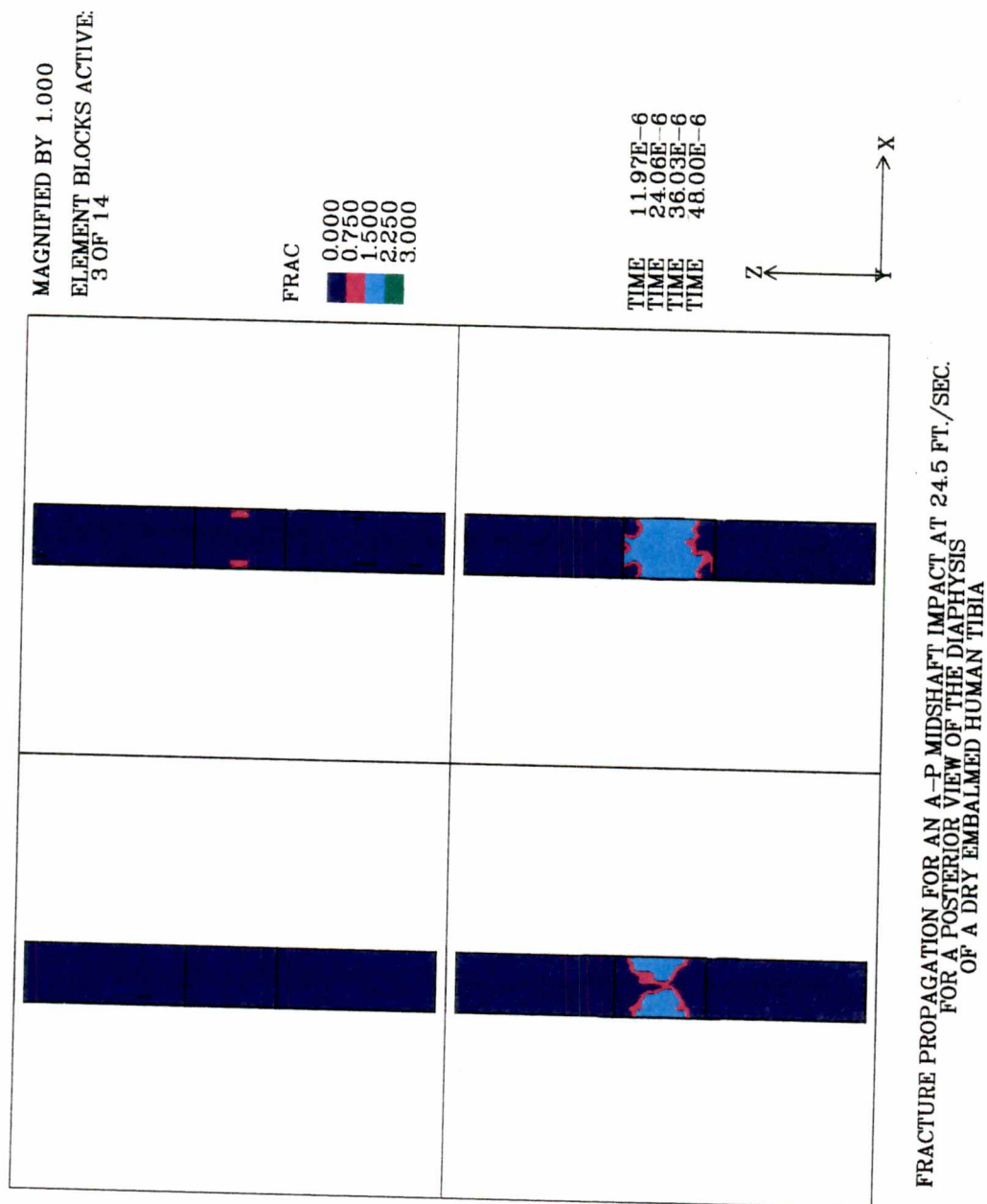


Figure 50. Fracture propagation contour plot from 11.97 to 48.0 micro-seconds of human tibia diaphysis viewed from the posterior side

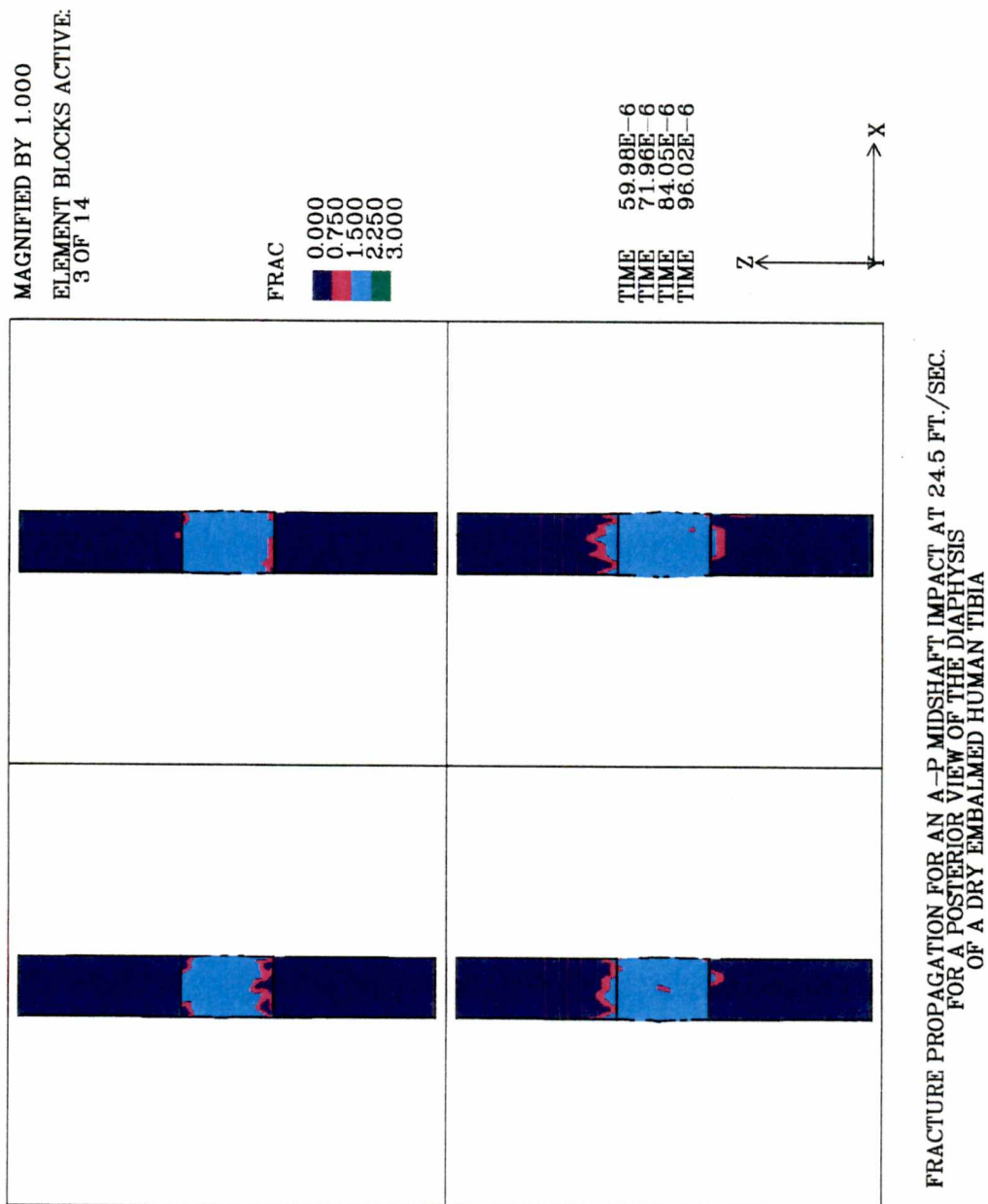


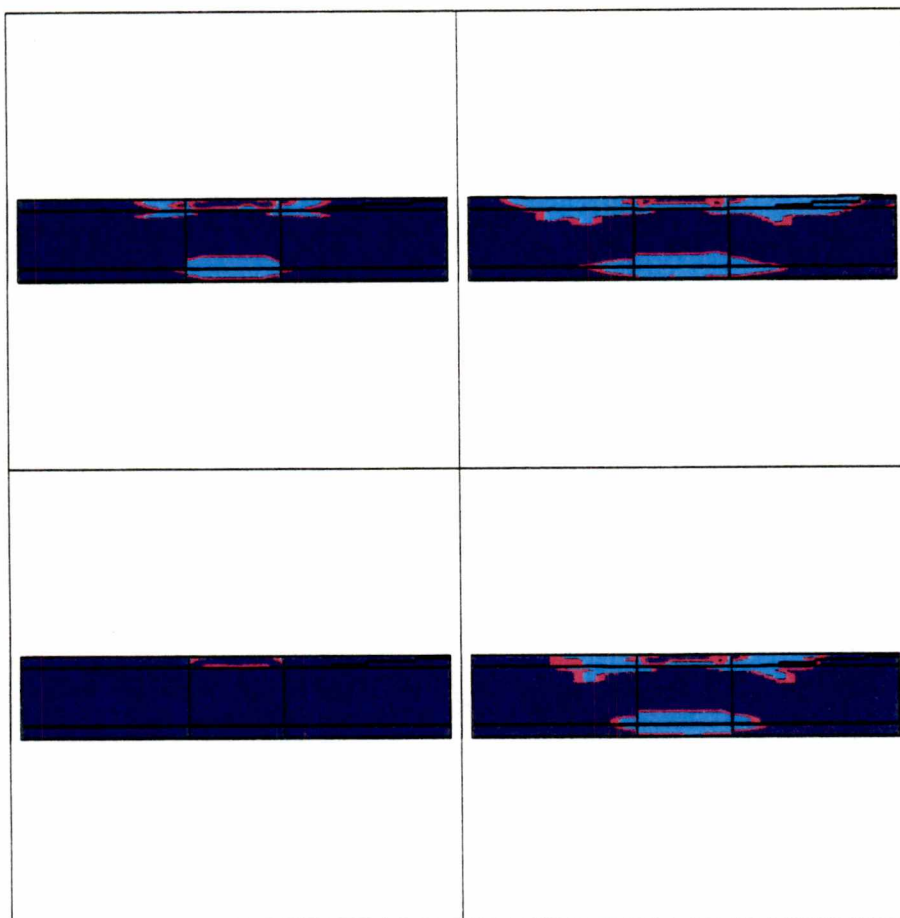
Figure 51. Fracture propagation contour plot from 59.98 to 96.02 micro-seconds of human tibia diaphysis viewed from the posterior side

MAGNIFIED BY 1.000
ELEMENT BLOCKS ACTIVE:
3 OF 14

FRAC
0.000
0.750
1.500
2.250
3.000

TIME 11.97E-6
TIME 24.06E-6
TIME 36.03E-6
TIME 48.00E-6

Z
Y



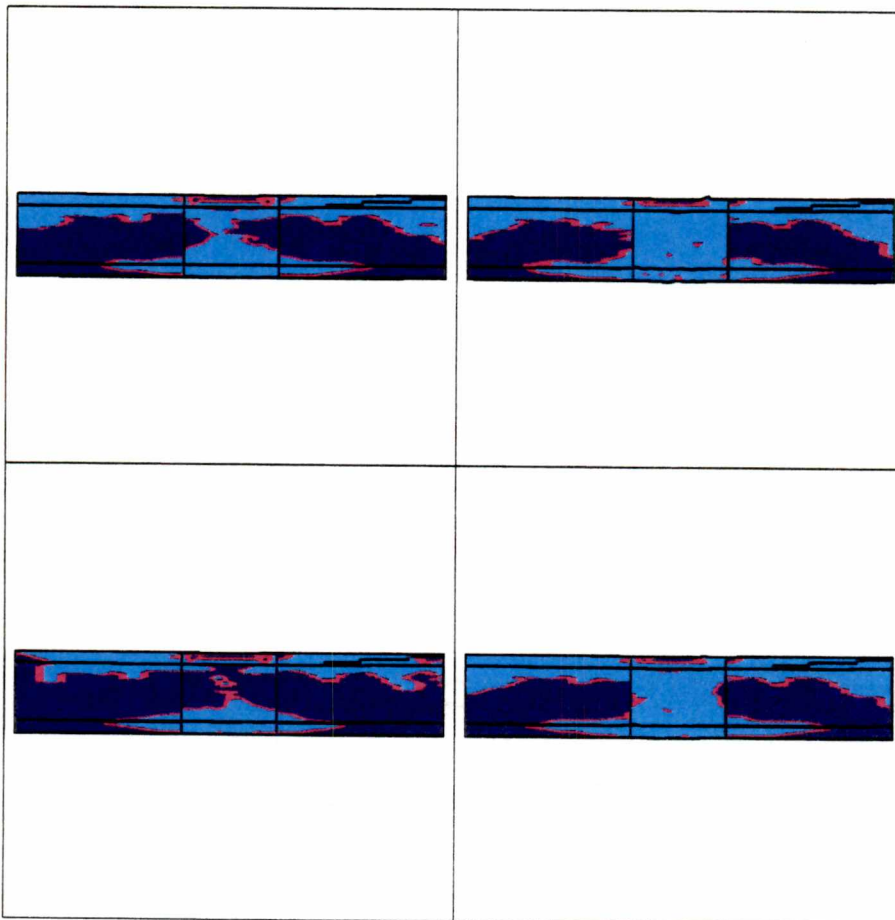
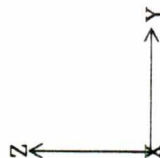
FRACTURE PROPAGATION FOR AN A-P MIDSHAFT IMPACT AT 24.5 FT./SEC.
FOR A MEDIAL VIEW OF A LONGITUDINAL CROSS-SECTION
OF A DRY EMBALMED HUMAN TIBIA

Figure 52. Fracture propagation contour plot from 11.97 to 48.0 micro-seconds of a longitudinal cross-section of a human tibia diaphysis viewed from the medial side

MAGNIFIED BY 1.000
ELEMENT BLOCKS ACTIVE:
3 OF 14

FRAC
0.000
0.750
1.500
2.250
3.000

TIME 59.98E-6
TIME 71.96E-6
TIME 84.05E-6
TIME 96.02E-6



FRACTURE PROPAGATION FOR AN A-P MIDSHAFT IMPACT AT 24.5 FT./SEC.
FOR A MEDIAL VIEW OF A LONGITUDINAL CROSS-SECTION
OF A DRY EMBALMED HUMAN TIBIA

Figure 53. Fracture propagation contour plot from 59.98 to 96.02 micro-seconds of a longitudinal cross-section of a human tibia diaphysis viewed from the medial side

Both fractures are in mode one failure. At the second time interval, the anterior midshaft fracture is propagating longitudinally with the fracture consisting primarily of mode two failure. This is consistent with the normal medial view seen in figures 46 and 47. However, a crack in mode two failure has occurred at the interior, posterior, midshaft, cortical bone region but has not entirely propagated through the cortical bone. This differs from what is seen in the medial view of figures 46 and 47. As time progresses to the end of the time run, the anterior crack has propagated longitudinally to involve the anterior proximal and distal regions. The posterior crack not only propagates anteriorly in the midshaft region but also propagates longitudinally to involve most of the proximal and distal interior regions. However, this posterior crack does not entirely propagate through the cortical bone in the proximal and distal regions. Therefore, it is not seen on the exterior medial or posterior views in figures 46, 47, 50 and 51.

Finally, a top cross-section of the midshaft of the tibia is seen in figures 54 and 55. This corresponds to a view as seen from sitting above the bone and looking down the long axis of the tibia. This view emphasizes the crack propagation in the cortical bone. Again, the impact bumper is moving in the negative "Y" direction. It can be seen that at the first time step, fracture has occurred at the anterior exterior region of the bone in mode one failure. The interior anterior region of the cortical bone is also fractured in both mode one and mode two failure. As time progresses, the anterior fracture first propagates through the entire cortical bone in mode one failure. Then mode two failure occurs throughout the anterior cortical bone although it is not as dominant as is normally seen. A fracture also

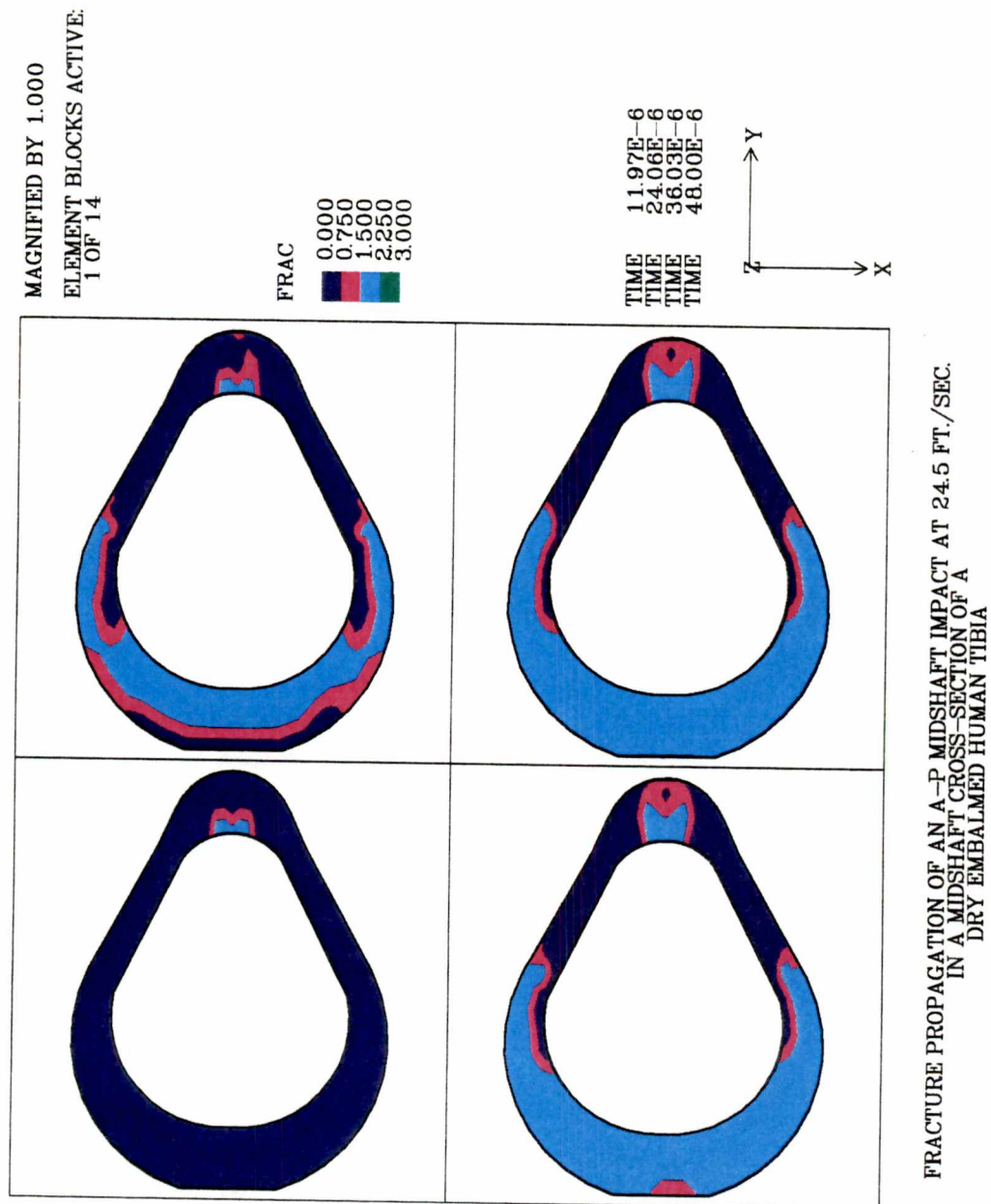


Figure 54. Fracture propagation contour plot from 11.97 to 48.0 micro-seconds of a midshaft cross-section of a human tibia viewed axially

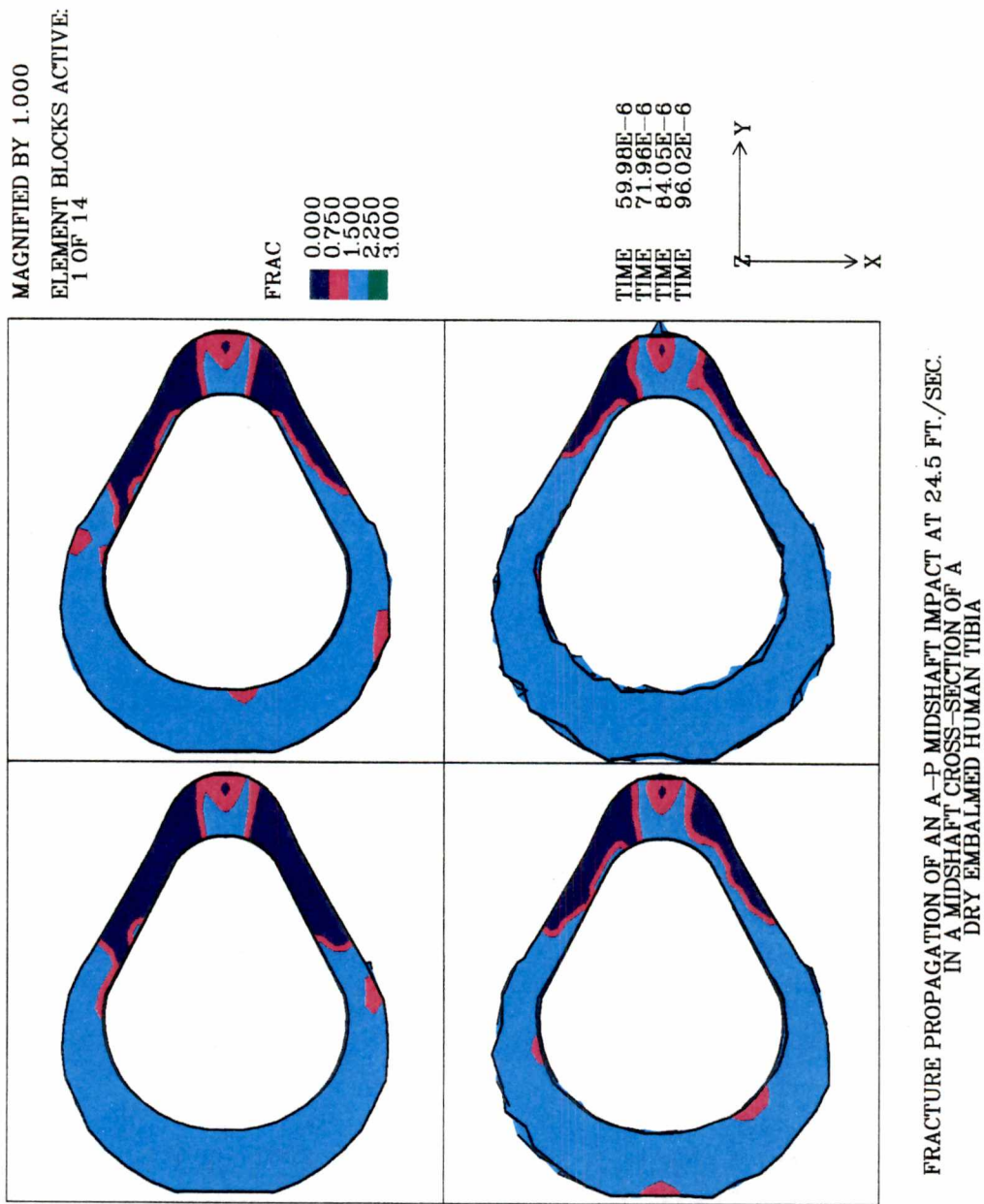


Figure 55. Fracture propagation contour plot from 59.98 to 96.02 micro-seconds of a midshaft cross-section of a human tibia viewed axially

appears posteriorly primarily in mode two failure. This fracture starts along the interior posterior region of the cortical bone and propagates both posteriorly to the exterior regions of the bone as well as wrapping around the bone to the anterior direction. It can be seen that the fracture never completely wraps around the entire bone to the anterior region which corresponds to the longitudinal intact bone segments.

2. Stress Analysis at Tibial Midshaft

Three types of stress were examined for a cross-section of the midshaft of the tibia. The effective stress, stress in the "YY" direction (direction of bumper movement) and stress in the "XX" direction (orthogonal to the bumper movement) were all examined in hopes of better understanding the crack propagation.

Figures 56, and 57 show the contour plots of the effective stress on the tibia cross-section. Notice that at the beginning of the time, the posterior end of the bone sees no effective stress. However, the anterior end sees relatively high stress. As time progresses, the anterior end of the bone feels a reduction of stress. This can be attributed to the fracture occurring at the anterior midshaft. As the bone fails for the effective stress fracture mode and the elements have undergone failure, the effective stress is reduced by the drop in the yield stress specified in the constitutive model. When the anterior end is seeing a reduction in stress, the posterior end is starting to build up stress on the order of about 17,000 psi. This build up continues up to the last time frame when a reduction in effective stress is, once again, seen. This is attributed to the bone's posterior elements starting to see fracture from the effective stress mode of failure. This can be seen more readily in the cross-sectional fracture

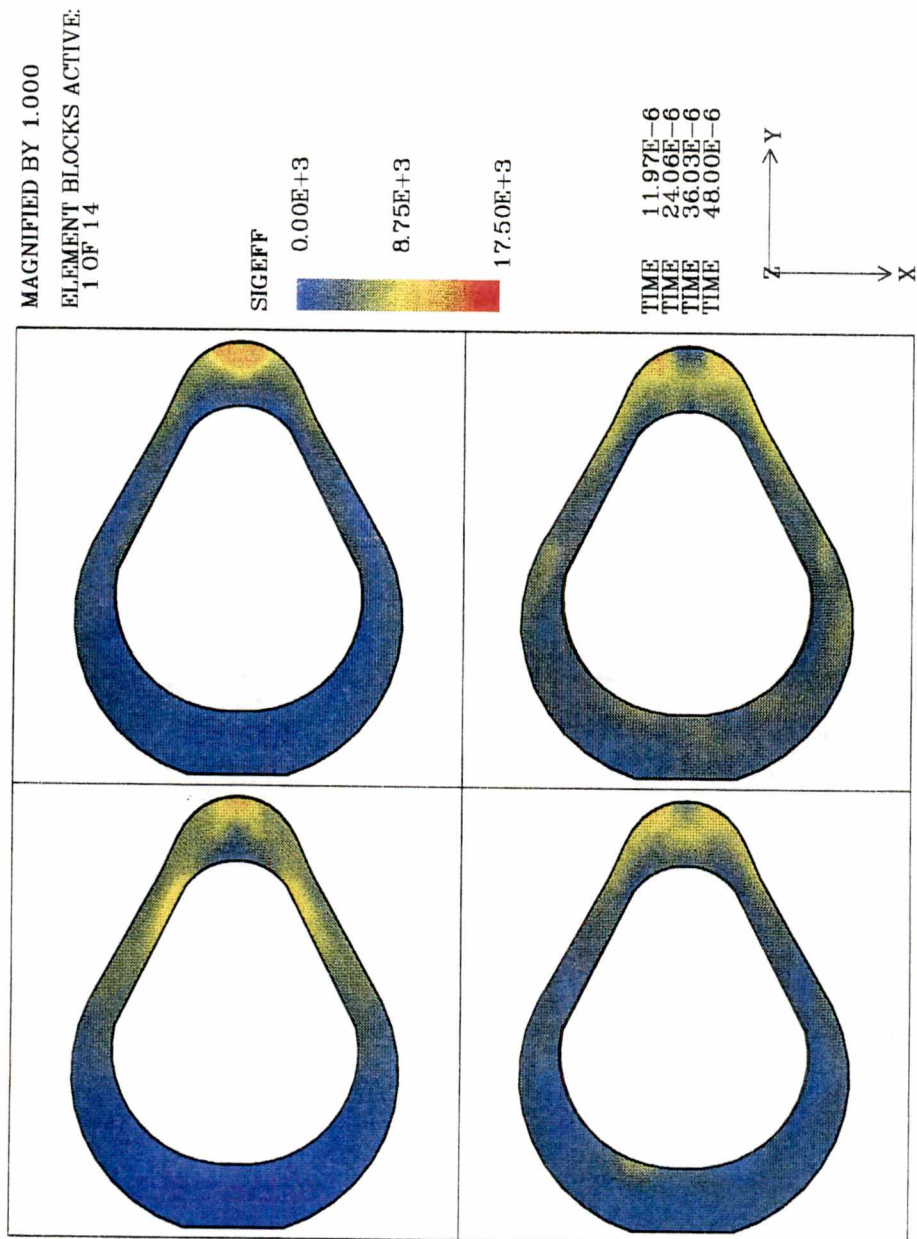


Figure 56. Effective stress contour plot from 11.97 to 48.0 microseconds of a midshaft cross-section of a human tibia viewed axially

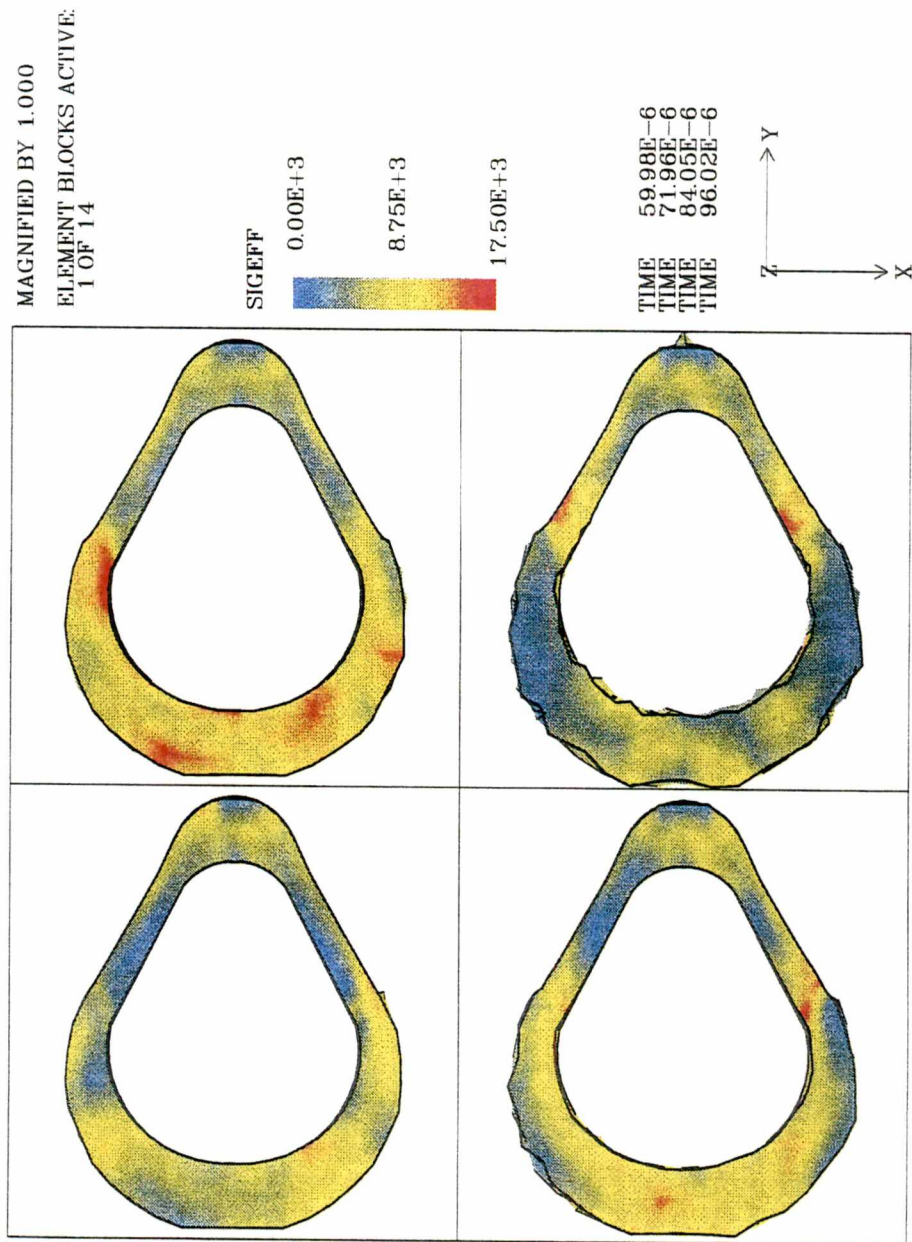


Figure 57. Effective stress contour plot from 59.98 to 96.02 microseconds of a midshaft cross-section of a human tibia viewed from the posterior side

picture (figures 54 and 55) where at time frame of 71.95 and 84.05 microseconds, the posterior elements are starting to see fracture from the effective stress failure mode. At the last time step, figure 55 shows the bone's posterior elements in complete fracture from the mean stress failure. This correlates with the reduction of effective stress seen for the posterior end of the bone at the last time step in figure 57. Also, it appears as if the bone may be in a chaotic stress state at this last time step. This is due to the large mesh distortion which will affect the accuracy of the answers at this point.

Figures 58 and 59 show the contour plots of the stress in the "YY" direction--direction of bumper movement. The plots show a compressive stress of about -15,500 psi at the point of impact on the exterior anterior side at the beginning time steps. The interior anterior cortical bone shows a tendency to build up tensile stresses. This may indicate a bending mode in the cortical bone at the region of impact. As time progresses, the stress in the anterior cortical bone reduces in stress and the posterior cortical bone sees an increase in stress in some places on the order of 5,000 psi. It can be seen that the stress, especially on the posterior side, appears in a chaotic state as the mesh becomes badly distorted.

Figures 60, and 61 show contour plots of the stress in the "XX" direction (orthogonal to the bumper movement). The stress slightly reduces in magnitude as time progresses on the exterior anterior cortical bone. The interior anterior cortical bone reaches slightly tensile stresses and remains fairly constant throughout time. The posterior cortical bone builds slightly compressive stresses and, once again, starts into a chaotic state as the mesh becomes badly distorted. Overall, the stress in the "XX"

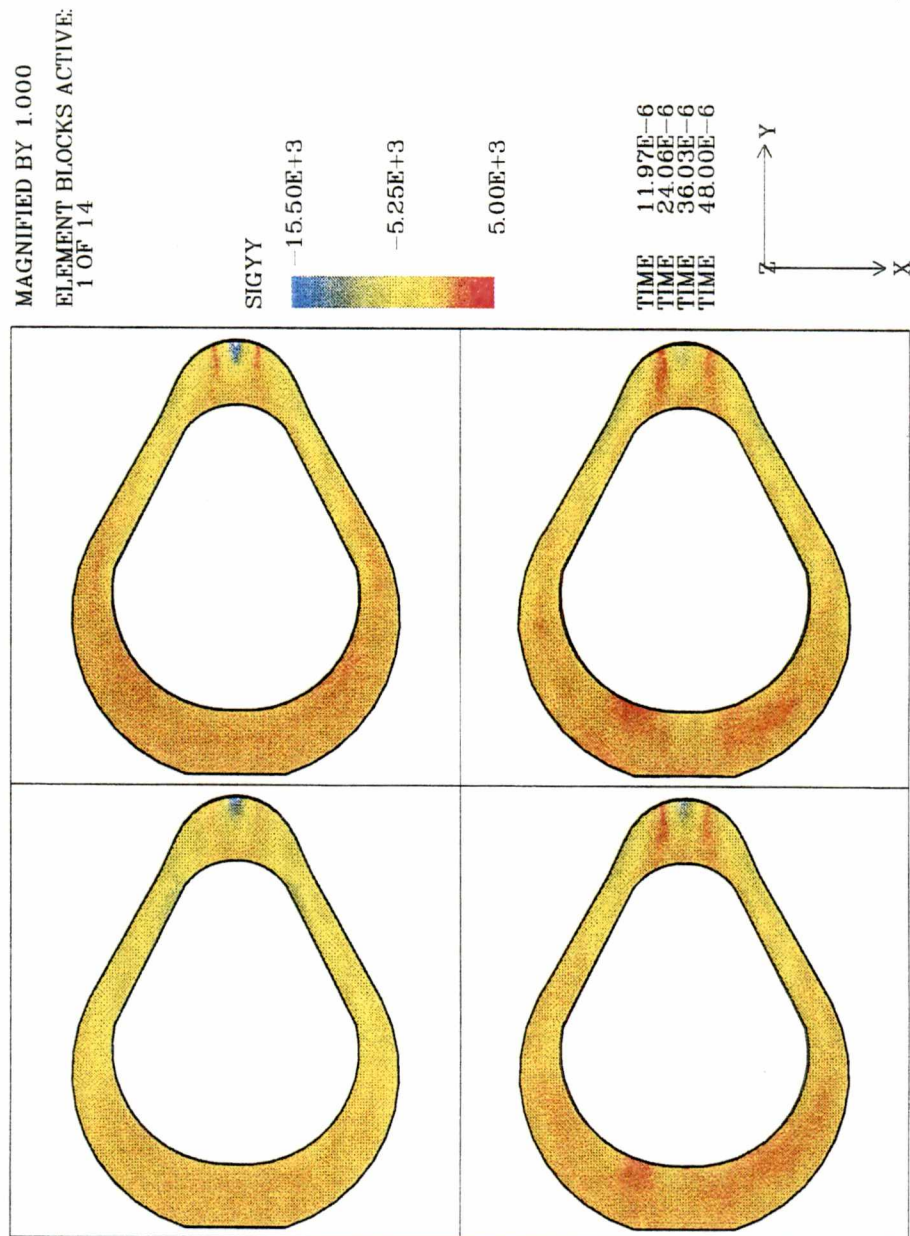


Figure 58. Contour plot of stress in the "YY" direction from 11.97 to 48.0 microseconds of a midshaft cross-section of a human tibia viewed axially

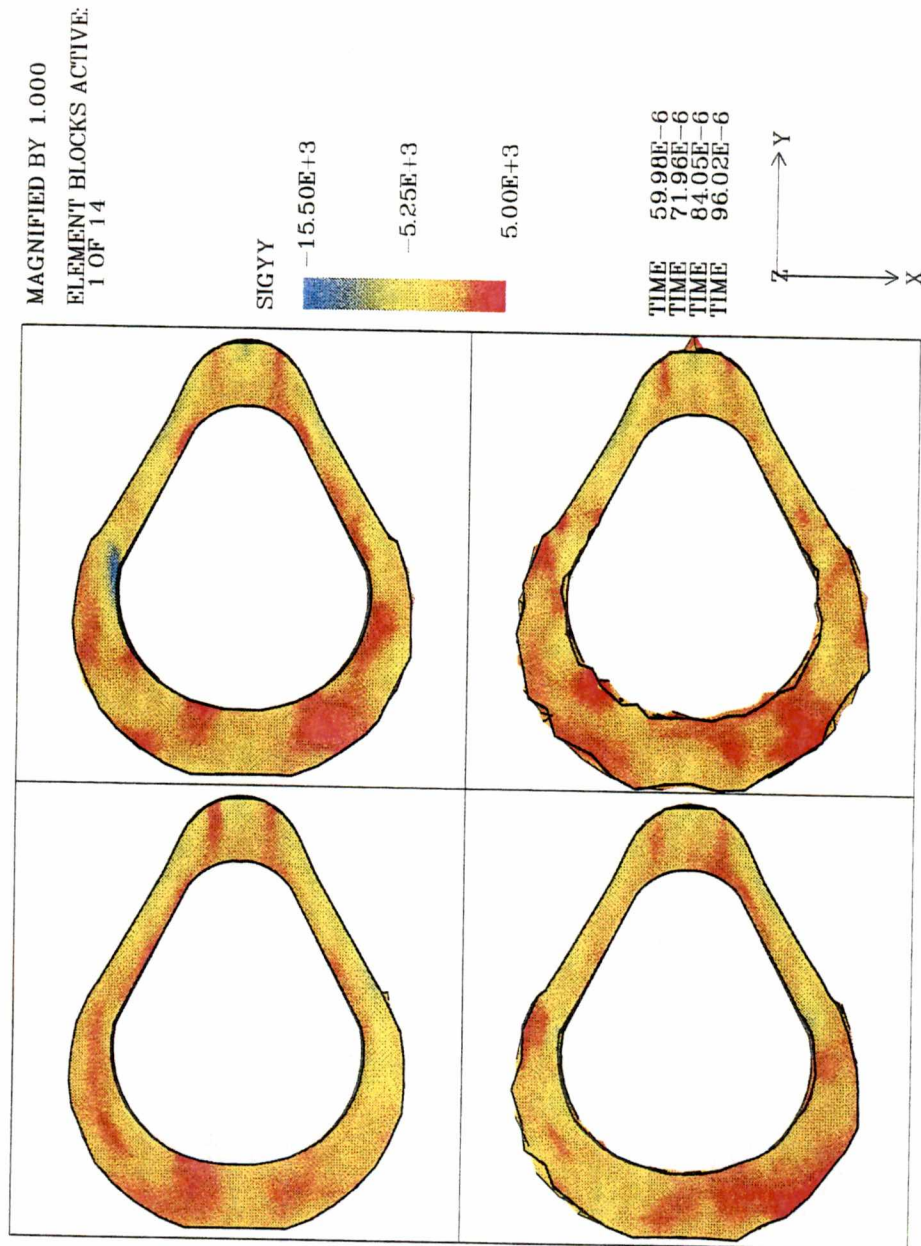


Figure 59. Contour plot of stress in the "YY" direction from 59.98 to 96.02 microseconds of a midshaft cross-section of a human tibia viewed from the posterior side

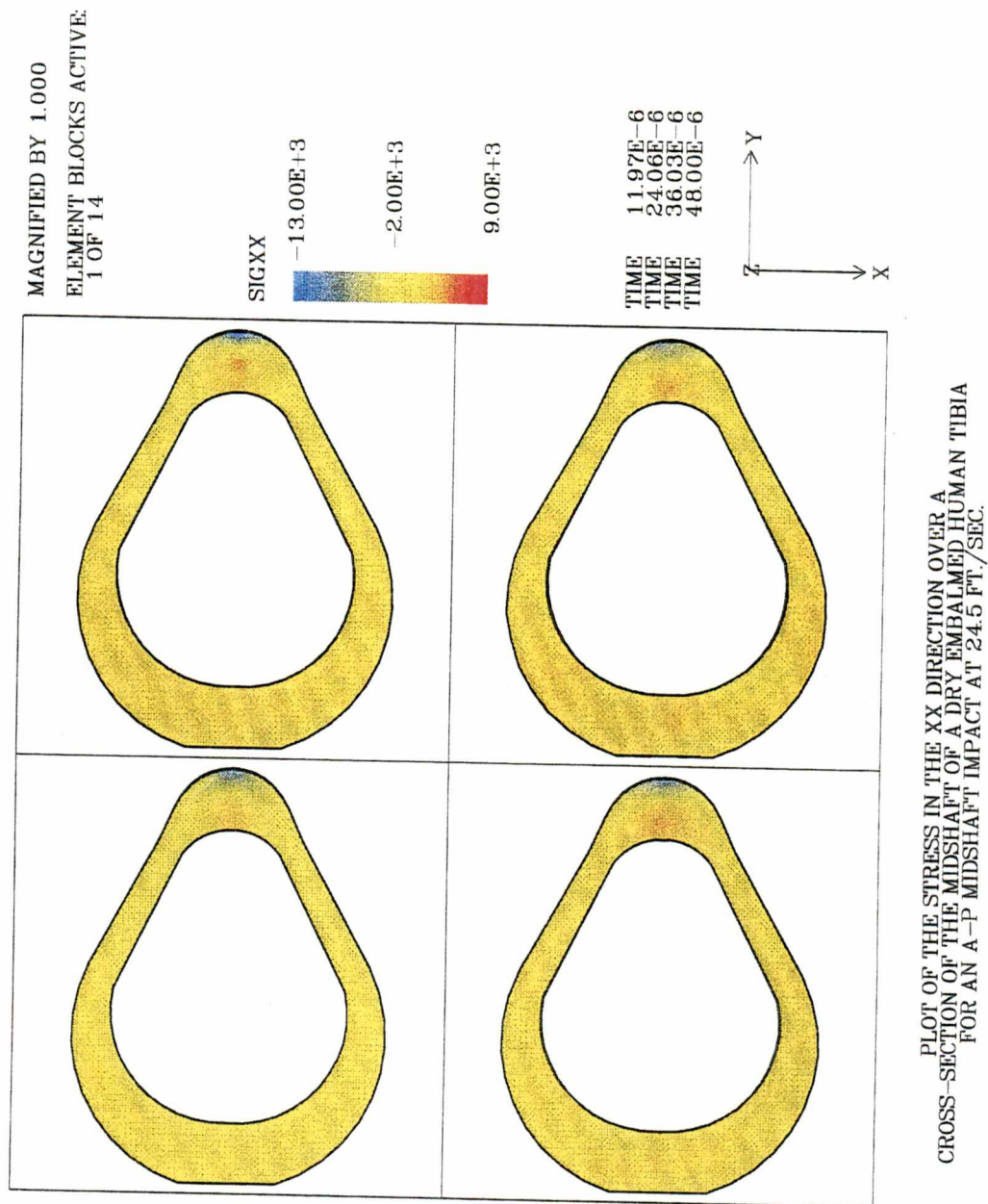


Figure 60. Contour plot of stress in the "XX" direction from 11.97 to 48.0 microseconds of a midshaft cross-section of a human tibia viewed axially

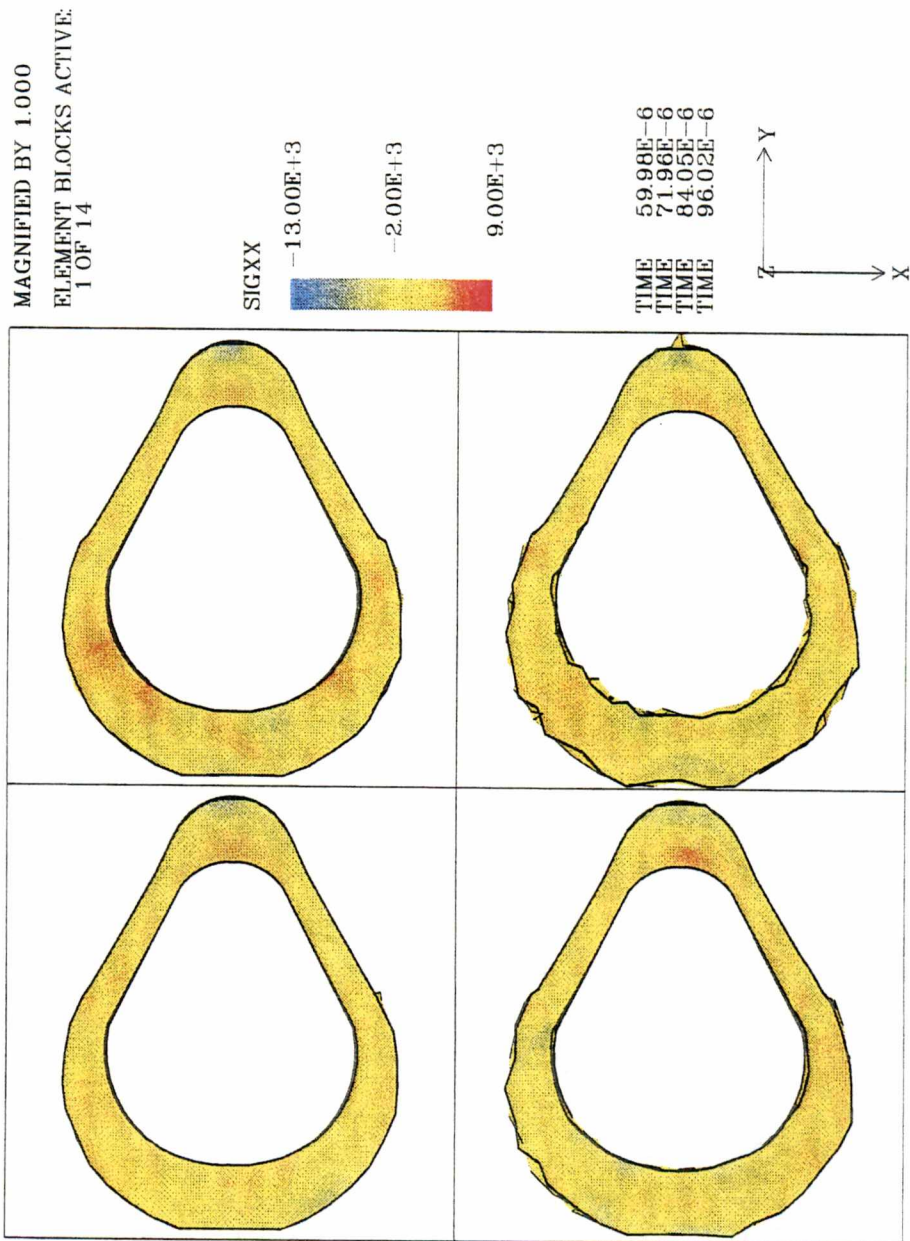


Figure 61. Contour plot of stress in the "XX" direction from 59.98 to 96.02 microseconds of a midshaft cross-section of a human tibia viewed from the posterior side

direction remains fairly constant over time.

3. Fracture Force Analysis

The fracture force from the finite element model was measured by a modeled load cell instrumenting the bumper. The load cell was modeled by a single H8 element attached to the front of the bumper and extending behind the bumper. The load cell was given a prescribed velocity of 24.5 ft./sec. in the negative "Y" direction. The load cell measured the compressive force required to push the bumper through the bone by measuring the stress in the "YY" direction and multiplying by the area of the front face of the load cell in contact with the bumper (8 in.). This simulates what the force transducer in the experimental testing measures.

Figure 62 shows the fracture force plot obtained from the load cell element. Notice the oscillations in the plot and the very large oscillation that occurs at about 0.028 milliseconds. The oscillations occurring before the 0.028 millisecond mark is actual vibration in the modeled system. However, the large oscillation at 0.028 milliseconds is caused by an error either in the finite element source code or in the Sun operating system. The data from the 0.028 millisecond mark to the end of the time run is ignored, and only the data from the beginning of the time run to the point of trouble is analyzed. This error occurs when a restart file is written for the finite element model. The restart file allows for the code to be halted as well as for the data to be analyzed or stored for future use and then the code can be restarted from the stopping point. Writing a restart file could not be avoided if an adequate data sample rate was to be attained. This was due to the lack of disk space available on the computer system. After

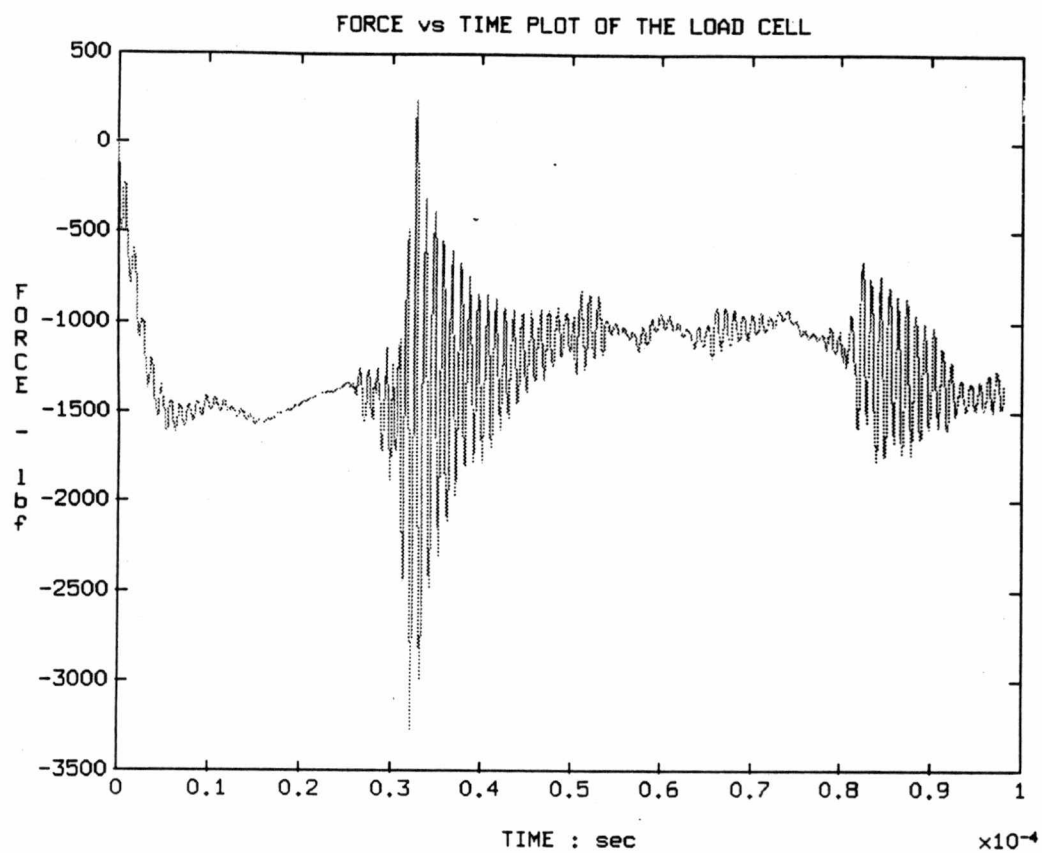


Figure 62. Finite element fracture force versus time plot for 0.0 to 0.1 milli-seconds

each stop, the data had to be removed or stored for later use and the computational model started again. It is important to note that the restart file was only written for analysis of the fracture force data. It was not necessary to use a restart file in analyzing the crack propagation data. Therefore, no extraneous errors should have been introduced in this data.

Looking at the time run from 0.0 to 0.028 milliseconds (figure 63), the data shows a trend in the fracture force to gently peak at about 1600 lbf before reversing direction and begin to reduce in magnitude. The oscillations seen in this data are actual vibrations in the modeled system. Efforts were attempted to identify these modes of vibration so that future finite element models could filter these vibrations or devise a better model to reduce these oscillations. To help in determining the vibration frequency present in the data, the fracture force data was analyzed by plotting the spectral density versus frequency (figure 64). This data shows that two different vibrations occur in the model. These predominant vibrations are represented as the two largest peaks on the figure. The first peak occurs at about 100,000 Hz and the second peak, which is the largest, occurs at about 1.0 MHz.

In the vibration identification effort, several finite element models and theoretical calculations were performed on each of the components of the bumper, load cell, and bone configuration. Table 7 shows the results of these calculations. The load cell and bumper component vibrations were measured computationally in each of its three principal axes. The bumper and load cell configuration were also tested for its natural frequency as one component. This is present in the finite element model in the direction of its movement. The bumper was also analyzed for its bending frequency

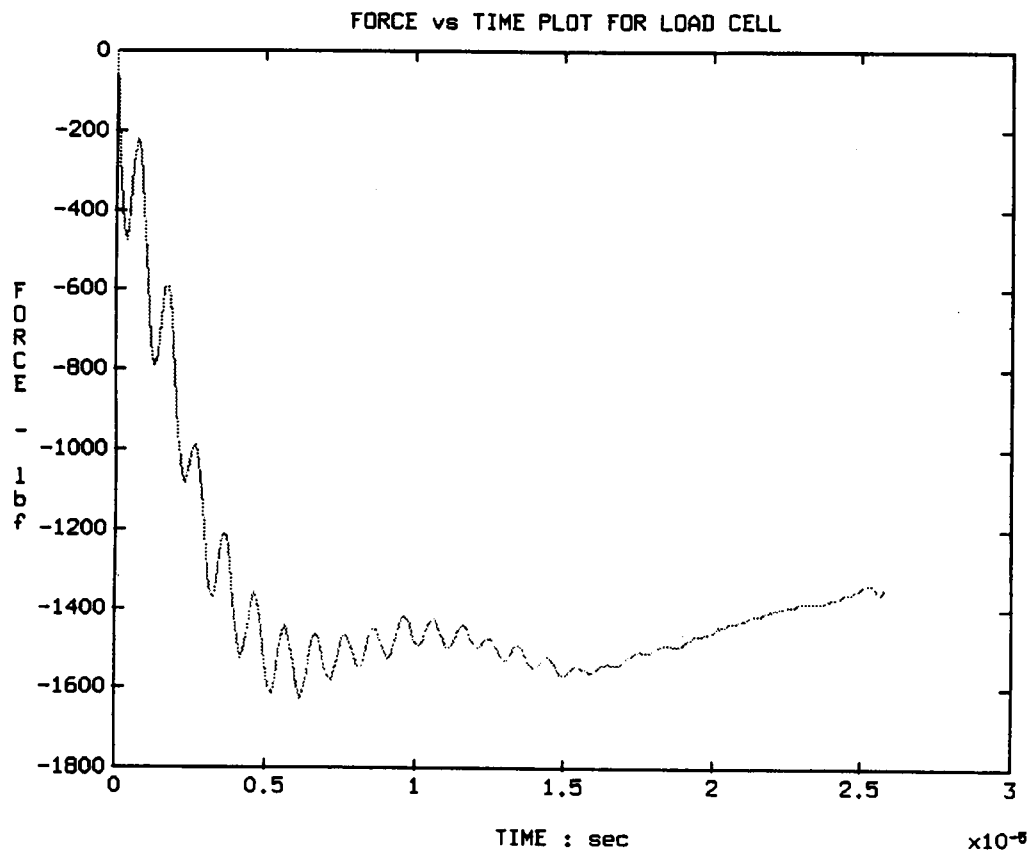


Figure 63. Finite element fracture force versus time plot for 0.0 to 0.028 milli-seconds

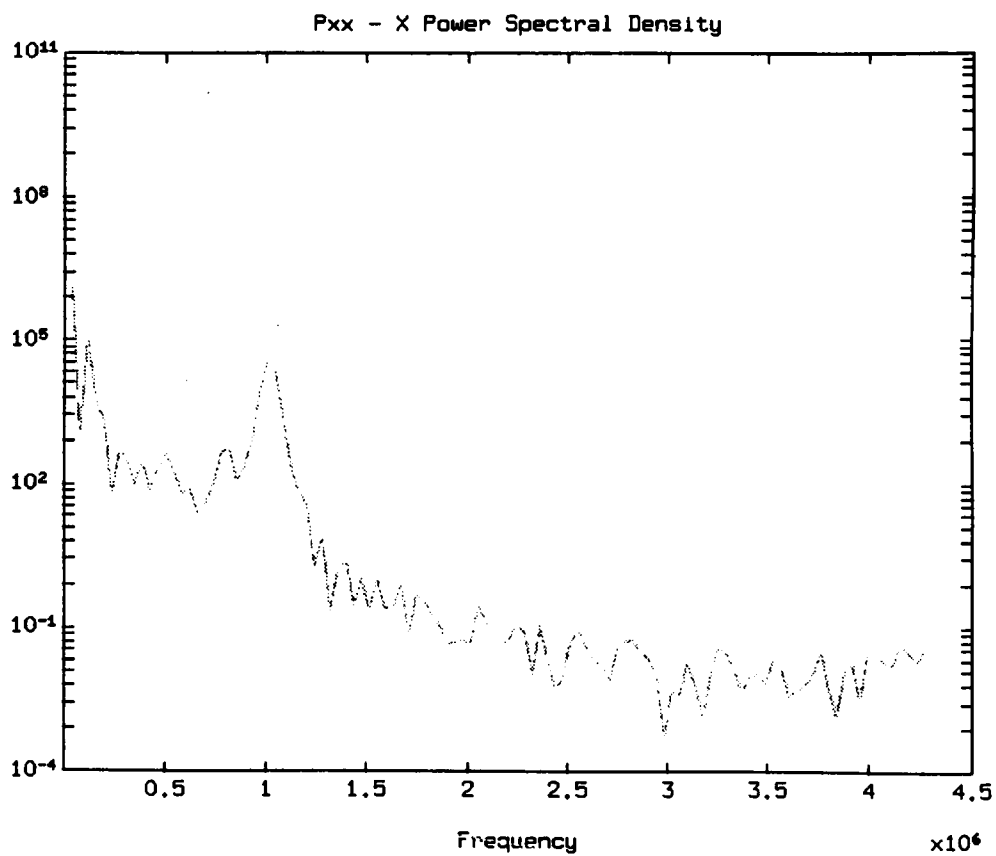


Figure 64. Power spectral density analysis of the finite element fracture force data.

Table 7

Frequency Values for Finite Element Model Components

Direction of Frequency Calculation	Impact Bumper Component	Load Cell Component	Complete Impact Bumper Configuration	Tibia Component
f_{xx}	25,396 Hz	204,918 Hz	-----	-----
f_{yy}	230,000 Hz	1,393,982 Hz	1,000,000 Hz	-----
f_{xx}	440,000 to 888,888 Hz	1,111,111 Hz	-----	-----
$f_{bending}$	A-110,661 Hz B-59,639 Hz	-----	-----	1,824 to 5,586 Hz

* Note: A indicates measurement one and B indicates measurement two.

using two different height measurements. The first measurement used was measured from the bottom of the rounded end of the bumper to the top of the rounded end; the second measurement value was measured along the rectangular section of the bumper excluding the rounded ends. Figure 65 shows the two height measurements used in the bending frequency calculation. This calculation was performed using the bending frequency calculation for the first mode of bending frequency in a free-free beam (Hartog, 1985). The ovaling frequency (first bending frequency) of the bone was calculated by assuming the bone was a thin walled ring (Hartog, 1985).

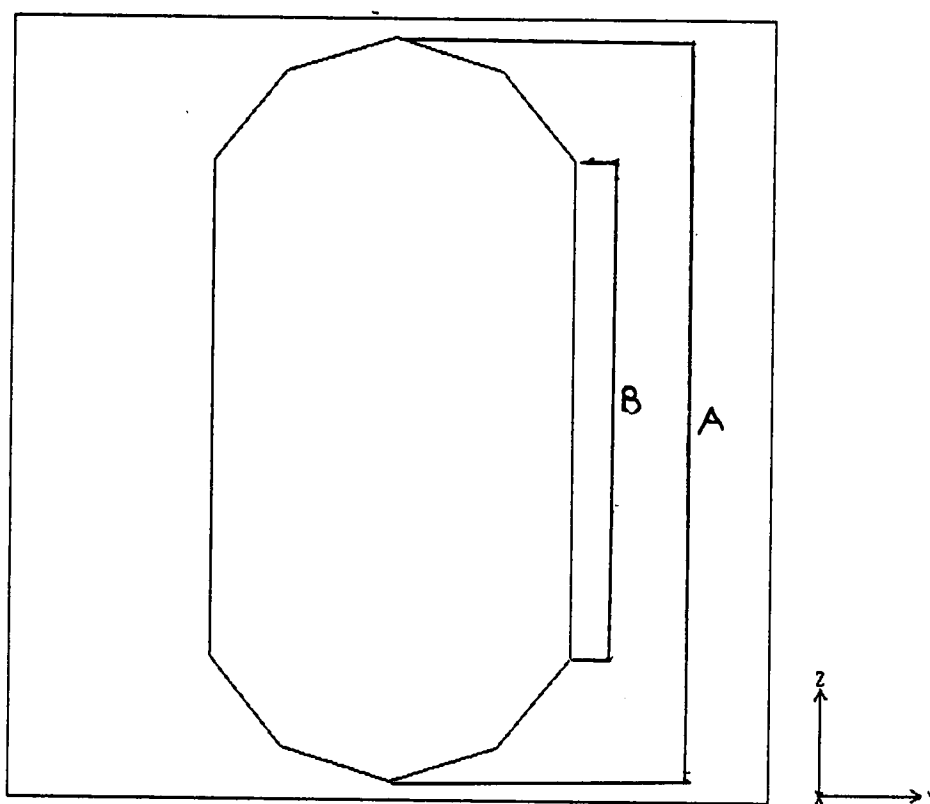


Figure 65. Height measurements for bending frequency calculation on the impact bumper

Chapter VII

Conclusions

It was hoped to verify the finite element model by the experimental testing. However, due to instrumentation problems with the experimental testing and due to the crude approximations used in the constitutive modeling of bone in the finite element model, only general trends and observations could be made in the comparison of the data.

A. Fracture Propagation

The fracture propagation general trend as observed from the experimental testing methods was found to be in the direction of a posterior to anterior direction. The post-impact photos of the bone specimens (appendix C) show a number of the tibias breaking into longitudinal pieces in a splintering effect. This is known as a comminuted fracture.

Relatively few tibia specimens showed transverse fracture at the tibia midshaft (appendix C). Both of these fracture trends were shown in the finite element model. The fracture contour plots of the finite element model do not show the direction of fracture, although, they do indicate longitudinal sections of bone that remain intact and do not undergo failure. This indicates a general trend of the bone to show longitudinal splintering effects. The plots also show complete failure across the midshaft of the bone which can indicate a transverse fracture. This is seen in a few of the experimental test specimens. If a constitutive model that tracked the direction of the fracture was used, it would probably show a combination of both types of fractures occurring. However, the present model can only show regions of failure in the bone. It would be too simplistic to state that the trend of bone fractures for this specific impact is either in the form of

longitudinal or transverse fracturing. More than likely, it is a combination of both. The trend of the fracture to propagate from the posterior to anterior direction, as observed from the experimental tests, needs further verification. It should be noted that the tibia specimen (977L) which showed a pre-impact longitudinal fracture, yielded results that were atypical of the fracture trends seen from the other tibias. The fracture propagation data from this tibia is disregarded. It was determined that the pre-impact fracture did indeed play a role in the direction that the fracture would propagate. The finite element model suggests that fracture is a combination of a variety of failure modes. It is often theorized that a straight posterior to anterior direction of fracture (transverse fracture) may indicate a bending mode failure. However, the computational model suggests that the fracture actually has a tendency to propagate from the middle of the midshaft toward the posterior side in a wedge shape and to propagate longitudinally along the anterior length of the tibia diaphysis. On the other hand, when looking at the interior or tibia cross-sections, the model suggests that the failure occurs along the interior posterior wall of the tibia and wraps around the tibial lateral to medial sides. The fracture also propagates along both the anterior interior and exterior cortical walls at the point of impact. Both the observations from the cross-sections and the exterior views of the bone definitely indicate multiple modes of failure.

B. Fracture Force

The fracture force data could not be compared between the computational model and the experimental testing. This was due to a file restart instability in the finite element code that caused the force versus time data to become corrupted. Only the force versus time data was

affected by this problem due to the fact that no restart files were necessary in the analysis of the fracture propagation. The data that was available from the finite element model before corruption occurred showed the trend of the data to reach its maximum at about 1600 lbf. This is comparable to the 1288 lbf seen in the experimental testing and provides for only a 24% difference in results. As a first time approximation and considering the crude approximations used in the constitutive model, this is considered a very acceptable result. The data past the 0.028 millisecond mark could not be analyzed. However, it showed a general trend to peak at the 1600 lbf mark.

The force versus time curve from the finite element model shows an oscillation in the data. To determine what causes this oscillation, each component of the finite element model was tested for its natural frequencies. The force versus time data was also plotted as spectral density versus frequency to observe the overriding vibrations in the model. These vibrations produce two peaks in the spectral density plot at values of 100,000 Hz and 1,000,000 Hz. The 100,000 Hz frequency is comparable to the frequency of the bumper in bending which occurs at about 110,666 Hz. However, the 1,000,000 Hz is a far more predominant frequency that corrupts the data. This frequency was found to be attributed to the impact bumper and load cell configuration. This configuration resulted in a frequency of 1,000,000 Hz. The cause of this frequency in the bumper configuration is due to the load cell. The load cell shows a frequency of about 1,393,982 Hz in the direction of movement ("YY" direction) and a frequency of 1,111,111 Hz in the plane perpendicular to the plane of movement of the bumper ("ZZ" direction).

Although both of these frequencies are higher than the 1,000,000 Hz frequency found in the force versus time data, the added mass of the bumper when combined with the load cell, lowers the frequency of the load cell. The primary direction of this visible frequency is in the direction of movement because it is the easiest mode of vibration to excite in the model. However, there is a slight possibility that this vibration can be coupled with the frequency in the "ZZ" direction. These vibrations can be filtered from the data at a later time, but it was more important to show these vibrations in raw form so that future finite element models can be developed to delete these vibration modes. These vibration modes have also been seen in the laboratory experimentation but have been filtered out of the data by the set sampling frequency of the digital analyzer. These vibrations were discussed by Kress in his thesis (1989).

C. Fracture Time

It should also be noted that fracture occurs very rapidly in the finite element model and that the peak force is reached much sooner than in the experimental testing. In fact, the whole finite element model never runs past 0.1 milliseconds before too much damage has occurred to the mesh to continue calculations. This is on the order of about 10 times faster than what is seen in the experimental testing results. The experimental tests show that the whole fracture process occurs over a 1.0 millisecond range with maximum fracture force occurring at about 0.5 milliseconds.

D. Summary

In summary, the finite element model showed promising results in the prediction of fracture propagation in the tibia. With the development of an accurate constitutive model, these results will only improve. The finite

element model also showed that there was a combination of failure modes in the tibia as it undergoes fracture and that fracture failure could not be described by a singular failure mode. The model showed that the general trend of the fracture propagation was in a probable longitudinal direction on the anterior bone surface. In addition, a fracture initiated at the middle midshaft position in the form of a wedge shape and propagated to the posterior midshaft position. The fracture force data was difficult to compare with the experimental testing. The trend showed that only a difference of about 24% in maximum fracture force data occurred; however, the entire fracture process occurred about 10 times faster than found experimentally. Overall, the finite element method shows promise for use in designing future tests, analyzing injury predictions, and designing injury mitigating devices. The concept of the finite element method allows for "measurements of data" to be taken everywhere simultaneously which provides for a more accurate explanation of results. This is something of which experimentalist can only dream. Nevertheless, without initial experimental testing, the finite element model is impossible to develop. Therefore, experimentation and computational modeling should always go hand in hand.

Chapter VIII

Recommendations

To enhance the research conducted in this thesis, there are several areas that can be improved. These areas fall within three main categories, finite element analysis improvements, experimental testing improvements, and future finite element analysis verification. These sections address the concerns and hopes for future work in the finite element analysis of human bone fracture.

A. Finite Element Analysis Improvements

The biggest problem with the finite element analysis of the human tibia is that the present constitutive model for the tibia is not an accurate representation of human bone under impact conditions. It is therefore recommended that if further finite element analysis is to be conducted, then an accurate constitutive model must be developed. It should address not only the fracture characteristics of the tibia, but also the material behavior of the bone under high strain rate activity. The problem in achieving this task is that, currently, there are few material properties available for high strain rates in human bone. There have been numerous papers published on the material behavior of specific human bones under very low strain rates using small cubes of bone material for testing. However, it is hard to find two researchers that agree even on these low strain rate material properties. Also, very little research has been conducted using whole intact bones. Results obtained from these tests will more than likely produce much different results than those results obtained from the small cube samples. Complicating the material property research is the anisotropic nature of bone. This anisotropy complicates conclusions on the

material properties of bone since the properties change along the length and across the cross-section of bone. Before a constitutive model can be developed, the material properties must be acquired for human tibia. A study of the relationship between the mechanical properties of bone for different age groups and different sexes, as well as, porosity in bone, and thickness of the cortical bone should be performed to determine if the constitutive model should account for these variables.

The constitutive model used in this research was a first time approximation to account for fracture in the tibia. The model uses an effective stress and mean stress calculation to address the fracture characteristics and only indicates a fracture has taken place. It does not indicate the direction of the fracture. These calculations are a crude approximation, and most fracture constitutive models use a maximum stress calculation for fracture characteristic. These models track the direction of the fracture and allow the material to crack open as the fracture propagates. Further constitutive modeling for the tibia should include these calculations.

In addition to constitutive modeling, the finite element model can be improved by using a more geometrically accurate mesh. The one presently used is based on crude approximations of measurements on the human tibia cross-sections and is symmetric from the lateral to medial side. This provides for an idealized cross-section of the human tibia. The mesh can be improved geometrically by digitizing points on several slices of tibial cross-sections. By taking the cross-section slices from the distal to proximal end of the bone, an accurate three-dimensional mesh can be created. These slices will more than likely show a non-symmetric geometry

across the cross-sectional slices. Although a geometrically accurate mesh does not guarantee accurate results in the finite element analysis, it is felt that it would be a welcomed improvement in the finite element model.

Implementing anisotropic behavior in the finite element analysis is another consideration that should be addressed in future research. It is not known how this would affect the finite element analysis and this addition may prove to be unnecessary with a more refined fracture constitutive model. It appears from the preliminary results in this research that fracture propagation and classification is a result of geometry more so than isotropic or anisotropic behaviors; the model predicted the general trends of fracture using an isotropic material.

B. Experimental Testing Improvements

The experimental testing instrumentation, setup, and protocol also need modification. One of the biggest problems encountered with the experimental testing involves the use of the crack gages. The main problem associated with the gages was their inability to adhere to the bone surface. They, therefore, could not measure the fracture propagation which was their intended purpose. Also, if the gages adhered to the bone, they would pull loose from their electrical connection wires at the solder joint. This disengagement at the solder joint broke the connection with the fracture measurement circuit and indicated a fracture had taken place across this gage when it had not. This situation provided erroneous data and complicated the data analysis process. Post-impact observation was the only method to decipher the adherence problem or the bad solder connection problem. It would be advisable to make the crack gages in the laboratory instead of purchasing them from Measurements Group. A crack

gage could probably be made cheaper than buying the gages. It could also be designed for the specific application of fracture measurement to human bones. A stronger adhesive is needed to maintain gage adherence to the bone. As a first attempt in the crack gage development area, burglar alarm tape may suit the purposes for a crack gage. This tape consists of a wire inside a thin tape construction. The tape is meant to tear easily, thus breaking the wire and resulting in a voltage drop measurement. Thus indicating fracture just as the present crack gages currently show. The advantage of this tape is that it is widely available in large rolls at a cheap price. The problem of adherence to the bone surface would still need to be addressed as well as how to avoid bad connections to the measurement instrumentation.

Due to limited resources and only a two channel, digital analyzer, only three crack gages were employed for fracture measurement. With only three gages, the bone could not be adequately instrumented to measure fracture propagation in all areas of the bone. Because of the splintering nature of the bone, the limited number of gages did not provide enough data to accurately determine fracture propagation. It is recommended that the bone be instrumented with as many gages as possible. An eight channel, high speed recorder can provide an increased number of gage measurements. A particular area of interest may be the woods metal and bone interface at the distal end; a fracture was often noticed in this area. The limited number of gages did not provide for adequate investigation as to whether the crack resulted from impact with the ground or during actual bumper impact with the bone.

The test setup should also be modified to restrain the distal cup from

impacting the floor after bumper impact. This impact with the floor often caused increased damage to the distal bone fragments still fixed in the distal woods metal. Often, the steel cup would flip 180 degrees, impacting the distal bone segment axially. This situation complicated the post-impact observations on fracture classification. The distal axial impact described above should be avoided in future testing.

C. Future Finite Element Analysis Verification

In order to further verify the finite element analysis of the bone fractures, it would be helpful to devise a scheme for post-test identification of each region of the bone. Combining this identification scheme with high speed filming of the impact and the crack gage data, accurate determination of the fracture propagation can be accomplished. An added benefit to the finite element analysis is that it identifies the area of fracture in the tibia. This allows for concentration of the high speed filming process in a specified region of interest. A wide camera angle would no longer be necessary to capture the entire bone and would avoid any loss of visual resolution.

Table 8 lists the necessary future plans needed to improve and expand the research described in this thesis. These steps can be accomplished in a time frame of approximately three to six years.

Table 8**Future Recommendations**

Steps	Finite Element Analysis	Experimental Testing
1	Use step 1 material properties from experimental testing for constitutive model	Determine high strain rate material properties and age, sex, and cortical bone thickness correlations
2	Perform finite element analysis with new constitutive model and compare to experimental testing step 5	Design new crack gages and fracture measurement circuits
3	Develop different geometric bumper meshes and compare the finite element analysis to experimental testing step 6	Develop a distal steel cup restraining device
4	Modify the tibia mesh for more accurate geometric modeling	Develop a fracture identification scheme
5	Examine finite element analysis steps 2 and 4 for change in accuracy	Perform high speed filming with experimental testing steps 2, 3, and 4 to determine fracture propagation
6		Perform experimental testing with different bumper geometries

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APPENDIXES

Appendix A

This appendix contains all of the crack gage plots of volts versus time and the corresponding force versus time plots for each of the eleven tibia specimen.

X=703.12 μ Sec
Ya=5.8489 KN

FILT TIME1
14.0

0%OV1p

Real

KN

-2.0

FxdXY -1.0m Sec

9.0m

CT991R

FILT TIME2
1.0

0%OV1p

Real

VOLTS

0.0

FxdXY -1.0m Sec

9.0m

CT991R

X=625.0uSec
Ya=6.21631 KN

FILT TIME1
14.0

0%OV1p

Real

KN

-2.0

FxdXY -1.0m Sec

CT971R

9.0m

FILT TIME2
1.0

0%OV1p

Real

VOLTS

0.0

FxdXY -1.0m Sec

CT971R

9.0m

X=703.124Sec
Ya=3.51005 KN

0%OV1P

FILT TIME1
14.0

Real

KN

-2.0

FxdXY -1.0m Sec

9.0m

CT926L

FILT TIME2
1.0

0%OV1P

Real

VOLTS

0.0

FxdXY -1.0m Sec

9.0m

CT926L

X=703.12 μ Sec
Ya=6.02929 KN
FILT TIME1
14.0

0%OV1p

Real

KN

-2.0

FxdXY

-1.0m

Sec

FILT TIME2

1.0

CT977R

0%OV1p

9.0m

Real

VOLTS

0.0

FxdXY

-1.0m

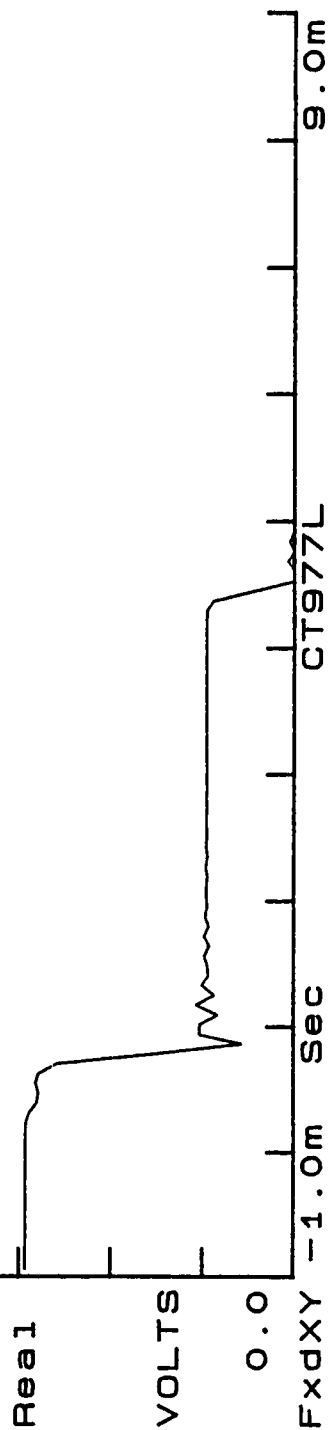
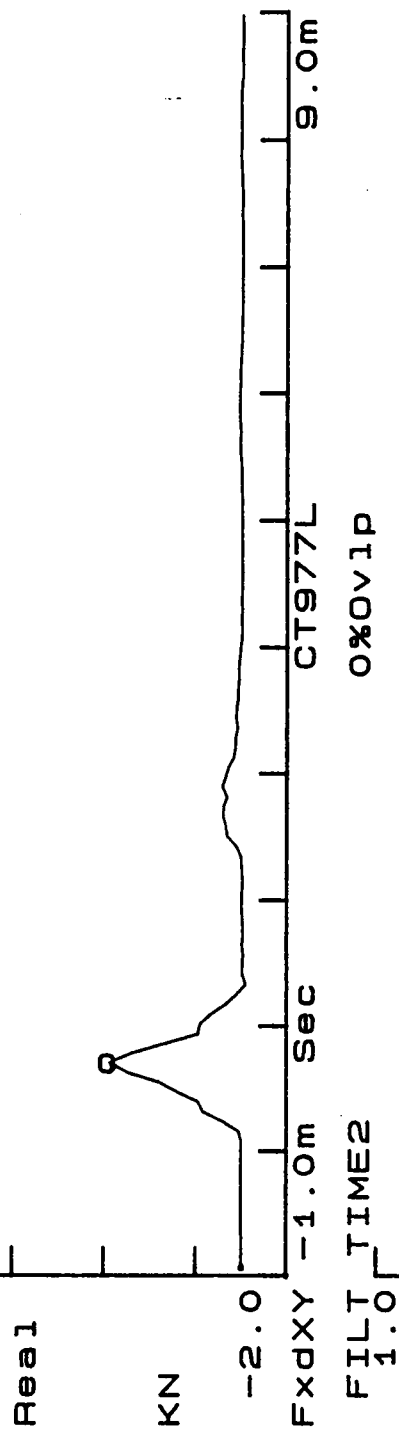
Sec

CT977R

9.0m

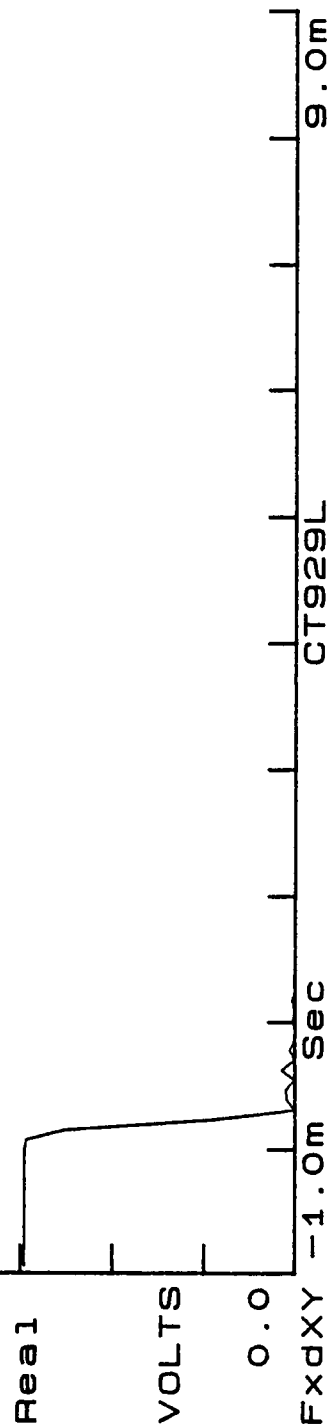
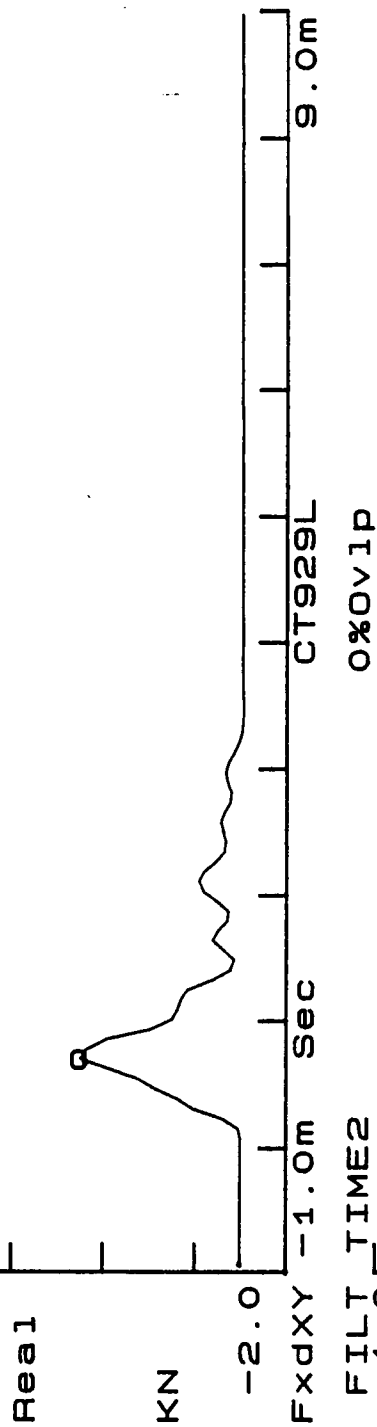
X=703.12 μ Sec
Ya=5.77964 KN
FILT TIME1
14.0

0%OV1P



X=703.12 μ Sec
Y=7.00064 KN
FILT TIME1
14.0

0%OV1P



X=703.12 μ Sec
Ya=6.91133 KN

FILT TIME1
14.0

0%OVlp

Real

KN

-2.0

FxdXY -1.0m Sec

9.0m

CT955L

FILT TIME2
1.0

0%OVlp

Real

VOLTS

0.0

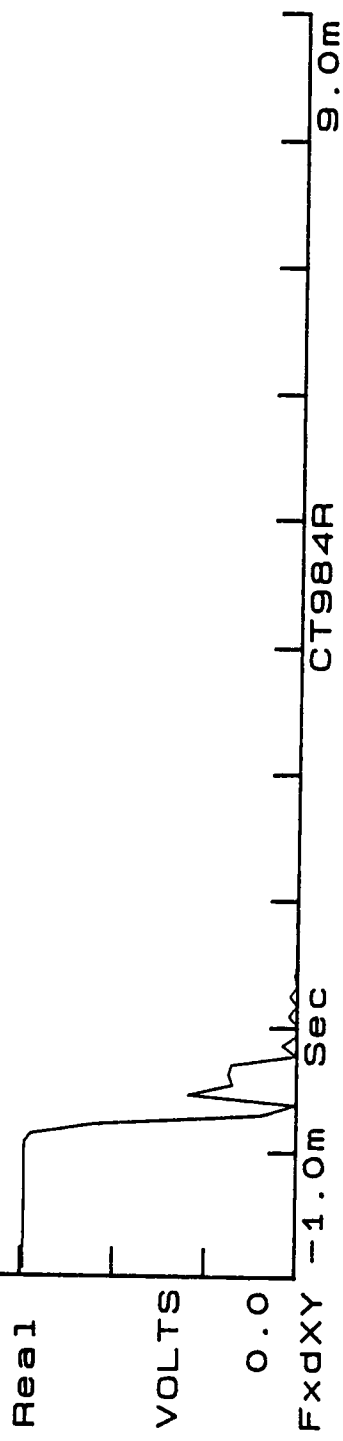
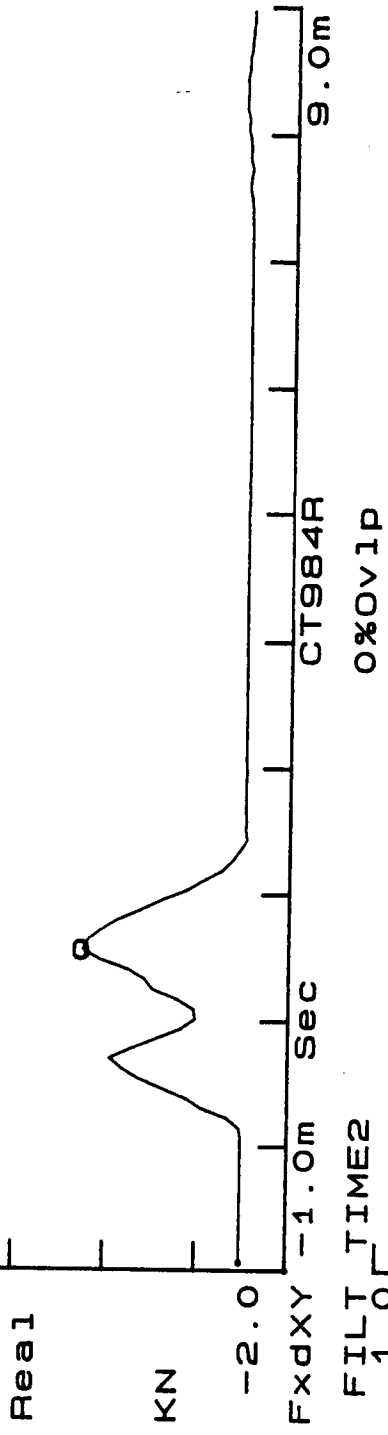
FxdXY -1.0m Sec

9.0m

CT955L

X=1.5625mSec
Ya=7.01833 KN
FILT TIME1
14.0

0%OV1p



X=703.12 μ Sec
Ya=7.33754 KN

FILT TIME1
14.0

0%OV1p

Real

KN

-2.0

FxdXY -1.0m Sec

9.0m

CT955R

FILT TIME2
1.0

0%OV1p

Real

VOLTS

0.0

FxdXY -1.0m Sec

9.0m

CT955R

X=468.75 μ Sec
Ya=3.37741 KN
FILT TIME1
14.0

0%OV1p

Real

KN

-2.0

FxdXY -1.0m Sec

CT002L

9.0m

FILT TIME2

1.0

0%OV1p

Real

VOLTS

0.0

FxdXY -1.0m Sec

CT002L

9.0m

X=781.25 μ Sec
Ya=4.00965 KN
FILT TIME1
14.0

0%OV1P

Real

KN

-2.0

FxdXY

-1.0m

Sec

CT002R

9.0m

FILT TIME2
1.0

0%OV1P

Real

VOLTS

0.0

FxdXY

-1.0m

Sec

CT002R

9.0m

Appendix B

This appendix contains various experimental measurements and laboratory observations. These are first impressions of the experimenter and serve as a laboratory notebook for future reference.

Bone Data Sheets

Cadaver #	Sex(m/f)	Age	Weight(g)
977L	f	65	387.0

Posterior midshaft	Medial Gage Configuration Distal	Anterior midshaft
Blue	Orange	Green

Gage Calibration Chart			
Blue R1	Orange R2	Green R3	Volts (mv)
C	C	C	740
C	C	O	712
C	O	C	676
O	C	C	334
C	O	O	640
O	C	O	245
O	O	C	130
O	O	O	-1

Speed(ft/sec)	Height Cup-Cup(cm)
24.9	25.0

Comments:

Bone was cracked longitudinal on lateral side from support to support before specimen was hung. We used it anyway.

The longitudinal crack before impact appears to have affected the bone impact data

Crack on impact propagates from A-P thus breaking the Green-Blue-Orange gages. Usually we see a P-A crack.

Bone Data Sheets

Cadaver #	Sex(m/f)	Age	Weight(g)
991R	f	42	435.2

Posterior midshaft	Medial Gage Configuration Distal	Anterior midshaft
Blu	Orange	Green

Gage Calibration Chart			
Blue R1	Orange R2	Green R3	Volts (mv)
C	C	C	748
C	C	O	719
C	O	C	681
O	C	C	337
C	O	O	643
O	C	O	247
O	O	C	131
O	O	O	-1

Speed(ft/sec)	Height Cup-Cup(cm)
25.4	26.0

Comments:

Blue gage never broke

Orange gage broke first.. Probably from Posterior side and crack propagaged anterior and up to green gage to break it.

Bone Data Sheets

Cadaver #	Sex(m/f)	Age	Weight(g)
971R	F	72	365.9

Posterior midshaft	Medial Gage Configuration Distal	Anterior midshaft
Blue	Orange	Green

Gage Calibration Chart			Volts (mv)
Blue R1	Orange R2	Green R3	
C	C	C	735
C	C	O	704
C	O	C	670
O	C	C	332
C	O	O	634
O	C	O	243
O	O	C	129
O	O	O	-1

Speed(ft/sec)	Height Cup-Cup(cm)
24.2	24.0

Comments:

Using duct tape to hold solder and wires firmly in place to help confirm the bad glue problem.

Orange gage broke loose from wire at solder joint and never broke

The bone break distally is caused by impact with floor

This was only bone to give a transverse fracture. All other bones had convoluted fracture. This may be a reason why crack gage worked and with convoluted it did not work. Also this was only time the glue held the gages on!

This one worked!

Bone Data Sheets

Cadaver #	Sex(m/f)	Age	Weight(g)
926L	f	85	252.0

Posterior midshaft	Medial Gage Configuration Distal	Anterior midshaft
Blue	Orange	Green

Gage Calibration Chart			
Blue R1	Orange R2	Green R3	Volts (mv)
C	C	C	736
C	C	O	706
C	O	C	671
O	C	C	332
C	O	O	633
O	C	O	243
O	O	C	129
O	O	O	-1

Speed(ft/sec)	Height Cup-Cup(cm)
25.4	20.0

Comments:

Using duct tape to hold solder and wires firmly in place.

Orange gage pulled loose and did not fracture across gage

The green gage breaking and then reconnecting caused by solder pulling off and then made contact again to make connection. The attaching wire to the green gage had lots of slack and as the bone lower part fell down the slack was taken up. The orange connection wire did not have slack and thus pulled loose from the solder before the green gage lost its slack and detached itself.

Green gage did not adhere to bone and was never actually severed. It came loose at the solder joint.

Orange gage never severed. It came loose at the solder joint.

Fracture convoluted and it was a good plot

Bone Data Sheets

Cadaver #	Sex(m/f)	Age	Weight(g)
977R	f	65	375.0

Posterior midshaft	Medial Gage Configuration Distal	Anterior midshaft
Blue	Orange	Green

Gage Calibration Chart			
Blue R1	Orange R2	Green R3	Volts (mv)
C	C	C	738
C	C	O	710
C	O	C	673
O	C	C	335
C	O	O	637
O	C	O	245
O	O	C	130
O	O	O	-1

Speed(ft/sec)	Height Cup-Cup(cm)
24.6	25.0

Comments:

Orange gage cracked in tension because there was no bone crack across the gage. This due to duct tape holding gage down maybe Blue gage broke first and then the green gage broke.
Crack propagates from P-A

Bone Data Sheets

Cadaver #	Sex(m/f)	Age	Weight(g)
929L	m	72	464.6

Posterior midshaft	Medial Gage Configuration Distal	Anterior midshaft
Green	Orange	Blue

Gage Calibration Chart			
Blue R1	Orange R2	Green R3	Volts (mv)
C	C	C	753
C	C	O	722
C	O	C	685
O	C	C	337
C	O	O	647
O	C	O	247
O	O	C	131
O	O	O	0

Speed(ft/sec)	Height Cup-Cup(cm)
24.3	26.7

Comments:

Blues gage tore from posterior-anterior direction
Orange gage broke at solder joints but was not torn
Green gage did not adhere to bone due to bad glue adhesion

Bone Data Sheets

Cadaver #	Sex(m/f)	Age	Weight(g)
955L	m	67	435.2

Posterior midshaft	Medial Gage Configuration Distal	Anterior midshaft
Blue	Orange	Green

Gage Calibration Chart			
Blue R1	Orange R2	Green R3(malfunction)	Volts (mv)
C	C	C	711
C	C	O	---
C	O	C	638
O	C	C	243
C	O	O	---
O	C	O	---
O	O	C	-1
O	O	O	---

Speed(ft/sec)	Height Cup-Cup(cm)
24.5	28.5

Comments:

Green gage is broken due to overheat on solder and can not be repaired

Bone Data Sheets

Cadaver #	Sex(m/f)	Age	Weight(g)
984R	m	80	520.2

Posterior midshaft	Medial Gage Configuration	
	Distal	Anterior midshaft
Green	Orange	Blue

Gage Calibration Chart			
Blue R1	Orange R2	Green R3	Volts (mv)
C	C	C	747
C	C	O	715
C	O	C	678
O	C	C	334
C	O	O	640
O	C	O	245
O	O	C	130
O	O	O	-1

Speed(ft/sec)	Height Cup-Cup(cm)
24.8	30.0

Comments:

Orange gage solder broke but gage was unbroken
 Bone broke at woods metal interface but did not break gage
 Bone break due to impact with floor
 Offcenter bumper hit caused bad force plots

Bone Data Sheets

Cadaver #	Sex(m/f)	Age	Weight(g)
955R	m	67	408.3

Posterior midshaft	Medial Gage Configuration Distal	Anterior midshaft
Green	Orange	Blue

Gage Calibration Chart			
Blue R1	Orange R2	Green R3	Volts (mv)
C	C	C	745
C	C	O	709
C	O	C	673
O	C	C	332
C	O	O	636
O	C	O	244
O	O	C	129
O	O	O	-1

Speed(ft/sec)	Height Cup-Cup(cm)
24.7	28.5

Comments:

After impact picture show that glue is bad and wont adhere to bone.

Bone Data Sheets

Cadaver #	Sex(m/f)	Age	Weight(g)
002L	f	53	238.5

Posterior midshaft	Medial Gage Configuration Distal	Anterior midshaft
Blue	Orange	Green

Gage Calibration Chart			
Blue R1	Orange R2	Green R3	Volts (mv)
C	C	C	740
C	C	O	711
C	O	C	676
O	C	C	332
C	O	O	639
O	C	O	243
O	O	C	130
O	O	O	-1

Speed(ft/sec)	Height Cup-Cup(cm)
24.9	19.0

Comments:

Crack gage did not appear to work.
The green gage shows broken from pulling at solder joints.
Tension wedge developed in bone crack

Bone Data Sheets

Cadaver #	Sex(m/f)	Age	Weight(g)
002R	f	53	261.0

Posterior midshaft	Medial Gage Configuration Distal	Anterior midshaft
Blue	Orange	Green

Gage Calibration Chart			
Blue R1	Orange R2	Green R3	Volts (mv)
C	C	C	718
C	C	O	687
C	O	C	650
O	C	C	333
C	O	O	605
O	C	O	244
O	O	C	130
O	O	O	-1

Speed(ft/sec)	Height Cup-Cup(cm)
25.1	19.0

Comments:

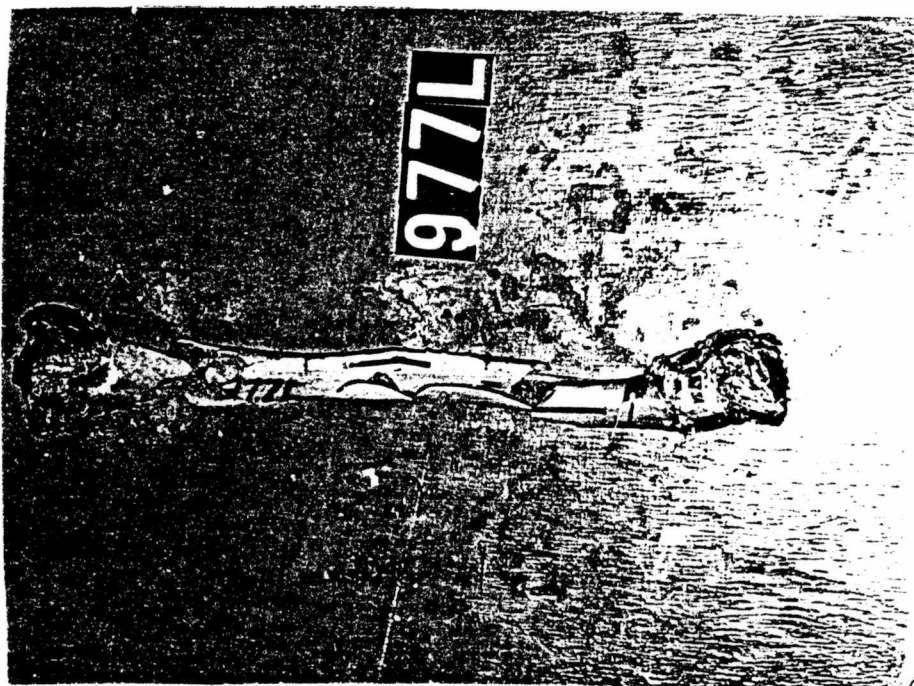
Solder connection very bad!
Duct tape is holding solder joint together to the crack gage for all three gages.

CADAVER #	CT 1	CT 2	CT 3	CT 4	CT 5	CT 6	VI	VI	DELTV	MASS	MASS	Density
	inches	mm	inches	mm	inches	mm	mm	mm	mm ³	grams	g/cm ³	g/cm ³
925L	0.537	14.148	0.163	4.140	0.260	6.603	0.231	5.867	8.565	0.310	7.874	1.304368
960R	0.529	13.284	0.246	6.248	0.312	7.925	0.400	10.160	0.285	7.239	730.000	1.455398
968L	0.551	13.965	0.137	3.490	0.355	9.017	0.335	8.509	0.280	7.113	710.000	1.488657
929L	0.553	14.045	0.166	4.216	0.320	8.126	0.265	7.264	0.265	6.759	775.000	1.395950
964R	0.693	17.548	0.267	6.792	0.475	12.065	0.265	6.731	0.344	8.738	710.000	1.333948
955L	0.320	8.128	0.155	3.937	0.324	8.200	0.237	6.020	0.256	6.502	1010.000	1.603365
955R	0.247	6.274	0.157	3.986	0.316	8.026	0.235	5.969	0.315	8.001	1010.000	1.512222
971L	0.436	11.125	0.176	4.470	0.340	8.636	0.267	6.762	0.201	5.105	950.000	1.419008
926L	0.489	12.421	0.216	5.486	0.281	7.135	0.253	6.426	0.270	6.958	865.000	1.355185
991R	0.215	5.461	0.103	2.616	0.135	3.429	0.145	3.663	0.255	6.477	910.000	1.44
971R	0.532	13.513	0.246	6.248	0.330	8.362	0.259	6.579	0.358	9.093	1005.000	1.450087
977R	0.453	11.760	0.486	12.395	0.244	6.196	0.262	6.655	0.203	5.156	1005.000	1.450994
977L	0.373	9.474	0.237	6.020	0.278	7.061	0.249	6.325	0.234	5.944	980.000	1.382143
002L	0.402	10.211	0.159	4.036	0.148	3.759	0.127	3.226	0.152	3.861	520.000	1.400225
002R	0.424	10.770	0.169	4.263	0.170	4.316	0.173	4.364	0.174	4.420	595.000	1.45
Avg	0.426	10.876	0.212	5.374	0.278	7.072	0.230	5.838	0.252	6.409	710.687	1.444

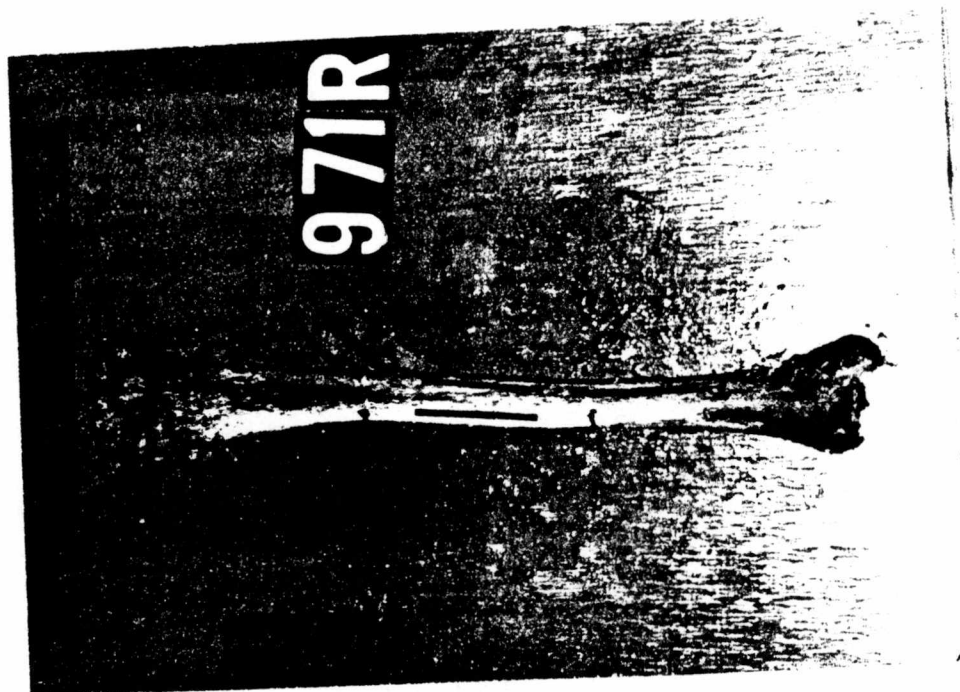
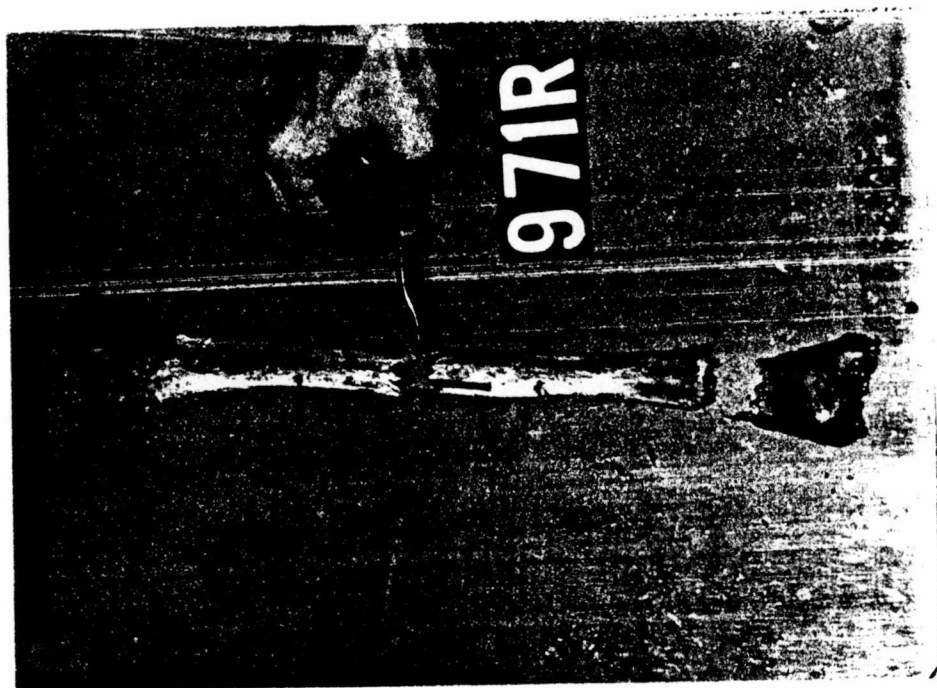
AVERAGE PROPERTIES	OF	TEST	TIBIAS
TOT AVG (in)	0.263543		
TOT AVG (mm)	6.695017		
		CORTICAL THICKNESS	
TOT AVG (cm ³)	261.4187		
TOT AVG (in ³)	15.96293		
		DISPLACED VOLUME	
TOT AVG (g)	375.4083		
TOT AVG (mg)	0.002145		
		MASS OF TIBIA	
TOT AVG (g/cm ³)	1.44413676		
TOT AVG (g/in ³)	0.00013502		
		DENSITY OF TIBIA	

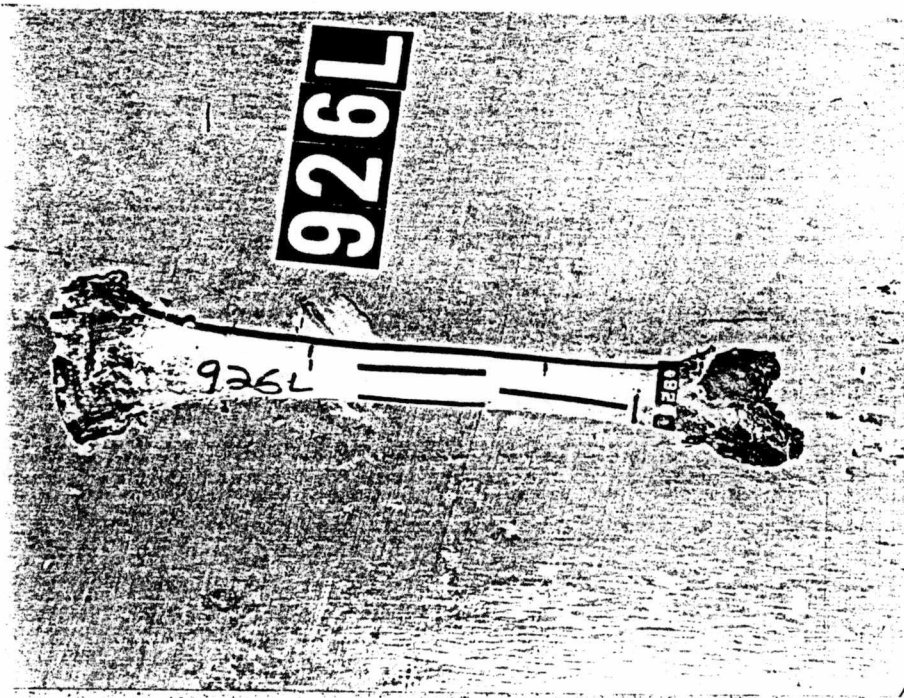
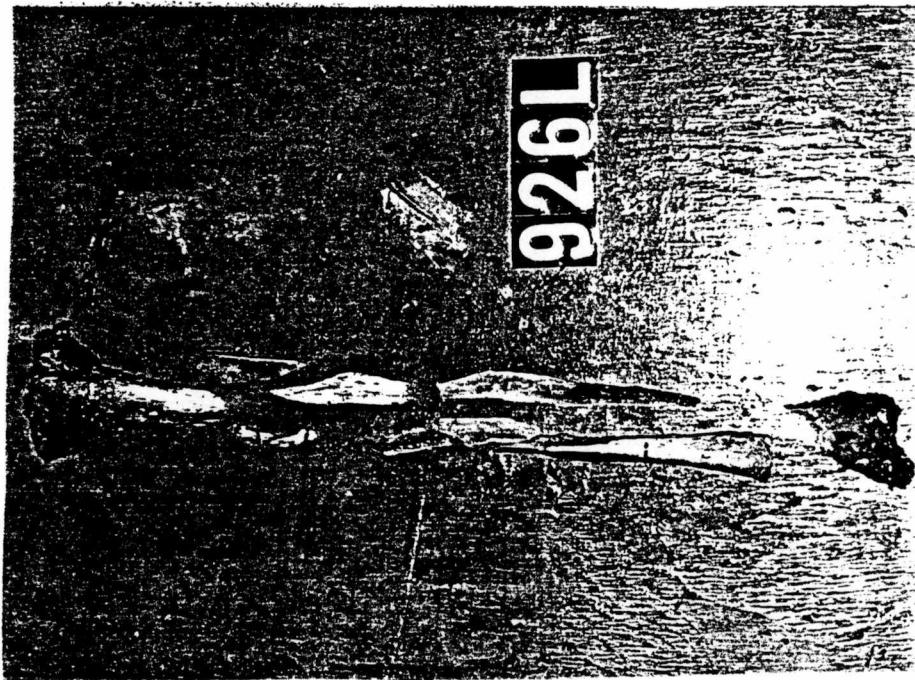
Appendix C

This appendix contains the pre- and post-impact photos for each of the eleven tibia specimens. Some pre-impact photos were not available for publication. The post-impact picture for Tibia 991R was mislabel as 971R at the time of photography.

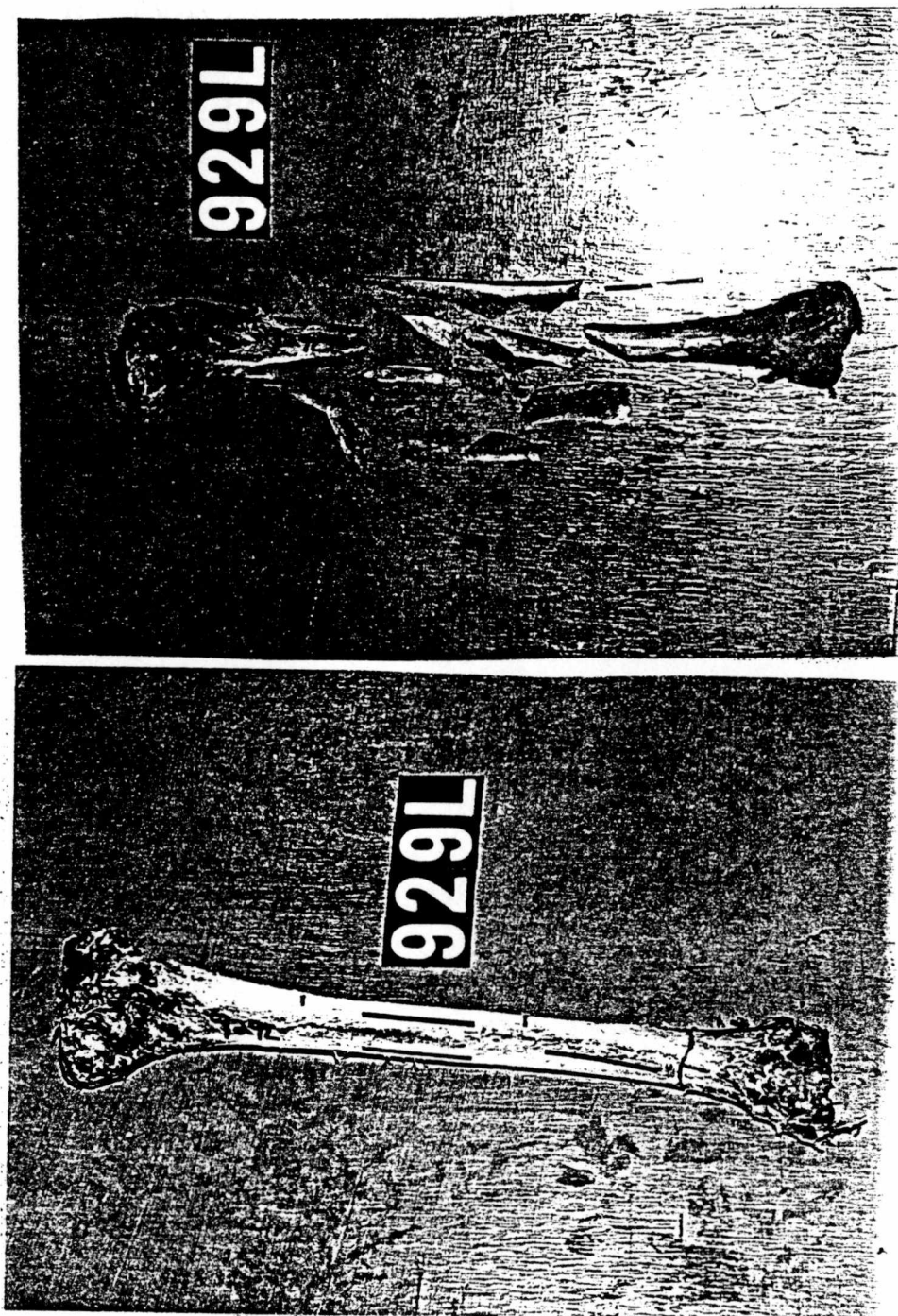




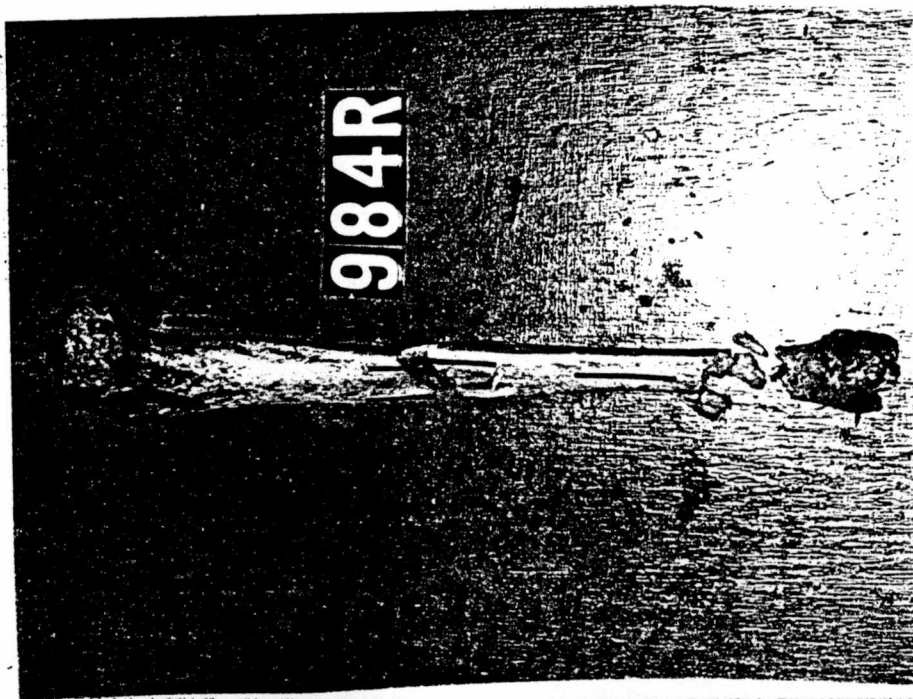


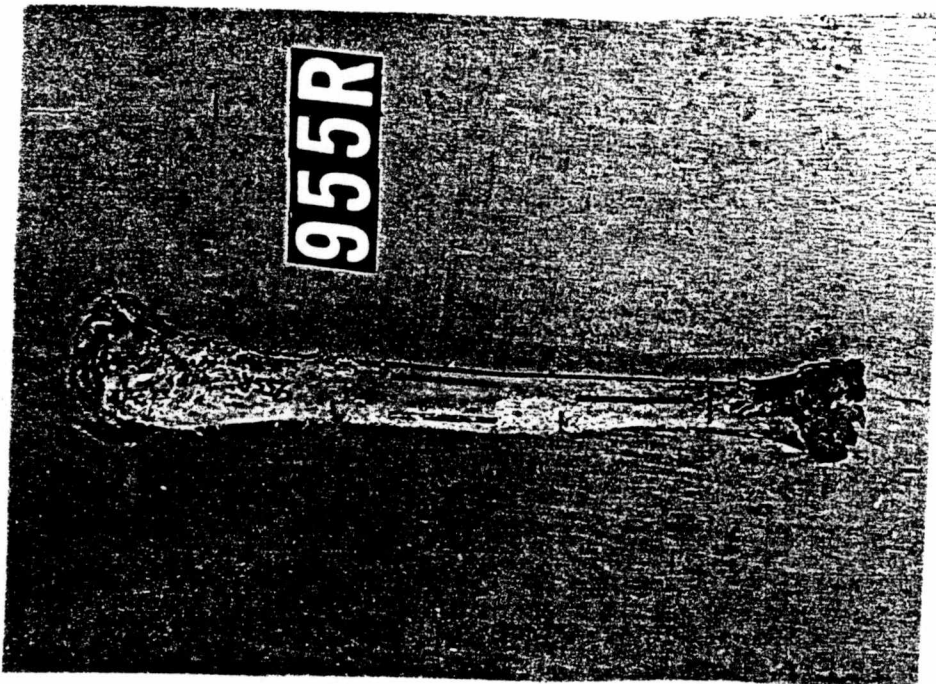


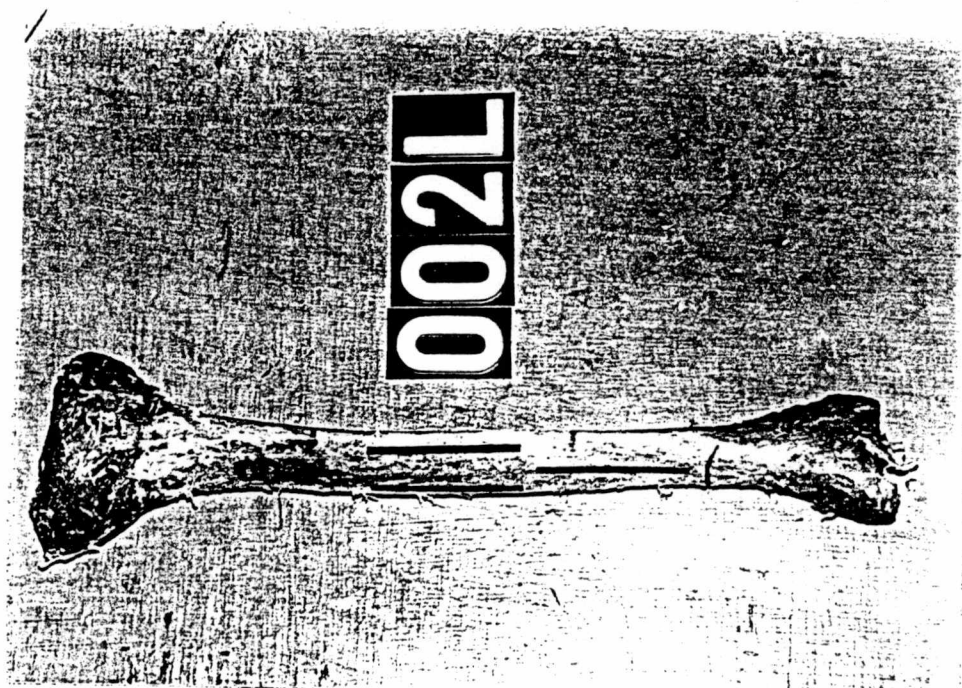
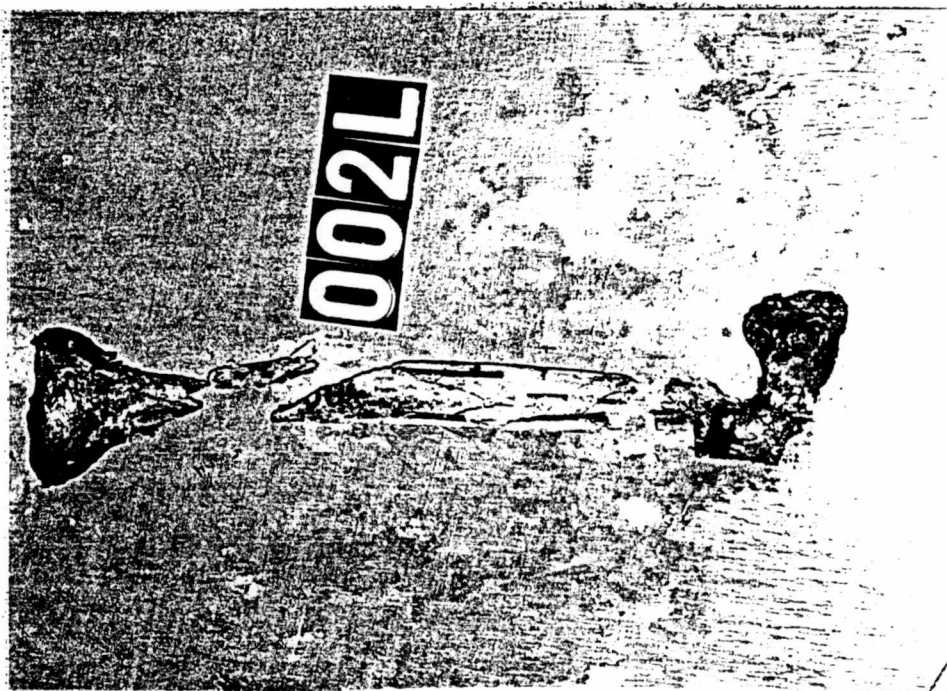


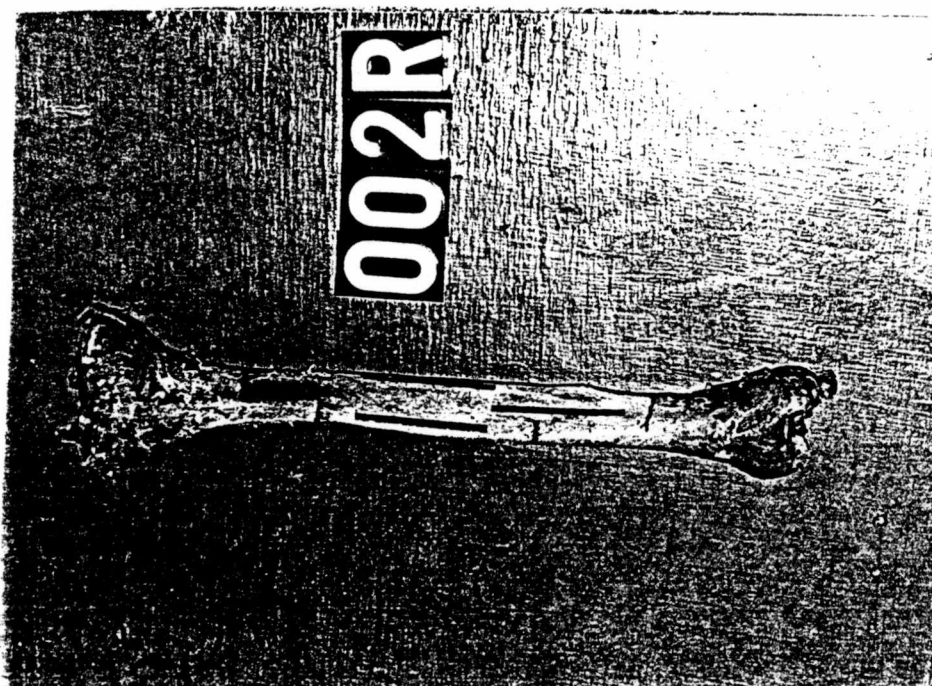












VITA

Michael Brian Cooper was born in Lewisburg, Tennessee on October 15, 1968. He attended Marshall County Senior High School where he participated strongly in the science club. During his sophomore year he placed first in the medicine and health category, as well as becoming a semifinalist in Vanderbilt's Middle Tennessee Science and Engineering Fair. During his senior year he was student council president and throughout his high school years he was lighting director for the high school and community theater productions. He graduated in May, 1986. The following September he entered the University of Tennessee's engineering curriculum and subsequently graduated with a Bachelor of Science in Engineering Science and Mechanics in May, 1991. Starting June, 1991, he entered the graduate program at the University of Tennessee in the department of Engineering Science and Mechanics; he wished to concentrate his efforts in the area of computational biomechanical engineering. In May, 1993, he received his Master of Science degree in Engineering Science and Mechanics.

Throughout his college career, Michael Cooper has always held one principle close in his pursuit for higher education; this principle is to make any sacrifice necessary to get to the golden ring.