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THORACIC MODEL IMPROVEMENTS (EXPERIMENTAL TISSUE PROPERTIES)

VOLUME I. EXECUTIVE SUMMARY

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16. Abstract

The general objective of this research program was to obtain mechanical properties, both stress, strain, and rupture strength, for various human tissues that are directly applicable to the thoracic injury problem as defined in the finite element model of the human thorax that is being developed by the Franklin Institute Research Laboratory (FIRL) under NHTSA Contract No. DOT-HS-243-2-424, "Thoracic Impact Injury Mechanism." The properties were determined at strain rates that can occur during fatal automobile accidents. The properties of Rhesus monkey tissues are also of interest in the modelling effort, and consideration was given to providing experimental data on selected Rhesus tissues as well as human tissues.

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1.0 INTRODUCTION

This report is a brief summary of the results of a project entitled, "Thoracic Model Improvements (Experimental Tissue Properties)" conducted by the Biomechanics Department of the Highway Safety Research Institute of the University of Michigan.

The general objective of this research program was to obtain mechanical properties, both stress, strain, and rupture strength, for various human tissues that are directly applicable to the thoracic injury problem as defined in the finite element model of the human thorax that is being developed by the Franklin Institute Research Laboratory (FIRL) under NHTSA Contract No. DOT-HS-243-2-424, "Thoracic Impact Injury Mechanism." The properties were to be determined at strain rates that can occur during fatal automobile accidents. The properties of Rhesus monkey tissues are also of interest in the modelling effort, and consideration was given to providing experimental data on selected Rhesus tissues as well as on human tissues.

The general approach to achieving the goals of the program were to:

1. Perform a detailed literature survey concentrating on soft tissue test techniques and mechanical properties data on the tissues of primary interest.
2. Develop basic test techniques, fixture design and test analysis methods.
3. Implement the test program with tissue priorities based on the needs of the modelling effort.
4. Analyze and synthesize the test data with respect to developing constitutive relations to describe the behavior of the tissues tested within the scope of the test program.

The following sections of this report summarize the methodology used in the conduct of this project and the results obtained.

1.1 BACKGROUND

Many investigators have performed mechanical tests on various biological tissues which exhibit the general characteristics of markedly non-linear behavior accompanied by finite deformations. The use of principles of solid mechanics to assist in the description of the mechanical properties and behavior of biological materials requires consideration of questions of isotropy, homogeneity, compressibility, elasticity and viscoelasticity, and universality. Soft tissues such as muscle and ligaments are clearly composite materials, preferentially structured and, microscopically, far from isotropic and homogeneous. However, in some tissues effective homogeneity may exist on a macroscopic scale. Soft tissues are generally accepted to be incompressible (1), although in some states of compression, a volume decrease can occur due to exudation of fluid (2). It is also recognized that soft tissues in general are viscoelastic (3), however, under certain conditions it is useful to consider the elastic behavior of some tissues only (1). The question of universal representation of the behavior of soft tissues poses a complex experimental problem. In cases where consideration is of elastic behavior only, the application of experimental and analytical techniques developed in the study of elastomeric polymers can be applied to soft tissue characterization. This approach utilizes the concepts developed in finite elasticity (4) and involves determination of a strain energy function W which will allow a description of the behavior of the material specifically through the parameters $\partial W/\partial I_1$, $\partial W/\partial I_2$ and $\partial W/\partial I_3$, where I_1, I_2 and I_3 are the strain invariants. These parameters are functions of the state of deformation in the material and require measurement of stress resultants and finite strains for a sufficiently large array of states of strain. For an incompressible material, the principal strains

are interrelated so that W becomes independent of I_3 thereby reducing the experimental work somewhat. Suitable use of thin membranes of material also allows further simplification of experimental procedure. Nevertheless it is necessary to conduct tests under at least three deformation states to describe an isotropic incompressible material. Experimentally, the three states usually chosen are: simple elongation, pure shear and two dimensional extension (or uniaxial compression for incompressible materials). Very few studies have utilized anything other than the uniaxial tensile test to describe the elastic behavior of soft tissues. A variety of forms for the strain energy function have been proposed and evaluated for rabbit mesentery (3, 5) and cat's skin (6) based on uniaxial tensile data. Fung (7) last year reviewed current efforts along these lines and has indicated that "the greatest need lies in the direction of collecting data in multiaxial loading conditions and formulating a theory for the general rheological behavior of living tissues in multiaxial loading conditions. None of the numerous theoretical proposals has received extensive experimental support." Lanir and Fung (8) have hust recently published the preliminary experimental results on the two dimensional mechanical properties of rabbit skin.

Typical soft tissues are not elastic and their viscoelastic nature must be accounted for as there is considerable difference in stress response to loading and unloading and to rate loading (3). Many soft tissues have been studied using simple states of strain such as uniaxial tension, simple shear and uniaxial compression to obtain viscoelastic data. These tests are usually either creep, relaxation or steady state small oscillation tests and the results are usually discussed in terms of the framework of linear viscoelastic theory relating stress and strain on the basis of the Voigt, Maxwell and Kelvin models (9, 10, 11). A nonlinear theory of the Kelvin type has

been proposed for tendons and ligaments (12) on the basis of a sequence of springs of different natural length, with the number of participating springs increasing with increasing strain.

For finite deformations, the nonlinear stress-strain characteristics of soft tissues must be accounted for. An alternative to the development of a constitutive equation by gradual specialization of a general formulation has been put forward by Fung (3). Utilizing special hypotheses, the history of the stress response in a material subjected to a uniaxial step elongation is called the relaxation function $K(\lambda,t)$ and is assumed to have the form

$$K(\lambda,t) = G(t)\, T^{(e)}(\lambda),\quad G(0) = 1$$

in which a normalized function of time, is called the reduced relaxation function, and $T^{(e)}(\lambda)$, a function of λ alone is called the elastic response. This formulation allows the function $T^{(e)}(\lambda)$ to play the role assumed by the strain in the conventional theory of viscoelasticity, thereby extending the machinery of the theory of linear viscoelasticity to use in characterization of nonlinear materials. In cases where the stress response to a loading process is insensitive to the rate of loading $T^{(e)}(\lambda)$ may be approximated by the uniaxial tensile stress response in a loading experiment with a sufficiently high rate of loading. Recent results suggest that the characteristic relaxation term may also depend on the strain level. This would result in $G(t)$ depending on λ.

Much of the experimental work on determining the mechanical behavior of soft tissues has not been cast in the framework of a complete solid mechanics description, thus it is possible to obtain only a partial characterization of tissue behavior from the existing literature. The most extensive summary of

work on the simple loading behavior of biological tissue is that of Yamada (13). The basic tests reported include uniaxial tension, biaxial membrane inflation and burst, tubular inflation, torsion and direct compression on a very wide range of human and animal tissues. The data presented in this extraordinary work provides no information of real use to a continuum mechanics characterization of the tissue behaviors. A survey of the existing literature on soft tissue properties reveals almost a complete lack of data on the stress-strain behavior to failure at anything other than quasi-static loading rates.

The main emphasis in this program was upon impact type high strain rate behavior of biological tissues and as such the main form of data acquisition was in terms of high strain rate stress-strain curves to failure. The experimental results obtained in this program were analyzed to provide the most useful data in terms of failure criteria and numerical parameters in forms most suitable for the FIRL finite element modelling effort. The details of the experiments performed on the various tissues, the test parameters measured, and the form of data analysis were coordinated to the extent possible with both FIRL and the CTM during the test technique development phases of the program. The limited scope and duration of the program (10 man months equivalent) was not sufficient to allow a complete continuum mechanics description of all the tissues of interest. In fact, the overriding consideration of data generation for the finite element model strongly influenced the tests performed and the subsequent data analysis. However, wherever possible the test configurations, measured test parameters and subsequent analysis were couched within the framework of proper solid mechanics characterization so that the data developed, while at this point in time may be incomplete in the sense of a total description, will be applicable to future efforts.

2.0 LITERATURE SURVEY

A literature survey which concentrated on the areas of soft tissue testing and analytical representation of soft tissue mechanical behavior was implemented early in the program as an aid in guiding the experimental design and data analysis techniques. In addition, the literature survey was used to study mechanical properties data on thoracic tissues of primary interest to the project.

A bibliography based on the papers located by the literature survey was compiled in the following manner:

1. Obtain the paper.
2. Code it.
3. Abstract the article.
4. Type the information on a Keysort card.
5. Punch the card according to the code.

Duplication of the Keysort card bibliography was done by Xeroxing the cards on a dark background at a slight reduction in order to allow attachment of the duplicate to a blank Keysort card for punching as indicated by the duplicate. The copies of these bibliography cards are contained in Volume III of this report. A total of over 500 cards were prepared during the project. In addition, selected articles of particular pertinence to the program have been microfilmed in their entirety and supplied in microjacket form to the CTM.

3.0 TEST TECHNIQUES

3.1 GENERAL

The effectiveness of a program in providing mechanical behavior data on varied tissues of the thorax depends on many factors. The test techniques and procedures must be kept as simple and efficient as possible in order to allow significant numbers of tests to be performed on each tissue with

statistical validity. Additionally, the measurements made in the tests must provide sufficient information to allow the complete determination of the state of strain and state of stress for a proper analytical characterization of the behavior of the tissue. In addition to these basic requirements, the test program must consider the questions of the effects of muscle tone, material storage, time after death, and in vivo configuration on the mechanical behavior of the tissues.

The test methods planned for use in the program can be classified into two types; basic mechanical properties tests and structural mechanical properties tests.

The basic mechanical properties tests consist of the following candidate loading states:

1. Uniaxial tension
2. Biaxial tension (membrane pressurization)
3. Uniaxial compression equivalent to biaxial tension for incompressible materials)
4. Combined axial tension and internal pressurization (with suitable tubular vessel and organ samples).

The structural mechanics tests involve those tissue configurations where it was impossible to produce suitable samples of the tissue and therefore were tested as a structure.

Selection of which of the above test states was to be applied to each of the tissues of interest depended on many factors. The structural geometry of the tissue sample precluded certain testing modes. The degree of anisotropy that a particular tissue exhibits may limit the application of the more complex loading states. In addition to material considerations, the primary goal of providing data for use in the FIRL finite element model played a major role in the selection of test modes.

Due to limited supply of suitable material samples and the need for significant numbers of tests in any one test mode and tissue type, only uniaxial tension and biaxial tension tests were performed routinely to obtain basic mechanical properties.

3.1.1 Uniaxial Tension Test

The uniaxial tension test could be applied to virtually any tissue of interest. Three types of uniaxial tensile specimens were used in the program. The first two consisted of die cut specimens (designated Type 1 and Type 2 specimens). The smaller of the two die shapes (Type 2) was used in some human tissues where there was a limited amount of uniform tissue and in Rhesus monkey tissues. The third type of uniaxial tension specimen was the ring type specimen which can be formed by making two parallel transverse cuts across a tubular vessel. It was used in monkey tissue tests only. This type of specimen can be loaded with simple pin loading while the die cut specimens required a special low mass air grip design. The two test configurations are shown schematically in Figure 1 for the die cut specimen and Figure 2 for the ring type specimen.

The strains in the uniaxial tension test were determined photographically in the case of the die cut specimens and by pin displacement in the ring type specimens. The photographic strain measurement employed a 45° angle mirror which allowed recording of thickness changes on the same photograph as the width and length changes. The specimen was stamped with a grid of lines with a 0.25 inch spacing using conventional stamp pad ink. Prior to testing, the thickness of the specimen was determined at three points along the specimen using an Ames 5642-1 thickness gauge. The camera used in static tests was a 35 mm Honeywell Pentax and in dynamic tests it was a 16 mm Photosonics 1B high speed (1000 pps) movie camera. Load measurement was done with

Figure 1. Schematic Representation of Tension Test with Die Cut Specimen

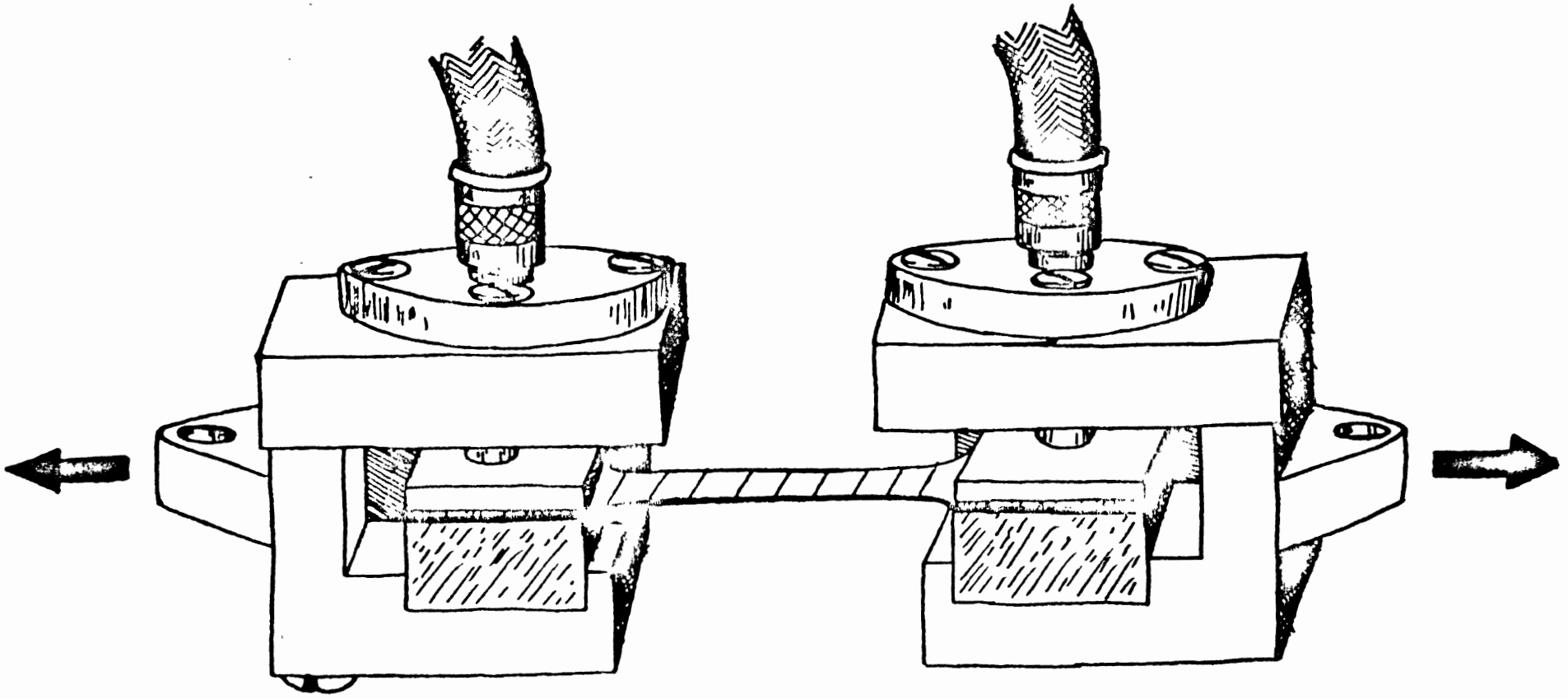
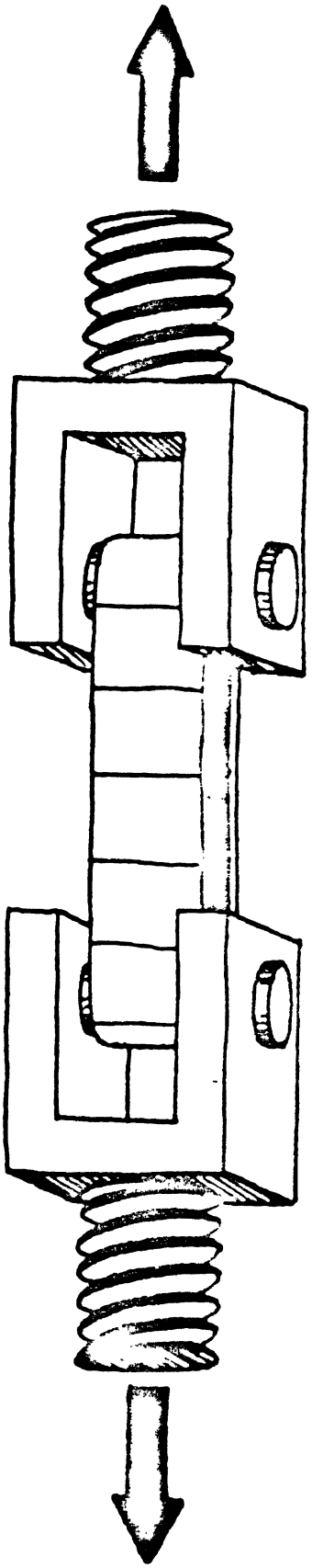


Figure 2. Schematic Representation of Tension Test with Ring Type Specimen



an Instron strain gage load cell in the static tests and a Kistler 931A piezoelectric load cell in the dynamic tests. Synchronization of the strain analysis pictures with the load trace was achieved in the static tests by placing a photodiode in front of the strobe light used to illuminate the specimen. For each picture the resulting flash of light produced a voltage spike in the photodiode which was displayed on a separate channel along side of the load trace. In the dynamic tests a timing pulse generator was used which produced timing marks on the film and on the load trace. In both static and dynamic tests the load trace and the synchronization pulses were recorded on a Honeywell Visicorder light beam oscillograph. In the dynamic tests the grip displacement was also recorded.

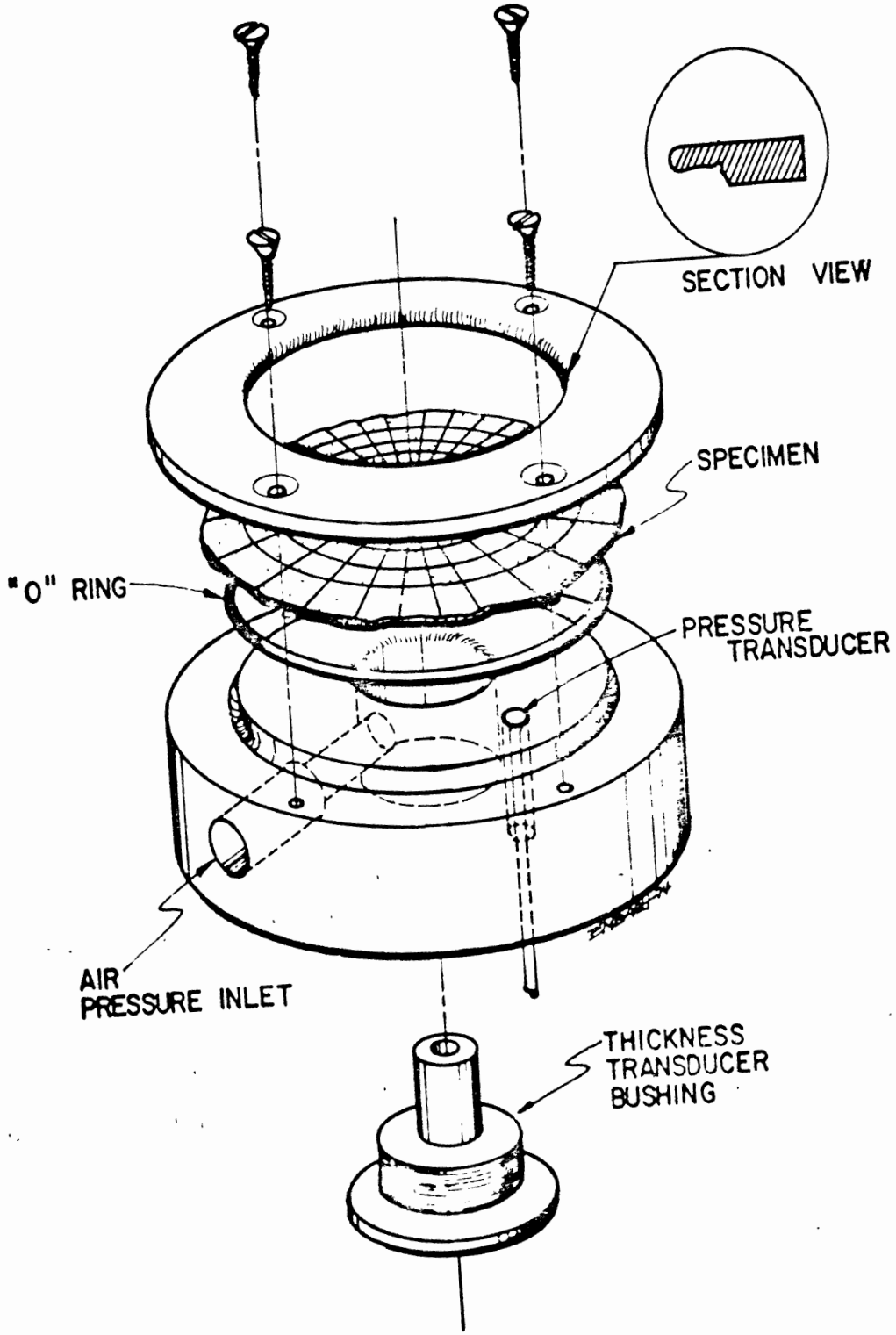
The static tests were performed in a Instron TTC floor model universal testing machine at a crosshead speed of 0.5 inches/minute. The dynamic tests were performed in a Plastechon high speed universal testing machine at ram speeds of nominally 360 and 3600 inches/minute.

The analysis technique used in reducing the uniaxial tension test data is given in Appendix A of Volume II.

3.1.2 Biaxial Tension Test

Biaxial tension tests could be performed on tissues which were membraneous in nature and had suitably uniform regions of material (approximately 2.5 inches in diameter). Figure 3 shows an assembly drawing of the finalized test apparatus developed for biaxial tension testing. The device was mounted on an air chamber with a quick-opening solenoid valve. For dynamic operation, the air chamber was charged with compressed air at about 100 psi with the solenoid valve closed. The valve was then opened and the membrane inflated to failure dynamically in approximately 10 msec.

Figure 3. Assembly Drawing of Biaxial Tension Test Apparatus


SECTION VIEW
SPECIMEN
"O" RING
PRESSURE TRANSDUCER
AIR PRESSURE INLET
THICKNESS TRANSDUCER BUSHING

The pressure acting on the membrane was measured by a Kistler piezoelectric pressure transducer and the deformation was recorded using a Photosonics 1B movie camera at 1000 pps. The synchronization of the pressure trace and the movie was obtained with an initial event strobe and timing marks on the film and light beam oscillograph trace. The analysis techniques used for this test are detailed in Appendix A of Volume II.

3.1.3 Structural Tests

Three tissue types which required testing in what could be termed a structural mode were the intercostal muscles, the intervertebral ligaments and the lungs. Each test had its own unique requirements as described below.

The intercostal muscles are short muscles lying between pairs of ribs. Due to the short length and the insertion of the muscle to adjacent ribs it was decided to test the muscle and its ribs as a unit. Short segments (about 1/2 to 3/4 inches in length) were cut along the ribs and the rib segments were used as the means of loading the muscle by pinning the rib segments into grips. The load was measured as in the uniaxial tension tests. The strain in the muscle was measured by grip displacement and was calculated as an extension ratio based on the initial rib to rib spacing.

The intervertebral ligament test required the design of a specimen holding fixture which would allow a specimen consisting of a sectioned vertebral column of two vertebral bodies and the associated rib, to be loaded in three directions in sequence. In order to measure the three-dimensional motion of the rib relative to the vertebral column, a target attached to the rib was used. This target system also served to apply the load to the specimen. The load was applied to the target fixture through a C-ring attached to a flexible wire from the load cell. A 45° mirror was used to obtain the second

view of the target system to provide complete three dimensional information. The motion of the target was photographically recorded. This system was used only in the Instron for tests.

The determination of the mechanical properties of the tissue of the lung presented many difficult technical problems. In the in vivo state, lung tissue is filled with air and is highly compressible. However, when an excised sample of lung tissue with a shape suitable for materials property testing is tested in vitro, the tissue is no longer air filled and thus is not representative of the in vivo state. This problem has plagued investigators in lung mechanics for many years.

The mathematical model of the thorax being developed at FIRL has characterized the lung as an elastic foundation which interacts accordingly with organs such as the heart during impact. In view of this simplified model of the lung, it would appear that a lung structural test would be of greater use in supplying data for the modeling effort. The lung structural test was performed on Rhesus monkeys. The experiment included in vivo and post mortem tests.

The monkey was anaesthetized with an I.V. injection of Sodium Pentobarbitol (300 mg). A tracheostomy was performed and a Harvard Respirator connected to provide respiration with a tidal volume of 30 ml, at a rate of 30 breaths per minute and an expiratory back pressure of 5 cm of water. The skin over the left half of the thoracic cage was then removed and half an inch portion of the fifth rib and surrounding musculature cut out to expose the middle portion of the pulmonale lobus medium. The monkey was placed on a specifically designed adjustable table and positioned so that the exposed lung surface could be impacted laterally with probe. The probe (3 inches length), in series with a Kistler 931 A load link, was mounted

on an Unholtz Dickie linear shaker. A linear accelerometer was also mounted on the shaker head and the two signals were added in a differential amplifier. With appropriate balancing of the signals, the inertial effect was cancelled out from the load cell output. The probe travel was measured optically by a Physitect GAGE-it unit by measuring the relative movement of the flaps mounted on the shaker head and base. Two types of tests were run with this apparatus; single pulse rapid load-unload tests and driving-point impedance sweeps.

4.0 TISSUE SOURCES AND TESTING PRIORITY

The human tissues tested in this program were fresh unembalmed tissues obtained at autopsy at the Veterans Administration Hospital in Ann Arbor. The material specimens were tested as soon as they were obtained, or in cases where this was not possible, they were stored in refrigerated physiological saline solution until used. Specimens were obtained from 13 individuals. It had been anticipated at the beginning of the program that more donors than the 13 would have been available. This did not occur for two reasons. The first was an unusual reduction in the number of autopsies performed during parts of the testing phase of the program. A more significant reason however, is that at the VA hospital the cause of death and gross pathologic diagnoses are such that in almost every case complications of one thoracic organ or another are involved. In many autopsies the situation was such that no suitable tissues could be obtained due to severe involvement of the thoracic organs in the pathology of the subject. Thus, only a fraction of the autopsies performed at the VA Hospital produced any suitable samples at all. The primary tissues studied in the program (ordered in decreasing importance to the FIRL modelling) are:

1. Intercostal muscle
2. Cardiac muscle (left ventricle)
3. Aorta
4. Pericardium
5. Lungs
6. Diaphragm
7. Vertebral ligaments
8. Esophagus
9. Trachea and Bronchi

In addition to human tissue samples, Rhesus monkey tissue samples of high priority tissues were also tested in the program. The monkey tissues were obtained from animals used on other HSRI studies. A limited number of live Rhesus monkeys were used in this program to study the lung structural characteristics.

5.0 SUMMARY AND CONCLUSIONS

5.1 GENERAL PROGRAM SUMMARY

This program has produced a bibliography of over 500 entries on the subjects of soft tissue mechanical properties and test techniques. The majority of these articles have been abstracted and are presented in Volume III of this report.

The knowledge obtained in performing the literature search has been incorporated into the experimental program carried out in this project. Test techniques on soft tissue testing have been specially adapted to the needs of the experimental program and unique test equipment developed to allow high speed dynamic testing of tissue samples in uniaxial tension and biaxial tension. The data produced in this project on the dynamic mechanical properties of thoracic tissues of the human and the Rhesus monkey are unique and represent

the first time that these tissues have been characterized at strain rates comparable to those produced in thoracic trauma associated with automotive accidents.

While the data presented here by no means represent a complete description of the tissues tested, either in the statistical sense or in the continuum mechanics sense, the data do represent a definitive first step in characterizing the dynamic behavior of these tissues and their failure mechanisms.

5.1.1 Test Results Summary

Examination of the test results reveals that the stress-strain response of the tissues varies over a wide range. In order to summarize the response for each tissue to provide average response curves the following reduction technique was used. First, it was noted that the major cause of the variability in response in most tissues was the different extensibility exhibited by each tissue in the low stress region. Following the lead of other investigators in soft tissue research (1,2) this high extension, low stress region of the curves was treated as a region separate from the stiffer high stress response region, since, in many cases, the pathological conditions found in these tissues manifests itself primarily in the low stress region (e.g., in atherosclerosis (hardening of the arteries) the main effect is the loss of this low stress extensibility while high stress response appears to be relatively unaffected). In the analysis of the uniaxial tension test data, an arbitrary stress level of 20 psi was used to define the upper limit of the low stress region (this corresponds to a typical wall stress in the aorta under physiological conditions). All dynamic stress-extension ratio curves for a particular tissue and loading direction were compared above this stress level by graphically shifting the curves to the $\lambda = 1.00$ point. This shift produced response curves which were generally quite similar. (This is most

likely due to the fact that the high stress response of the tissues is controlled by the oriented collagenous connective tissue present in the tissue. This orientation process occurs during the low extension.) In order to produce a representative curve for the tissue response, the resulting shifted high stress curves were averaged by dividing each curve into four proportional regions represented by the quarter point, the midpoint, the three quarter point and the failure end point. The resulting stress and extension ratio values for those four points were respectively averaged for all the curves of a given type and the average curve plotted. Similarly, the extension ratios for each curve at the 20 psi level were averaged and an average low stress region added graphically to the average high stress response.

The resulting average dynamic response curves for human intercostal muscle, cardiac muscle, aorta, pericardium, diaphragm and esophagus are shown in Figures 4, 5, 6, 7, 8, 9, 10 and 11. The static response of the intervertebral ligaments is shown in Figure 12. Similar curves for Rhesus monkey aorta and esophagus are presented in figures 13 and 14. The assumption of incompressibility was checked on many of the tissue tests when good values of thickness extension ratios could be obtained and, in general, the product of the three principal extension ratios was within 10% of unity (i.e., $\lambda_1 \lambda_2 \lambda_3 = 1$ implies incompressibility). In view of the limited statistical nature of the data obtained on any one tissue and the limited number of different strain rates used in the testing program, it is felt that any attempts to fit a constitutive relation to describe the behavior of any of the tissues would be ill-advised. Only with a more detailed long term study of each tissue could a statistically valid constitutive relation be generated. For the purposes of providing data for the FIRL modelling program, it is felt that the tabular summary data presented in Volume II are most appropriate at this time.

Figure 4. Summary Curve of Average Dynamic Response of Human Intercostal Muscle

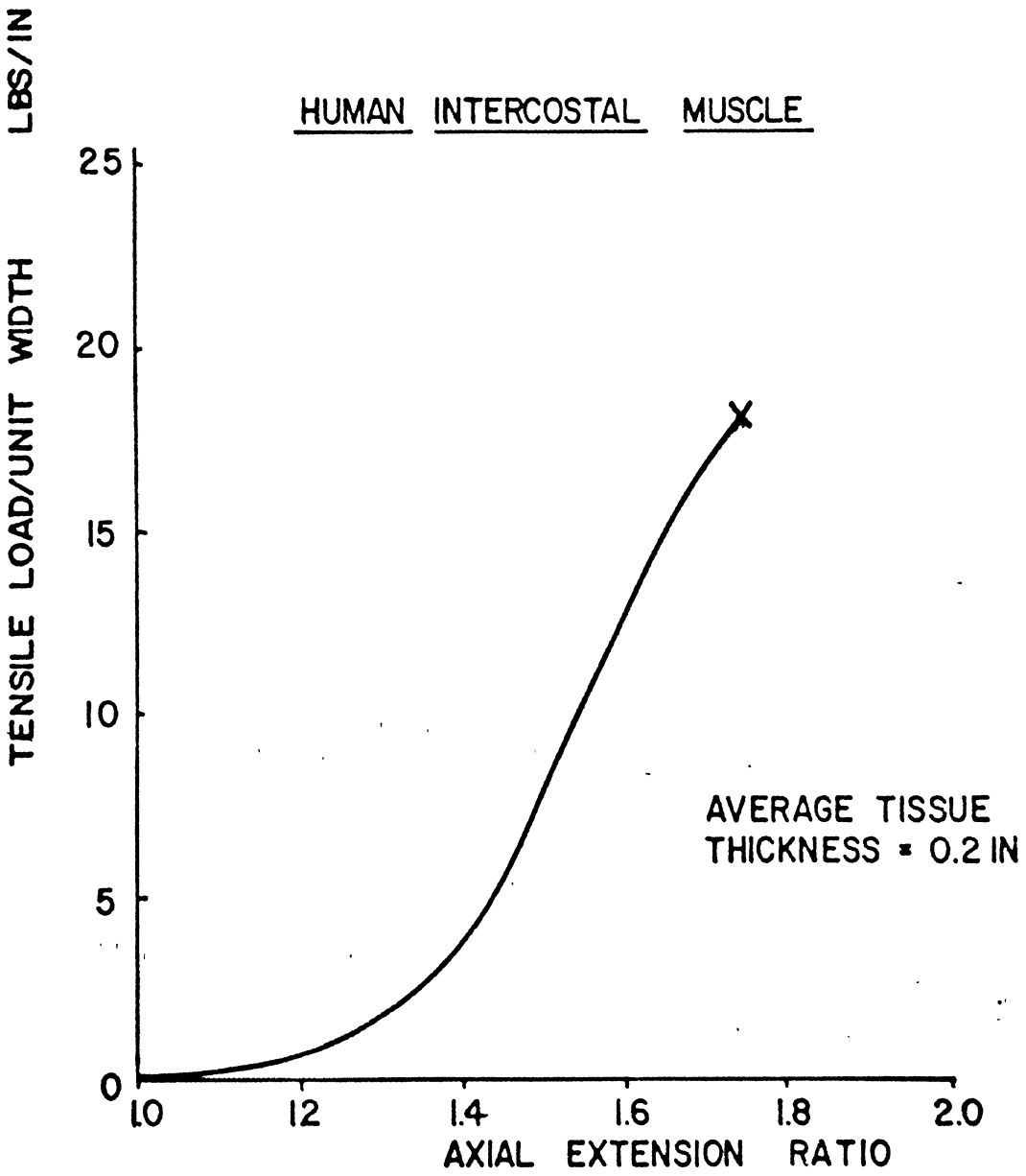

HUMAN INTERCOSTAL MUSCLE
TENSILE LOAD/UNIT WIDTH LBS/IN
25
20
15
10
5
0
1.0
1.2
1.4
1.6
1.8
2.0
AXIAL EXTENSION RATIO
AVERAGE TISSUE
THICKNESS = 0.2 IN

Figure 5. Summary Curves of Average Dynamic Responses of Human Cardiac Muscle

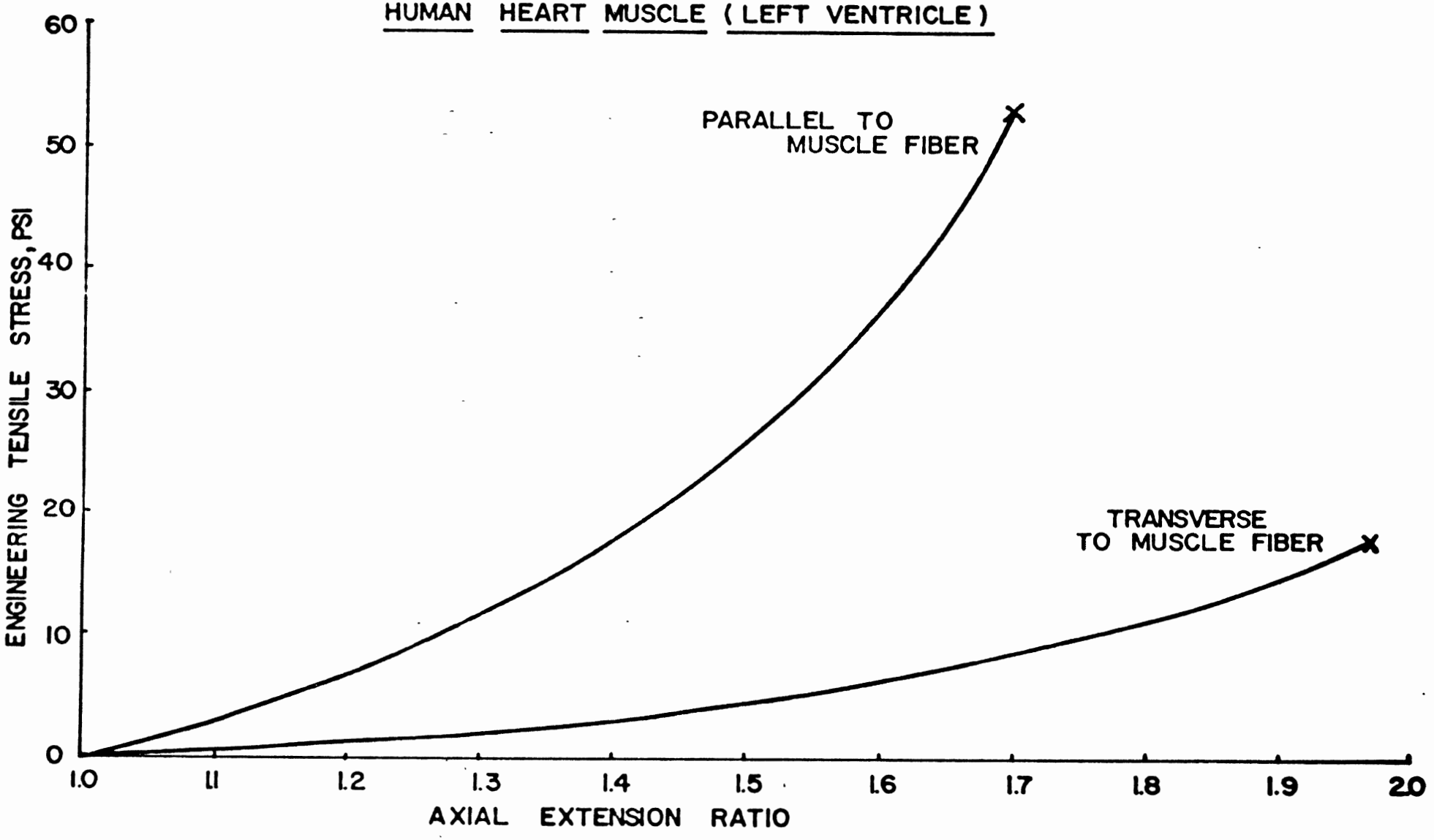

HUMAN HEART MUSCLE (LEFT VENTRICLE)
PARALLEL TO MUSCLE FIBER
TRANSVERSE TO MUSCLE FIBER
ENGINEERING TENSILE STRESS, PSI
60
50
40
30
20
10
0
1.0
1.1
1.2
1.3
1.4
1.5
1.6
1.7
1.8
1.9
2.0
AXIAL EXTENSION RATIO

Figure 6. Summary Curve of Average Dynamic Response of Human Aorta in the Longitudinal Direction

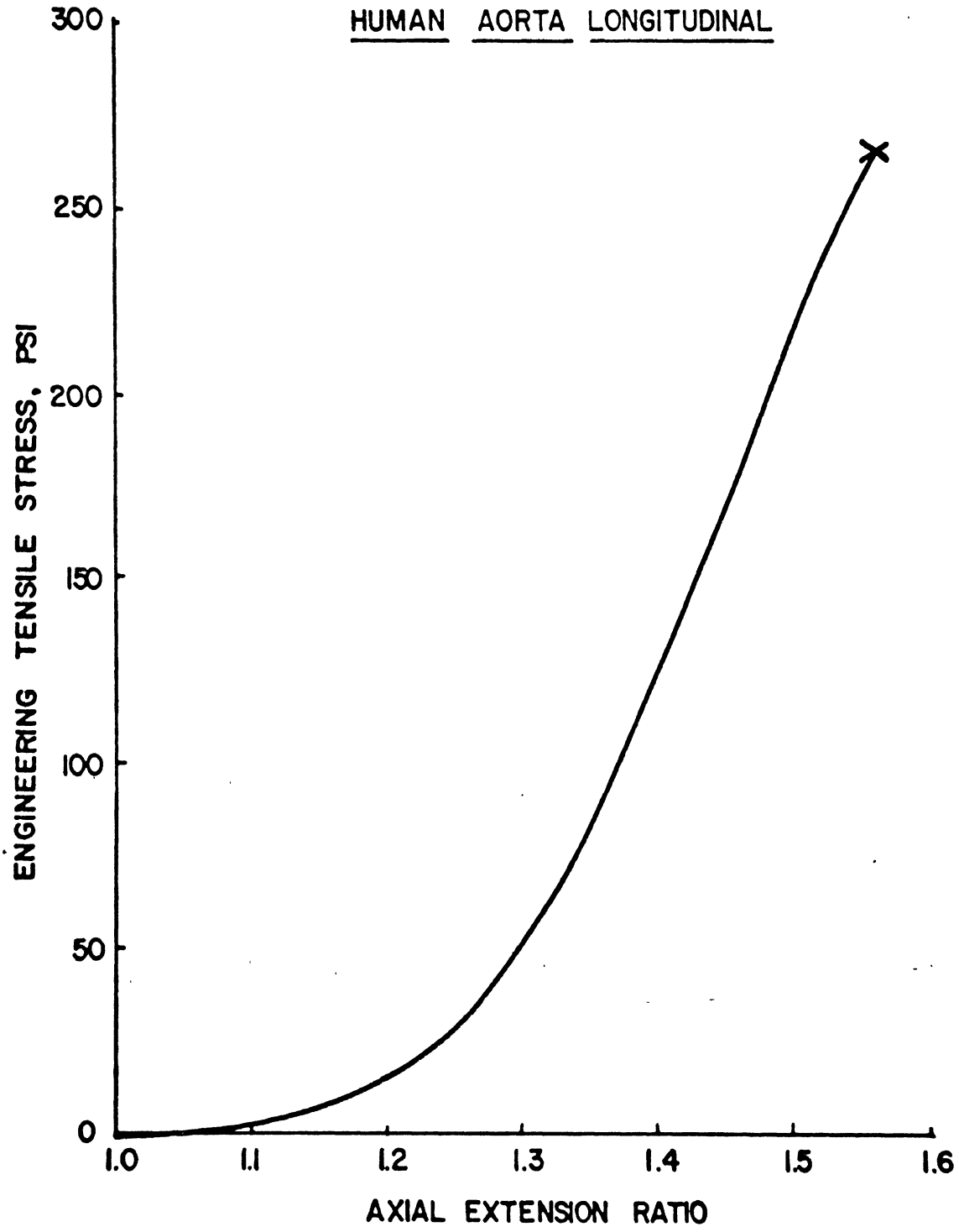

HUMAN AORTA LONGITUDINAL
300
250
200
150
100
50
0
ENGINEERING TENSILE STRESS, PSI
1.0
1.1
1.2
1.3
1.4
1.5
1.6
AXIAL EXTENSION RATIO

Figure 7. Summary Curve of Average Dynamic Response of Human Aorta in the Transverse Direction

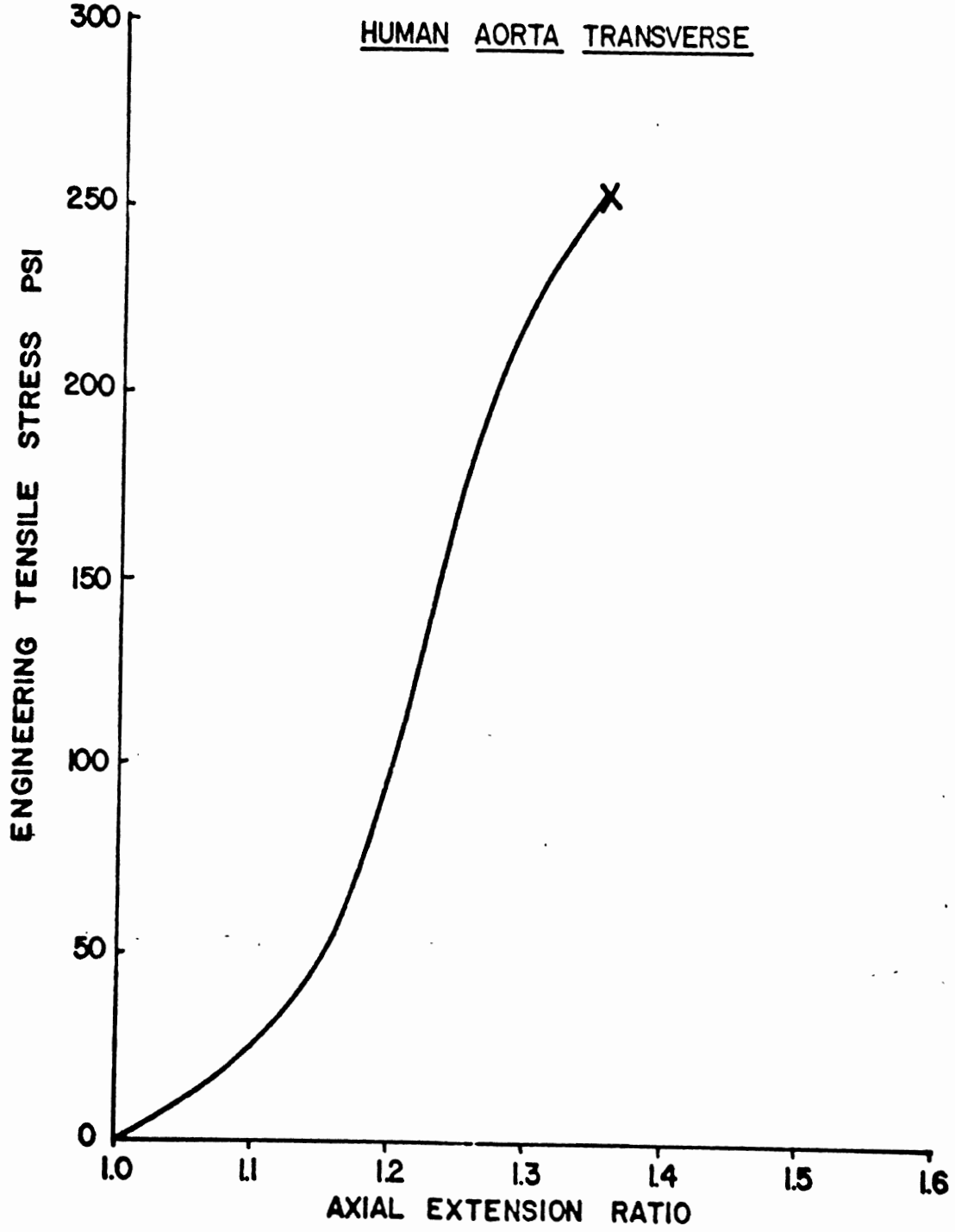

HUMAN AORTA TRANSVERSE
ENGINEERING TENSILE STRESS PSI
300
250
200
150
100
50
0
1.0
1.1
1.2
1.3
1.4
1.5
1.6
AXIAL EXTENSION RATIO

Figure 8. Summary Curve of Dynamic Response of Human Pericardium

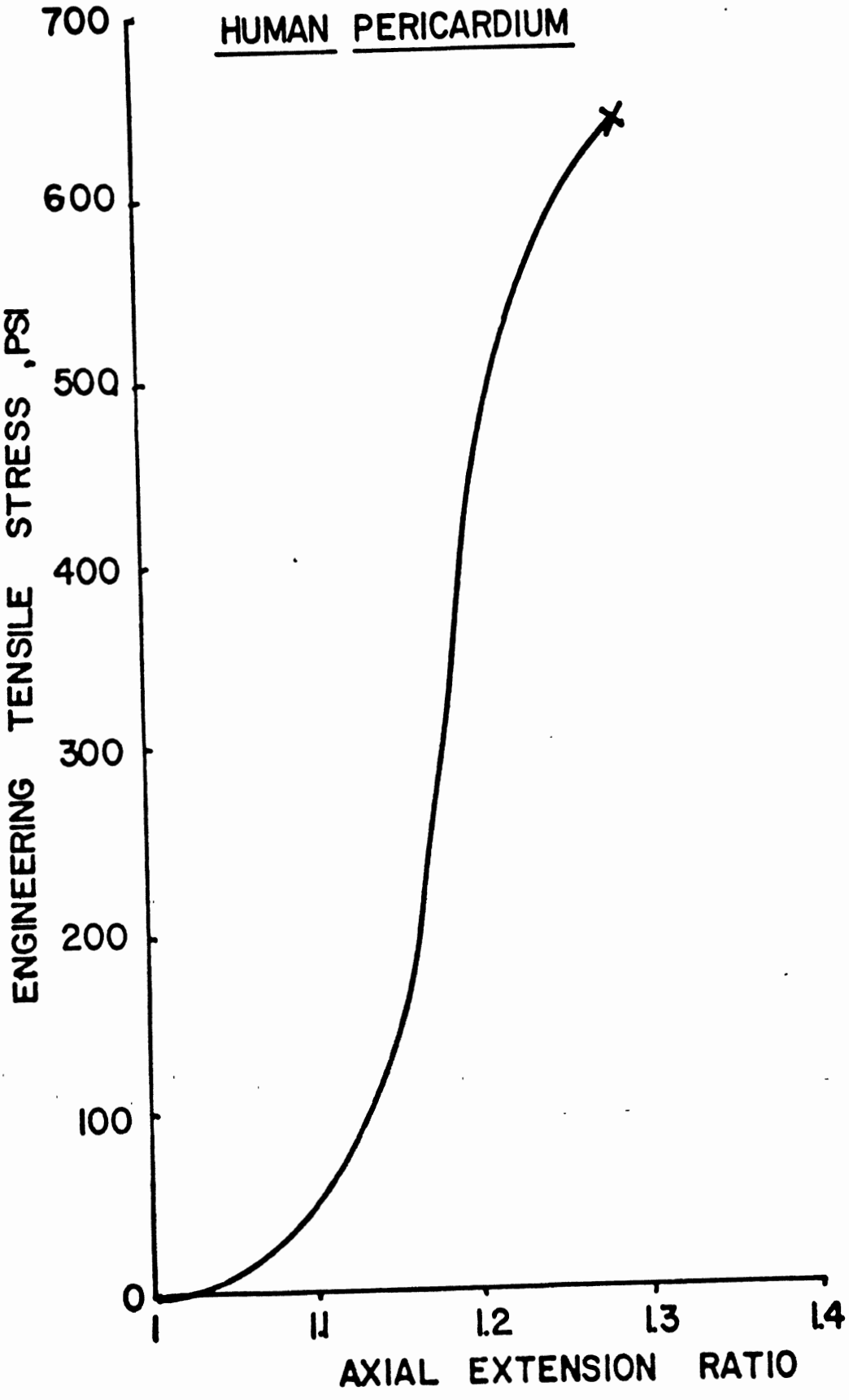

HUMAN PERICARDIUM
700
600
500
400
300
200
100
0
ENGINEERING TENSILE STRESS, PSI
1
1.1
1.2
1.3
1.4
AXIAL EXTENSION RATIO

Figure 9. Summary Curve of Average Dynamic Response of Human Diaphragm Parallel to Muscle Fibers

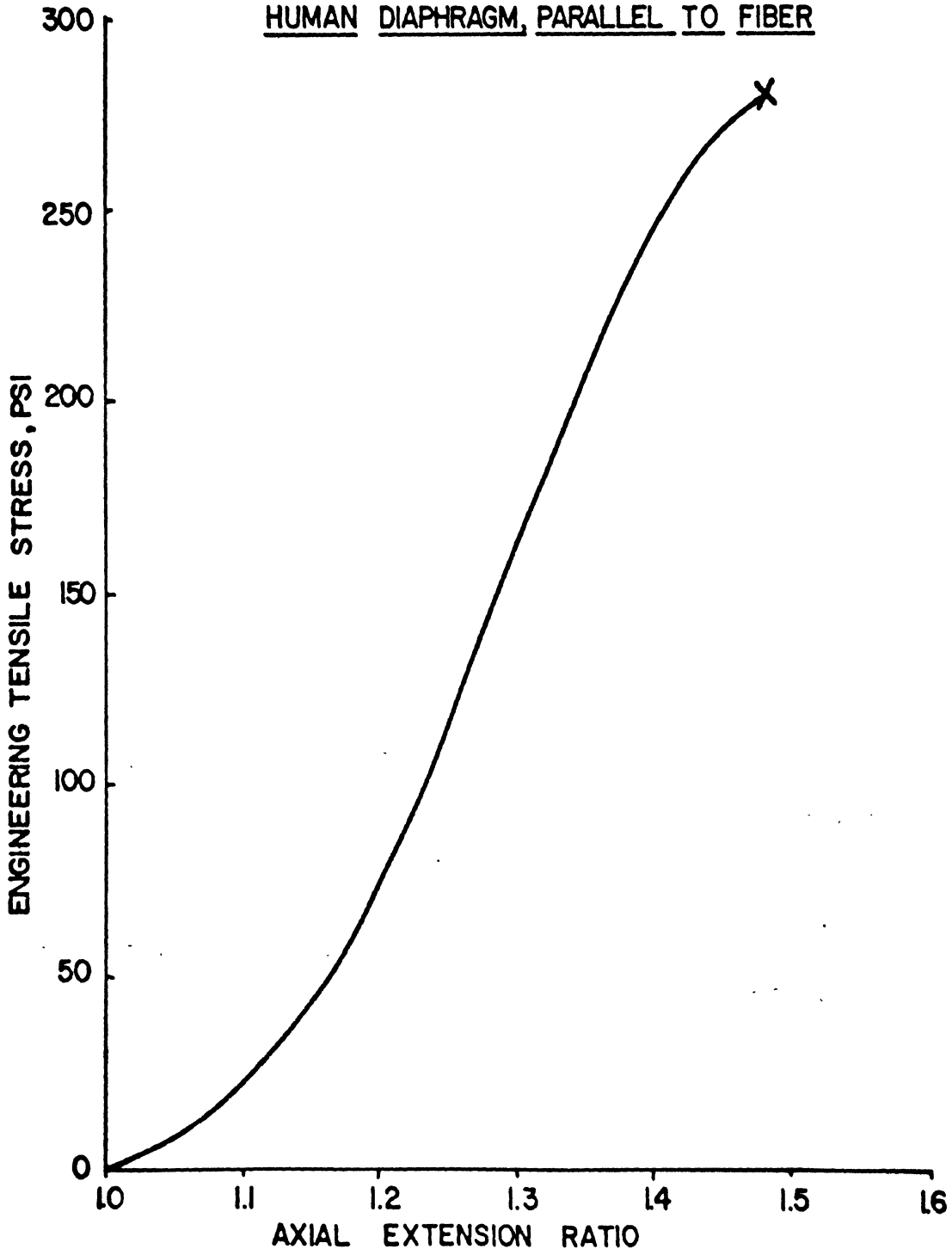

HUMAN DIAPHRAGM, PARALLEL TO FIBER
300
250
200
150
100
50
0
ENGINEERING TENSILE STRESS, PSI
1.0
1.1
1.2
1.3
1.4
1.5
1.6
AXIAL EXTENSION RATIO

Figure 10. Summary Curve of Average Dynamic Response of Human Diaphragm Across Muscle Fibers

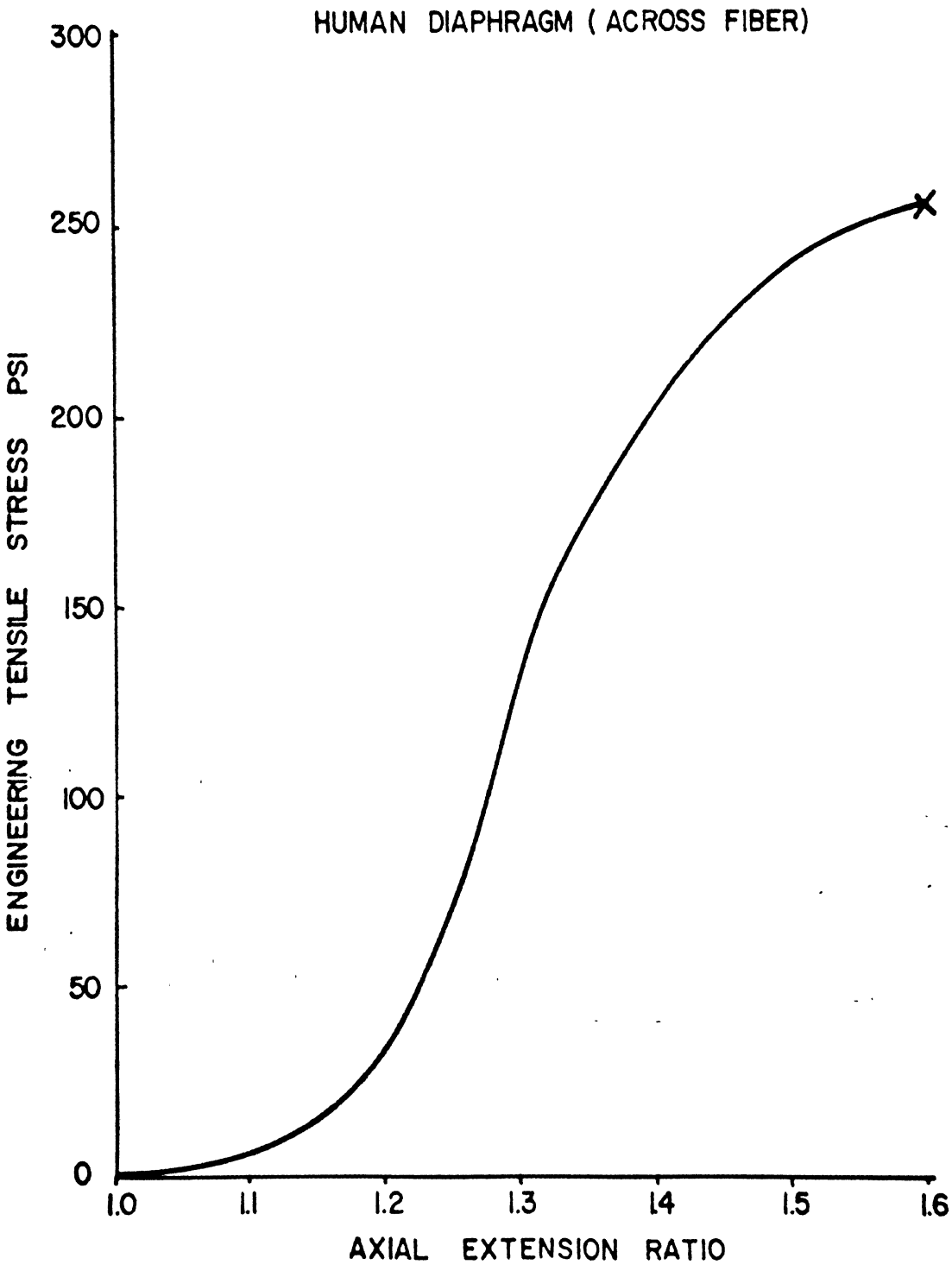

HUMAN DIAPHRAGM (ACROSS FIBER)
ENGINEERING TENSILE STRESS PSI
300
250
200
150
100
50
0
1.0
1.1
1.2
1.3
1.4
1.5
1.6
AXIAL EXTENSION RATIO

Figure 11. Summary Curve of Average Dynamic Response of Human Esophagus in the Longitudinal Direction

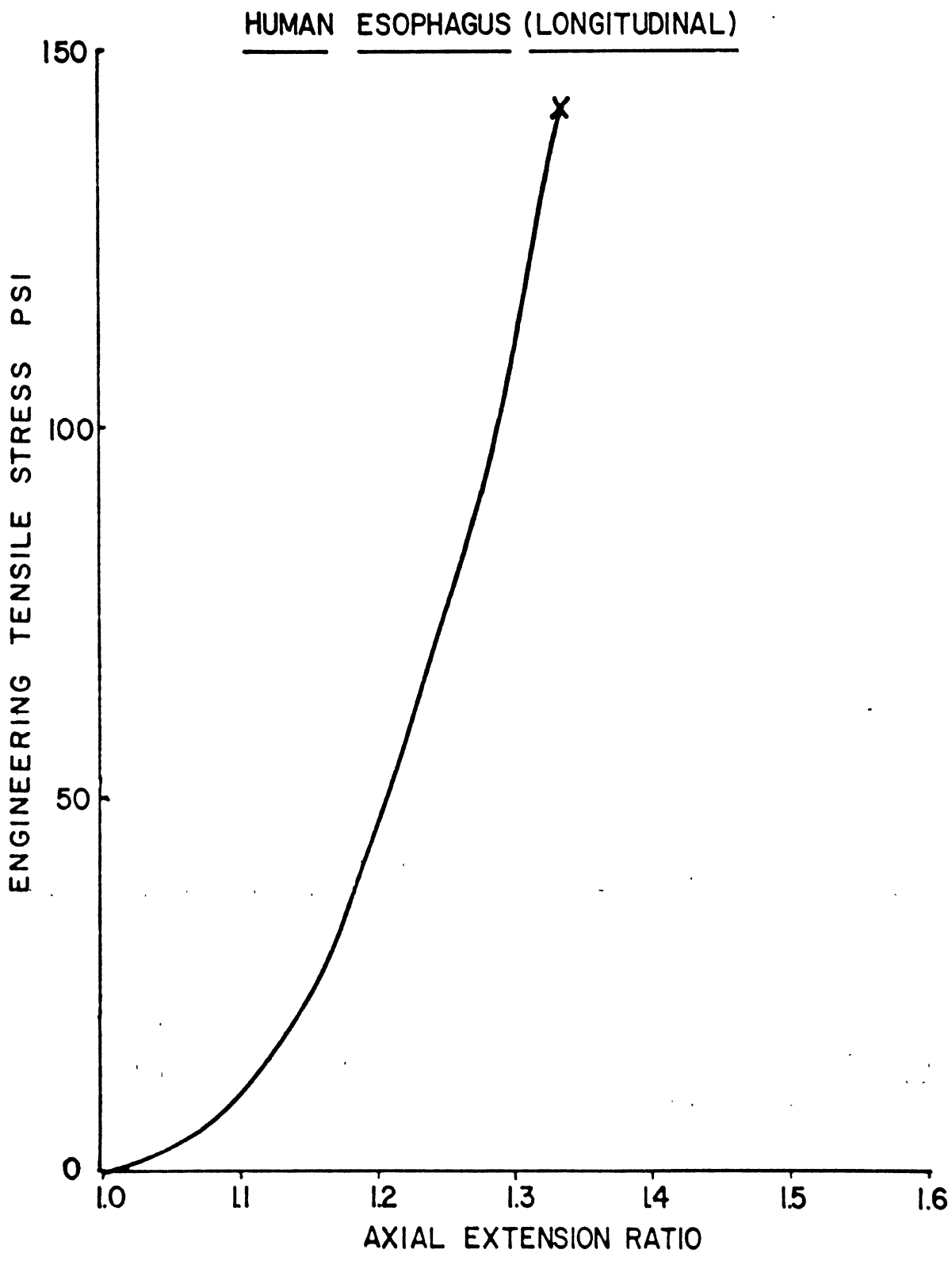

HUMAN ESOPHAGUS (LONGITUDINAL)
150
100
50
0
ENGINEERING TENSILE STRESS PSI
1.0
1.1
1.2
1.3
1.4
1.5
1.6
AXIAL EXTENSION RATIO

Figure 12. Human Intervertebral Ligament Test Results, Costo-vertegral Joint Response to Superior-Inferior (Downward) Loading

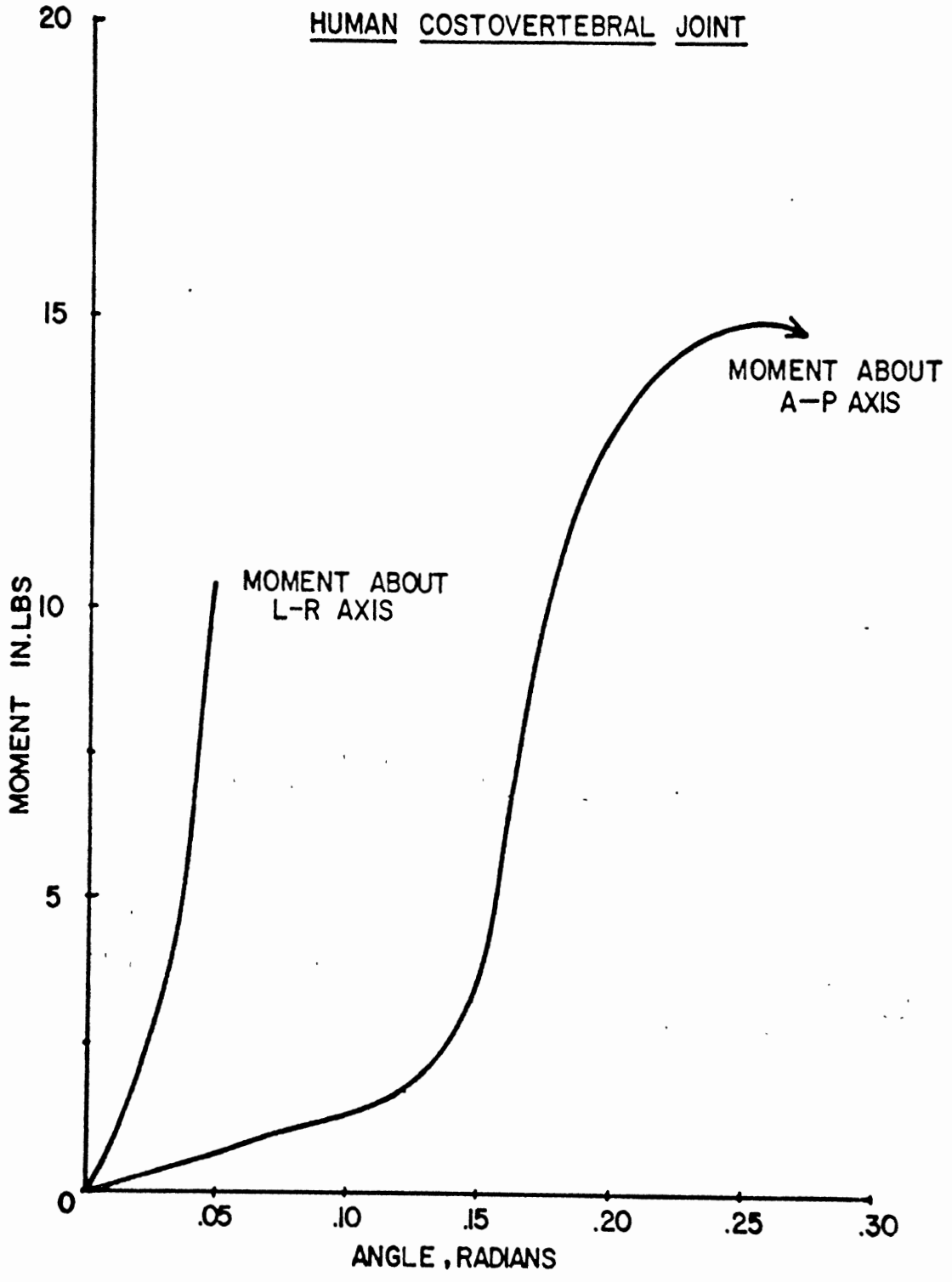

HUMAN COSTOVERTEBRAL JOINT
20
15
10
5
0
MOMENT IN.LBS
MOMENT ABOUT
L-R AXIS
MOMENT ABOUT
A—P AXIS
.05
.10
.15
.20
.25
.30
ANGLE, RADIANS

Figure 13. Summary Curve of Average Dynamic Response of Rhesus Monkey Esophagus in the Transverse Direction

RHESUS ESOPHAGUS, MEAN OF DYNAMIC TESTS

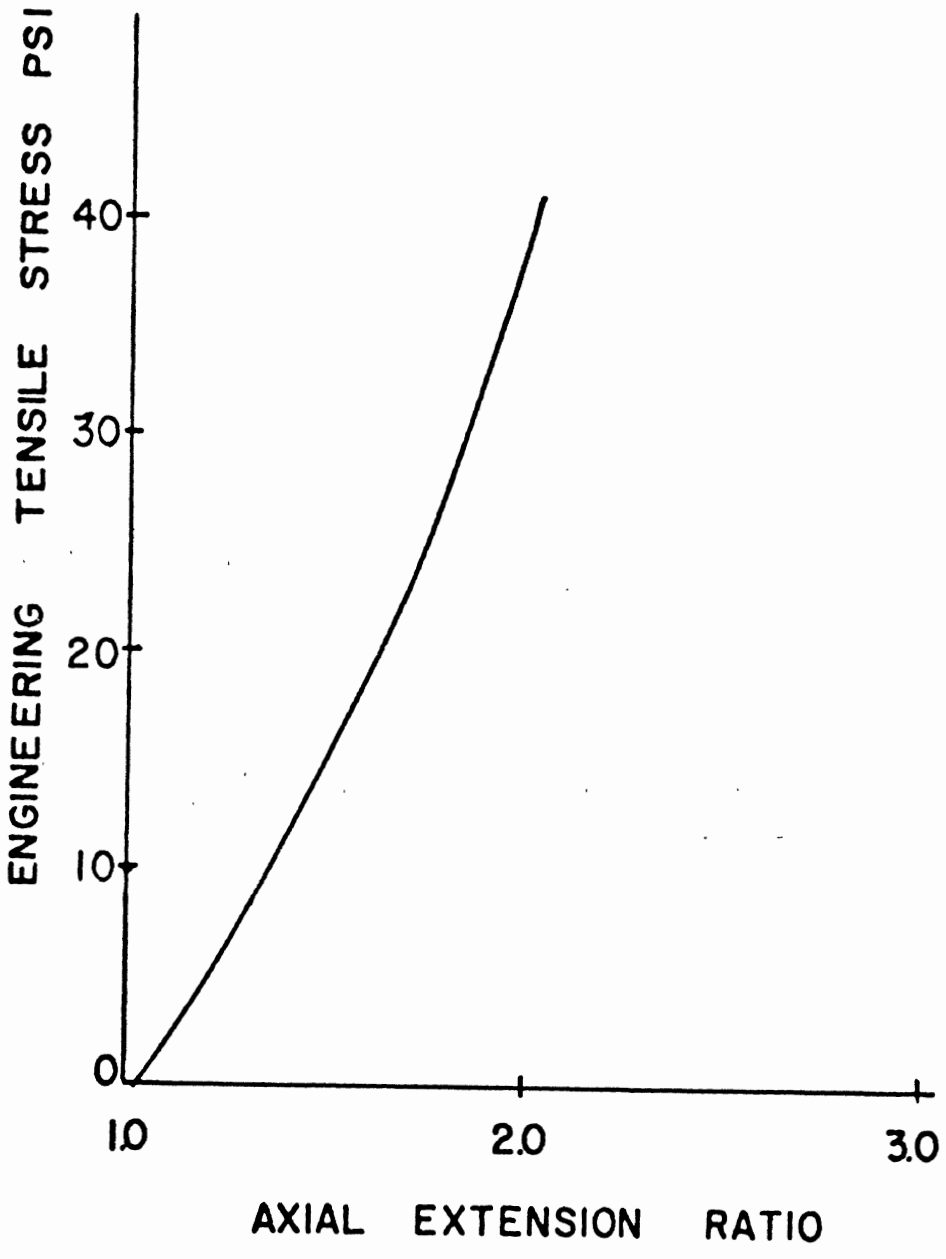
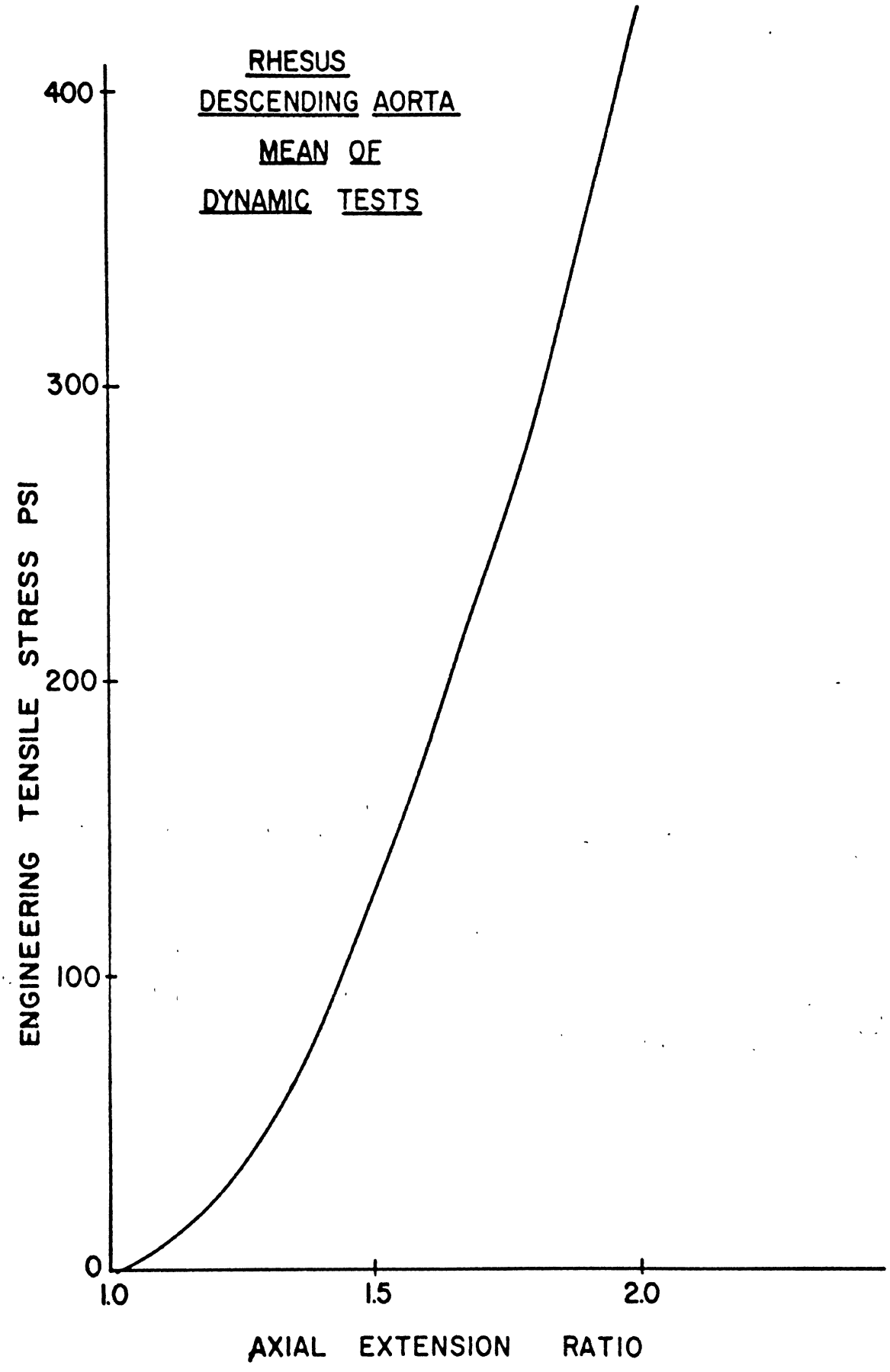


Figure 14. Summary Curve of Average Dynamic Response of Rhesus Monkey Descending Aorta in the Transverse Direction


RHESUS
DESCENDING AORTA
MEAN OF
DYNAMIC TESTS
400
300
200
100
0
ENGINEERING TENSILE STRESS PSI
1.0
1.5
2.0
AXIAL EXTENSION RATIO

Comparison of the dynamic response of the tissues tested with the static tests performed in the program and with the static data of Yamada (13) indicates that although the dynamic stresses produced in the tissues are as much as twice as great as those produced statically, the failure strains tend to be similar. This leads to the conclusion that the most appropriate failure mechanism theory for the tissues studied in the program would be a maximum tensile strain theory of failure. This is born out in the biaxial tension tests of the human diaphragm where the failure strains were similar to the uniaxial tension failure strains. It should be noted, however, that the pathological state of many of the tissues tested reflects the older age group (average age 69.5 years). The static failure strains obtained in the program are comparable to those obtained in the older age groups reported by Yamada (13) and thus, for younger people larger failure strains would be expected.

Lung tests were performed only on Rhesus monkey lungs. The tests were performed both in vivo, post mortem in situ and in vitro. In all cases the lung was inflated by a respirator. The tests which consisted of single pulse loading of the lung showed very pronounced rate effects where a 50 Hz equivalent pulse produced almost five times the force of the 5 Hz equivalent pulse with the 50 Hz force peak corresponding to maximum probe velocity rather than maximum deflection. The driving point impedance tests results on two monkeys also show the viscous nature of the lung. At low frequencies (below approximately 20-30 Hz) the response is spring-like while above these frequencies large damping effects predominate. The general response of the lung in vivo and in vitro is similar although specific details vary with probe diameter and animal.

5.2 CONCLUSIONS AND RECOMMENDATIONS

The results of this program to obtain information on the mechanical proper-

ties of thoracic tissues at high strain rates indicate that the dynamic response of such tissues is considerably different from the static response in terms of stress, but that the strain response is more dependent upon pathological condition of the tissues than upon strain rate. While the scope of the program was not large enough to permit a complete statistical and continuum mechanics representation of each of the nine tissue types studied, the information developed in the program will be of direct use in finite element modelling of the thorax and in helping to understand more clearly the injury mechanisms associated with thoracic trauma. In this connection it would appear that a maximum tensile strain theory of failure would be most appropriate to describe the failure mechanisms observed in many of the tissues studied in the program.

The work carried out in the program must be considered as an initial step towards the complete characterization of the response and failure of thoracic organ tissues. Work should be continued in the future to include a more complete quantification of anisotropy effects and multiaxial load response, to include a more complete statistical quantification of variations in tissue properties due to sex, age and/or physical condition; and to include the quantification of the effects of physiological deformation states on tissue response and injury modes. Consideration should also be given to extending this type of study to other injury prone areas of the body.

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