RESEARCH and

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LIFE SCIENCES DIVISION

ANNUAL TECHNICAL REPORT

1 November 1966 through 31 October 1967

Contract NASw-1313

THE INVESTIGATION OF VERTEBRAL INJURY SUSTAINED DURING AIRCREW EJECTION

Phase 2a: Basic Science Experimental Design and Investigation of Dynamic Characteristics of Vertebral Columns Considered as an Engineering Structure.

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SUMMARY

This report summarizes research conducted on isolated human vertebrae by Technology Incorporated, Life Sciences Division, during the year 1 November 1966 to 31 October 1967. The dynamic strength was determined for 2 vertebrae - 1 disc sets of various categories. Details of the basic science experimental design used in this continuing investigation of dynamic characteristics of the human vertebral column as an engineering structure are included. Recent studies include moisture and ash determinations of the segments tested. The results are presented as a series of figures, tables and discussions.

SECTION I

INTRODUCTION

Justification of Research Effort

Since the introduction of forcible ejection from disabled aircraft during World War II, there has been a continued high incidence of back injury among otherwise successful escapes.

This report describes some of the results of a continuing examination of the mechanical response of the human vertebral column. Ultimately, this study should allow specification of safe parameters for successful escape from disabled high speed aerospace vehicles. The experimental approach may serve as a model for solution of similar problems in automotive crash protection.

In the process of ejection from a disabled high performance aircraft, the pilot initiates the detonation of a controlled burning propulsive device which causes the seated occupant to be forcibly catapulted from the craft and displaces him far enough to allow clearance of all structures.

Studies by Koochembere[1] and others have shown that a peak acceleration of 17 - 21 g for periods of 0.15 to 0.25 seconds will allow this escape to take place. The acceleration forces resulting are applied through the seat pan to the buttocks and pelvic structures, thence to the sacrum and spine.

Compression of the spine results, and depending on the spine's relation to the acceleration vector, end-plate fractures, fractures of the anterior vertebral bodies and herniation of nucleus pulposus of the intervertebral discs into the body of adjacent vertebrae have resulted. Severity of injury has varied from brief loss of flying clearance to permanent disability. The usual sites of fracture have ranged from the second thoracic to the fifth lumbar vertebrae and coccyx. In American aircraft, the largest proportion of injuries occurs in the small of the back--at T12--with approximately equal incidence from T8 to L1. Why fracture should occur more often at certain specific sites than at others remains obscure. An examination of the static strengths of individual vertebrae (Table 1) does not help, even when related to the per cent of body weight borne.

TABLE 1: COMPRESSIVE STRENGTH VERSUS VERTEBRA (AFTER RUFF[2])

Vertebra	Compressive Strength Range kg	lb.	% of Body Weight Borne
T-8	540-640	1188-1408	33
T-9	610-720	1342-1584	37
T-10	660-800	1452-1760	40
T-11	720-860	1584-1892	44
T-12	690-900	1518-1980	47
L-1	720-900	1584-1980	50
L-2	800-990	1760-2178	53
L-3	900-1100	1980-2420	56
L-4	900-1200	1980-2640	58
L-5	1000	2200	60

Methods of Investigation

Since the static strengths of individual vertebrae do not show any marked lowering at the levels known to fracture, two possible fields of inquiry suggest themselves: several groups have studied the characteristics of the input acceleration time history, alignment of spine relative to the acceleration vector, and aspects of restraint systems;[3] a few others have examined the dynamic characteristics of the individual spinal segments and the spine as a whole.[4] The work reported here deals with the second area.

Mathematical Modeling

In order for any materials testing program to be meaningful, a definite purpose must be had for the experimental findings. To this end, Toth[5] has devised a multiple degree of freedom model of the human spine. In this model arbitrary spring constants, masses and viscous elements are combined mathematically in a set of nonlinear differential equations.

The model assumes uniaxial stress and absence of bending. The axial loads on each vertebral element can be predicted, given the input acceleration time history, spring constants, viscous coefficients, and masses of elements. These can then be compared with yield points, breaking strengths, or other indices of failure if such are known.

It should be emphasized that the equations used nonlinear coefficients for spring constant and viscous damping, but these were derived from static measurements from published literature only. These coefficients, therefore, must be determined dynamically for the population at risk. Equally important is a knowledge of the rate dependence, if any, of these structures.

METHODS AND MATERIALS

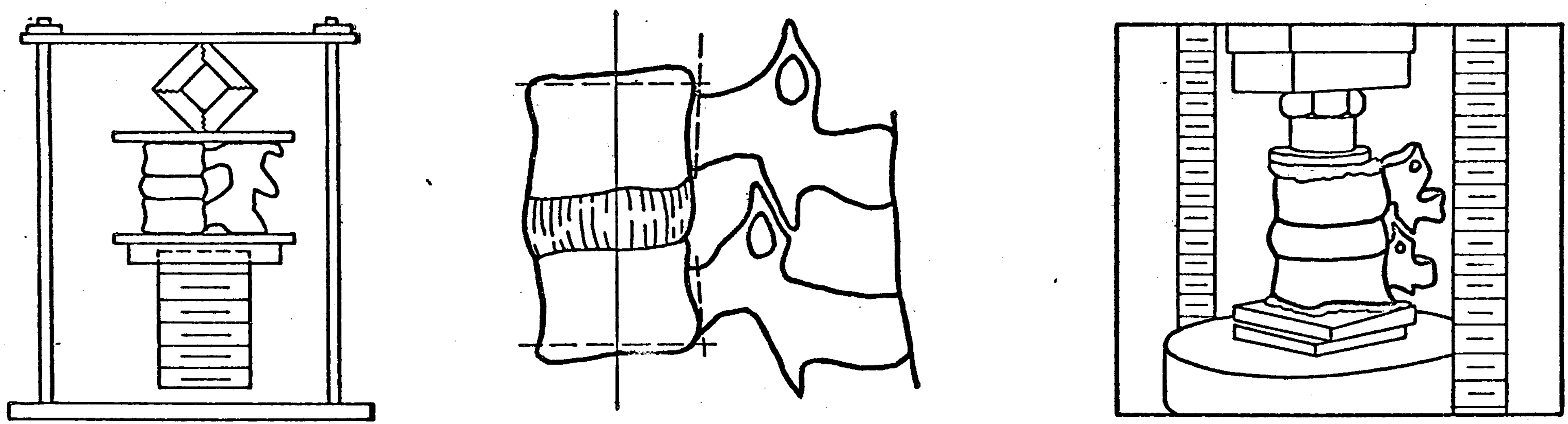
Portions of human vertebral columns, fresh and embalmed, were obtained from autopsy sources. The age of specimens was centered in two areas: 18-45 years (fresh specimens), and greater than 50 years (preserved specimens). For the preserved specimens, the usual cause of death was arteriosclerosis; for fresh frozen and fresh unfrozen specimens, trauma not involving the spine was the usual etiology of death. Vertebral columns were sectioned into two-vertebrae sets, hereafter called "segments." These consisted of two centra with spines intact and one intact intervertebral disc between. Each end thus represented a bare centrum.

Following dissection, the formalin preserved specimens were placed in a plastic bag half-filled with 10% formalin and incubated at 37°C for 24 hours. Fresh frozen specimens were allowed to thaw completely in a 37°C oven while kept moist with periodic physiological saline washes. The more recently acquired fresh unfrozen specimens were tested within one hour after removal at autopsy.

Each segment in all categories was mounted in an alignment jig, and the ends potted in quick-setting dental plaster* such that a pair of parallel surfaces, essentially perpendicular to the long axis of the vertebral bodies, resulted.

Figure 1 shows the test specimen preparation used, and is compared with preparations used by previous investigators. Preparations used by Perey[6] were tested statically without potting, but the posterior spines were partially cut off. The samples tested by Brown's[7] group were sliced top and bottom through centra and the spines removed entirely, as shown by the dashed lines. Potting material was used on each face of the specimen. No interruption of the integrity of the 2-bone plus intervertebral disc set resulted from the preparation used by our group; it should also be noted that the posterior spines were left intact.

* French's Impression Dental, Manufactured by Samuel H. French & Co., 475-477 York Ave., Philadelphia.



PEREY

BROWN, et/al

HIGGINS, et/al

Figure 1

METHODS OF
BONE PREPARATION

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Figure 2 shows a specimen mounted in a standard materials testing machine, along with the recording apparatus. This machine was modified by the addition of an accumulator and solenoid valve to provide higher strain rates, when desired. Physiological saline was continuously dripped on the specimen throughout the test procedure. Tests were then run at varying strain rates, in which the full capability of the modified testing machine was utilized. Four to six arbitrary settings of rate of deformation were used. Since this is not a constant strain rate machine, the actual strain rates were calculated from the time history of specimen compression recorded on an oscillograph.

Adjustable stops on the testing machine were set low enough to cause the deflection to remain within the straight line portion of the force-deflection curve. For preserved and fresh frozen bone, the order of testing was from slowest to fastest rate of loading in each case.

Providing the total load was kept below 500 lbs. (228 kg) repeated tests at various strain rates could be performed without significant change in the force-deflection read out for any machine setting. Later tests with fresh unfrozen bones were performed with randomly selected strain rates. Force-deflection data were plotted individually as functions of time on the CEC® ** oscillograph. Force-deflection curves were then plotted as X-Y graphs.

Prior to potting, the initial length was measured with calipers. The areas of the top and bottom free surfaces were determined planimetrically and averaged.

DATA REDUCTION

Measurement of the slope of the X-Y plot was accomplished at several points yielding stiffness values. These were subsequently normalized for initial length and area in order to compute the elastic moduli.

Strain rate was determined at the same point from the oscillographic record of deflection versus time.

Presentation of data for elastic modulus assumes the material is elastic over the range tested; calculation of strain rate assumes a linear deformation of the specimen with time.

** CEC Oscillograph Model 5-124A, Consolidated Electrodynamics Corp.

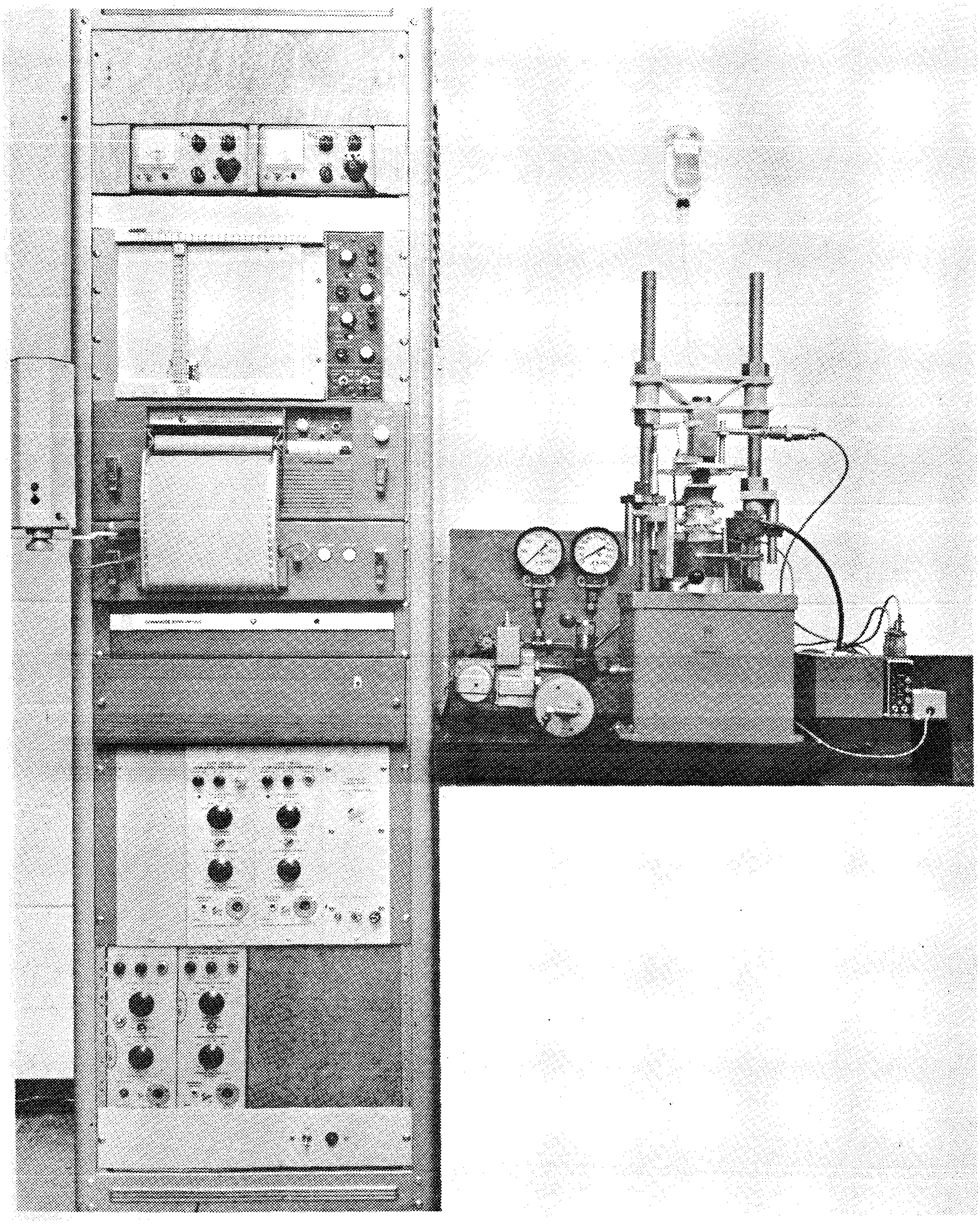

Figure 2

Load Vertebral Testing Machine
and Recording Equipment

RESULTS

Table 2 summarizes the data thus far collected.

For better orientation, Figure 3 shows force-deflection curves at the same machine rate setting for aluminum, nylon, dry pine wood and a typical vertebral bone segment. It can be seen that the elastic modulus for the vertebra falls considerably below that for any of the three inanimate materials. The strain rate, although different for each material at the same machine setting, is of the same order of magnitude. The reader should note the change of scale for bone deflection.

Figure 4 shows a typical set of force-deflection curves of fresh-unfrozen T12-L1 vertebrae from a 28 year old, 140 lb. white male who died from a gunshot wound of the head. Five such similar specimens in the age range of 18 to 40 years have been tested less than one hour following autopsy. Although our previous experience with preserved bones showed that serial loadings below 250 kg did not change the shape of the force-deflection curve, it was decided to test the fresh unfrozen specimens in random order, so that any effect of fatigue or early undetected fracture could be found.

For this specimen, the order of testing is shown in the upper left-hand corner with the elastic moduli and corresponding strain rate determined at the midpoint of each curve. It can be seen that the stiffness is, indeed, rate dependent.

Although the testing machine used is not a constant strain rate device, examination of the deflection-time traces revealed only a slight departure from linearity over the majority of the test.

Computation of elastic moduli from tangents taken at other portions of the curves demonstrated a considerable variation, although the strain rate at the same points varied little. Although the data are too few for a quantitative conclusion, persistence of this trend may indicate considerable non-linearity of the force-deflection response of this structure.

In order to explore the possible nonlinear elastic response further, the data were examined for elastic modulus as a function of total deflection. Figure 5 shows the force-deflection curves resulting from initial tests on fresh unfrozen T12-L1 vertebral segments at strain rates from .0020/sec to .0090/sec (Table 2). Figure 6 shows the elastic modulus derived from the Figure 5 curves plotted against total deflection of the segment. Of the curves shown, only one, 31-7M, shows a significant linear portion in the middle of the curve. The remaining segments show a generally increasing stiffness with increasing total deflection, regardless of strain rate.

TABLE 2: ELASTIC MODULUS, STRAIN RATE AND STIFFNESS OF VERTEBRAL SEGMENTS TESTED*

Specimen Number	Segment Tested	State of Segment	Elastic Modulus 10^6kg/m^2	Strain Rate/sec	Stiffness kg/mm
31-7M	T-12, L-1	Fresh Unfrozen	A 5	.0075	77
			B 14	.0082	241
			C 21	.0071	363
			D 24	.0074	409
			E 27	.0067	555
			F 27	.0090	422
33-7M	T-12, L-1	Fresh Unfrozen	A 8	.0034	218
			B 15	.0041	409
			C 21	.0043	559
			D 21	.0032	559
			E 21	.0034	577
			F 24	.0024	645
34-7M	T-12, L-1	Fresh Unfrozen	A 11	.0020	273
			B 19	.0036	486
			C 21	.0054	522
			D 25	.0054	645
			E 31	.0054	786
			F 27	.0074	695
35-7M	T-12, L-1	Fresh Unfrozen	A 7	.0444	173
			B 11	.0461	259
			C 15	.0444	363
			D 23	.0461	540
			E 34	.0461	818
			F 41	.0524	973
22-6M	L2-3	Fresh Frozen	53	.0010	1023
			53	.0010	1023
			53	.0011	1023
			59	.0011	1136
			59	.0011	1136
			65	.0011	1250
23-7M	T11-12	Fresh Frozen	60	.0005	1250
			60	.0005	1250
			60	.0005	1250
			60	.0005	1250
			60	.0007	1250
			60	.0007	1250

24-7M	L2-3	Fresh	55	.0009	1023
		Frozen	61	.0009	1136
			67	.0008	1250
			67	.0010	1250
			79	.0010	1477
			73	.0010	1364

TABLE 2: ELASTIC MODULUS, STRAIN RATE AND STIFFNESS OF VERTEBRAL SEGMENTS TESTED (CONT.)

Specimen Number	Segment Tested	State of Segment	Elastic Modulus $10^6 kg/m^2$	Strain Rate/sec	Stiffness kg/mm
4-6M	T8-9	Formalin Preserved	31	.0015	710
4-6M	T10-11	Formalin Preserved	24	.0014	639
4-6M	T12-L1	Formalin Preserved	60	.0016	1250
			60	.0014	1250
			60	.0015	1250
			60	.0014	1250
			60	.0013	1250
			60	.0013	1250
4-6M	L2-3	Formalin Preserved	37	.0008	1023
			37	.0008	1023
			37	.0009	1023
			47	.0009	1305
			47	.0009	1305
			47	.0010	1305
5-6M	T8-9	Formalin Preserved	44	.0006	964
			44	.0007	964
			44	.0007	964
			44	.0006	964
			44	.0005	964
			44	.0005	964
5-6M	T10-11	Formalin Preserved	54	.0009	1023
			61	.0009	1023
			67	.0008	1250
			73	.0009	1364
			73	.0010	1364
			79	.0010	1477
5-6M	T12-L1	Formalin Preserved	61	.0004	1023
			67	.0004	1136
			67	.0006	1136
			74	.0008	1250
			74	.0008	1250
			74	.0010	1250

5-6M	L2-3	Formalin Preserved	41	.0004	1136
			37	.0005	1023
			37	.0005	1023
			37	.0006	1023
			43	.0008	1191
			45	.0009	1250
6-6M	T10-11	Formalin Preserved	41	.0011	909
6-6M	T12-L1	Formalin Preserved	35	.0006	909
6-6M	L2-3	Formalin Preserved	30	.0005	757
7-6F	T8-9	Formalin Preserved	60	.0020	826
7-6F	T10-11	Formalin Preserved	21	.0012	405
7-6F	T12-L1	Formalin Preserved	35	.0013	508
7-6F	L2-3	Formalin Preserved	45	.0011	715
9-6F	T10-11	Formalin Preserved	39	.0009	671
9-6F	T12-L1	Formalin Preserved	31	.0006	694
9-6F	L2-3	Formalin Preserved	26	.0008	591
17-6F	T8-9	Formalin Preserved	27	.0016	757

* The multiple values shown are measurements made along the force-deflection curves and force-deflection time histories.

On fresh-unfrozen specimens, 31-7M through 35-7M, the capital letters preceding the elastic modulus refer to points on Figure 5.

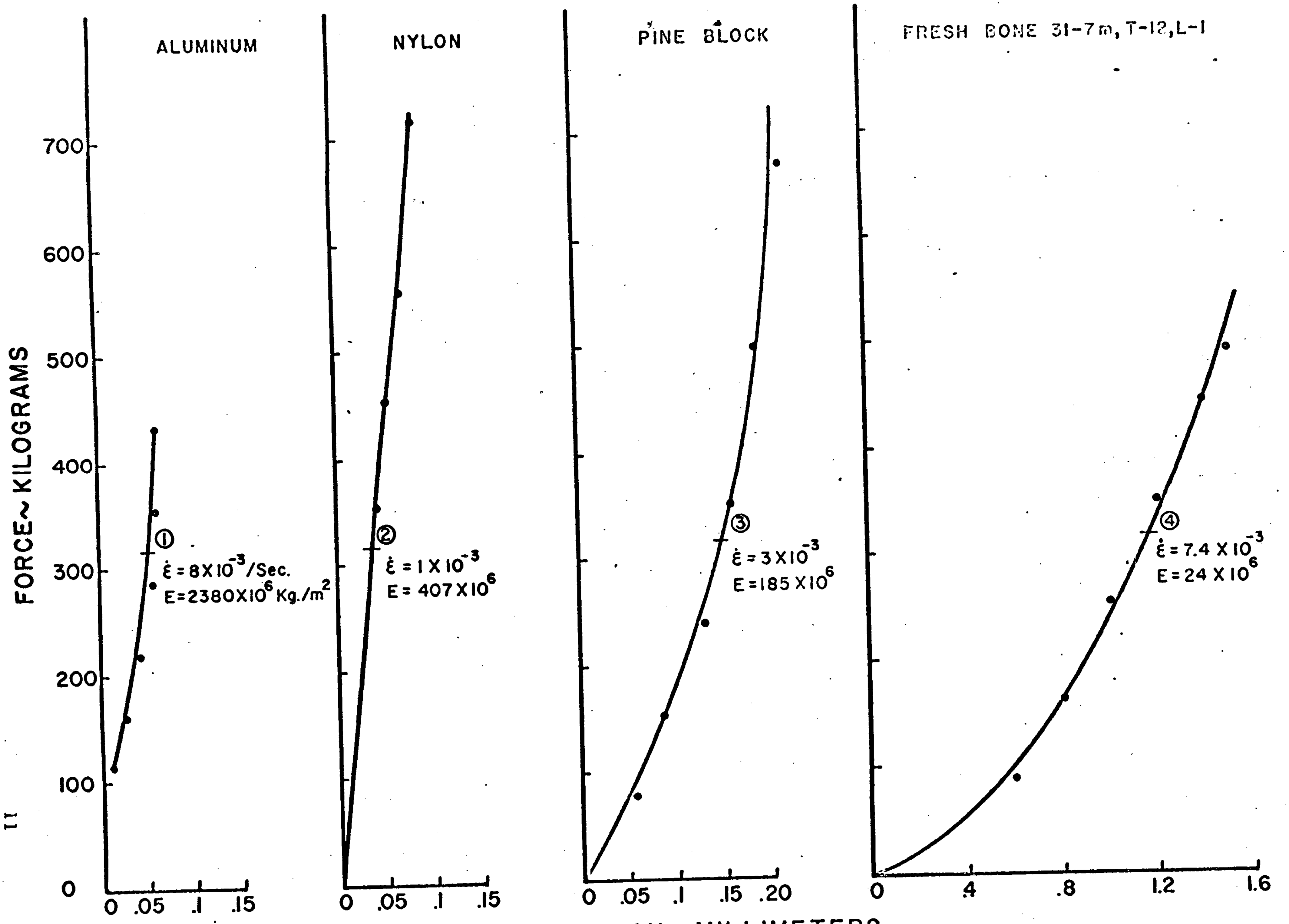
ALUMINUM
NYLON
PINE BLOCK
FRESH BONE 31-7m, T-12, L-1
FORCE~ KILOGRAMS
0
100
200
300
400
500
600
700
① $\dot{\varepsilon} = 8 \times 10^{-3}$/Sec.
$E = 2380 \times 10^{6}$ Kg./m^2
② $\dot{\varepsilon} = 1 \times 10^{-3}$
$E = 407 \times 10^{6}$
③ $\dot{\varepsilon} = 3 \times 10^{-3}$
$E = 185 \times 10^{6}$
④ $\dot{\varepsilon} = 7.4 \times 10^{-3}$
$E = 24 \times 10^{6}$
0 .05 .1 .15
0 .05 .1 .15
0 .05 .1 .15 .20
0 .4 .8 1.2 1.6
DEFLECTION~ MILLIMETERS
A-00159-27

Figure 3

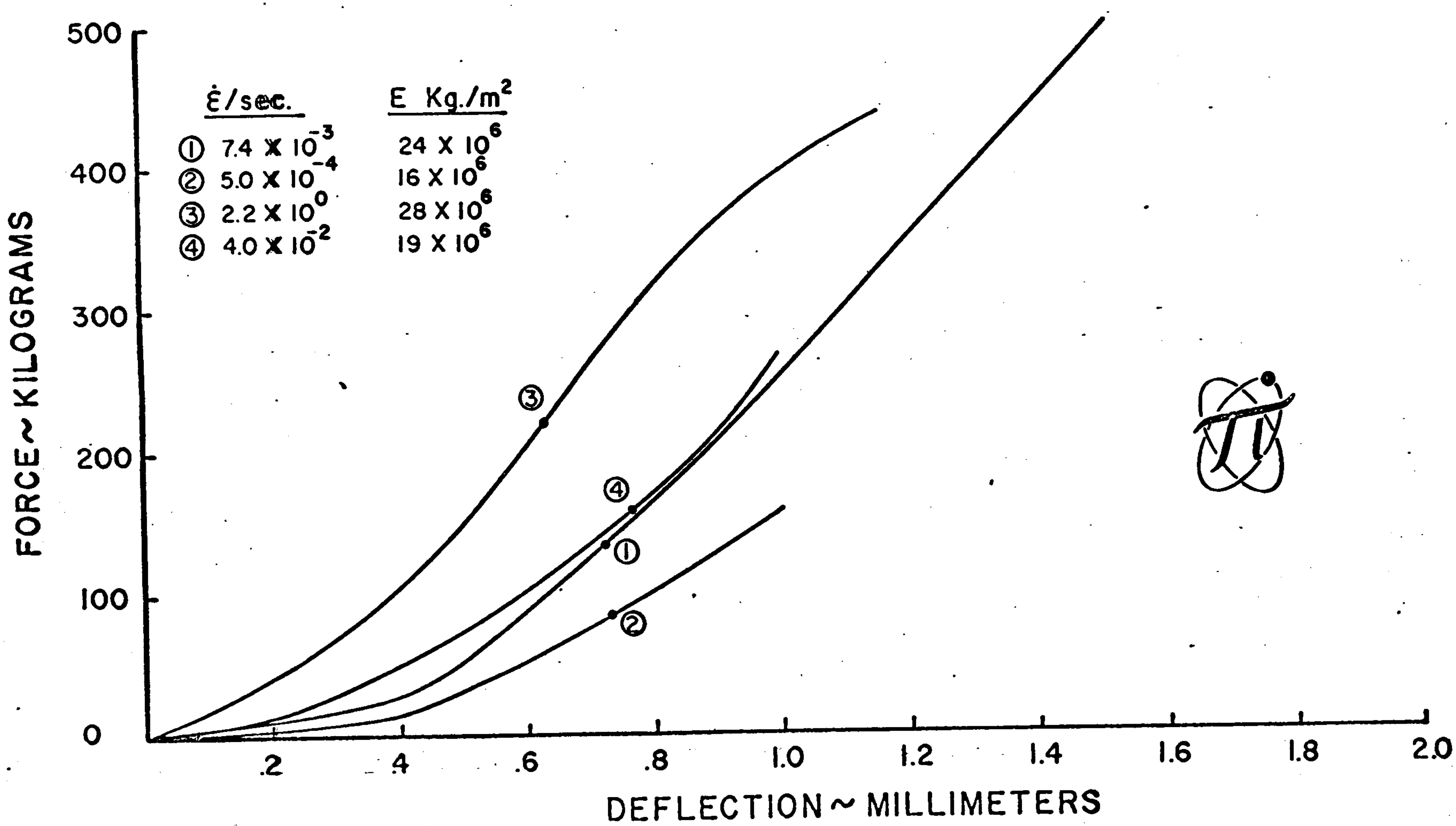

Figure 4

FORCE - DEFLECTION CURVES FRESH-UNFROZEN VERTEBRAE (T12 - L1)		
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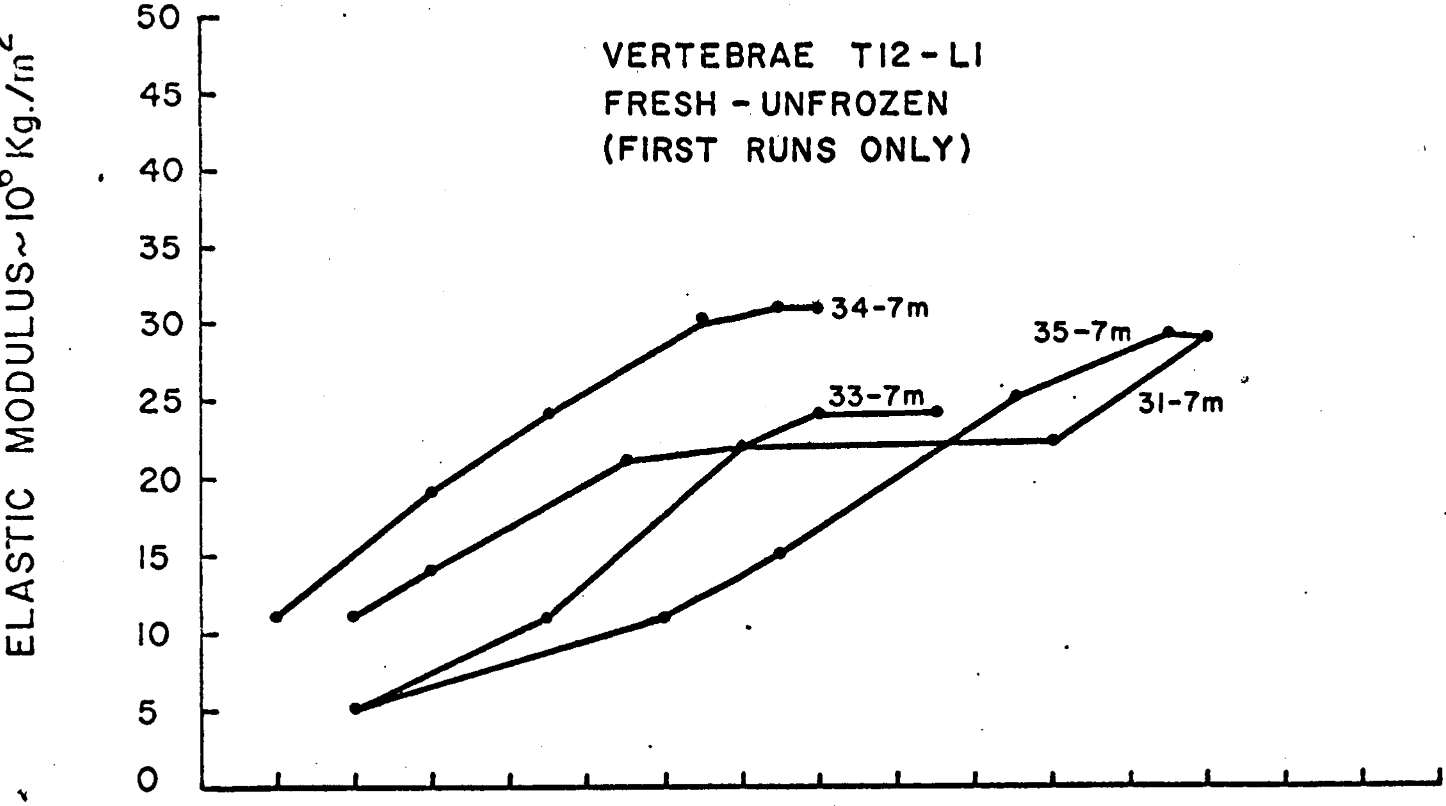

Figure 6

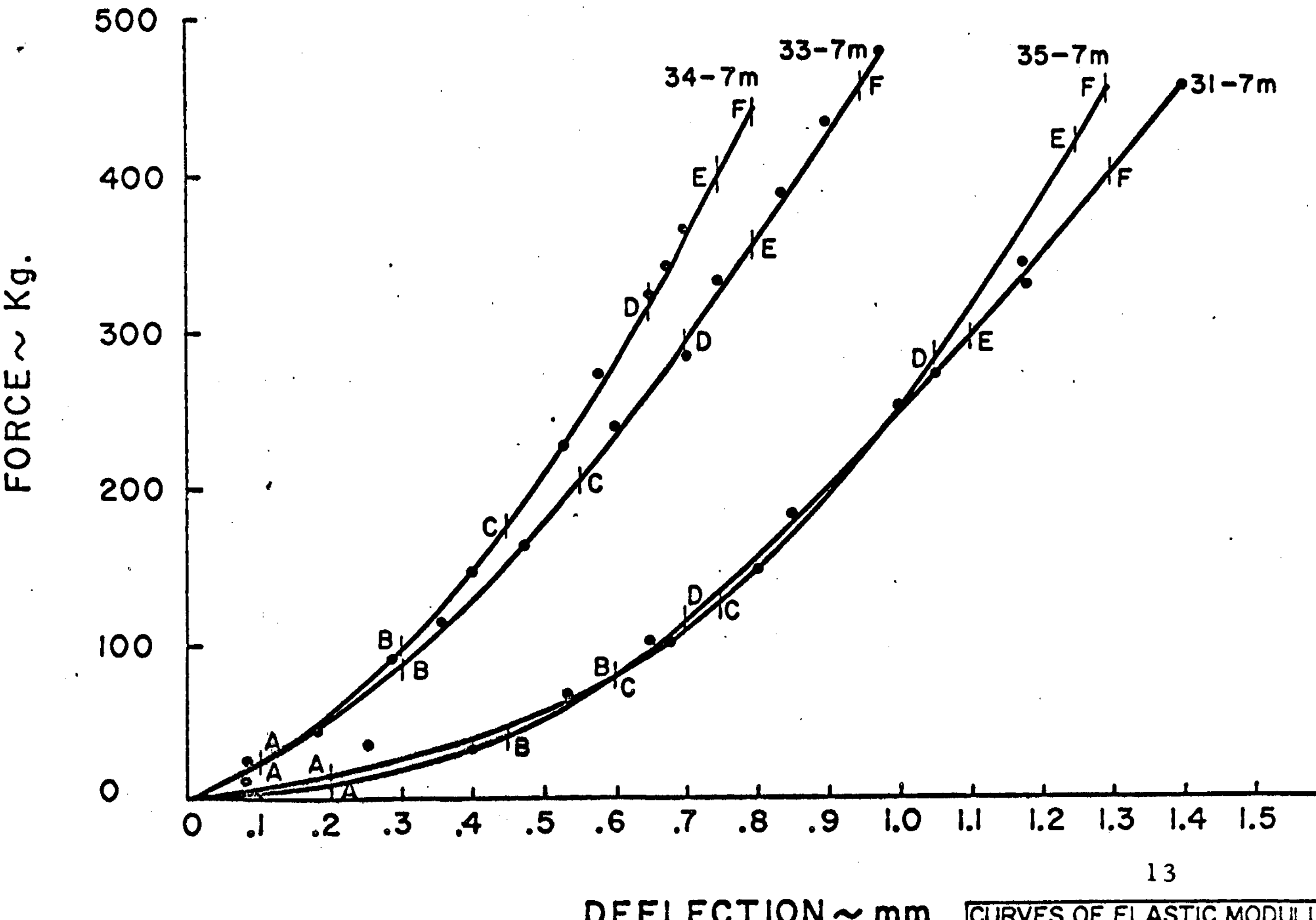

Figure 5

CURVES OF ELASTIC MODULUS vs. DEFLECTION & FORCE vs. DEFLECTION FOR SPECIFIED VERTEBRA.		
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From an examination of Figure 7, the difference in average stiffness of fresh frozen, fresh unfrozen and preserved material is also apparent. The number of specimens available for testing is, however, too small to allow quantitative conclusions.

Determinations of yield point or fracture loads are too few for presentation at this time. On the average, loads exceeding 250 kg cause a change in force-deflection characteristics when the test is repeated at the same machine settings or strain rate. Audible snapping and extrusion of blood from the specimen accompany the changes in mechanical characteristics. Some bones take a permanent set under these conditions. Bone failure has preceded any visible failure of the disc in all cases; this has been confirmed by sectioning of bones after completion of testing.

Typical failure curves, as seen for the usual engineering materials (Figure 8), are exceptional. Instead, tracings similar to those in Figure 9 are demonstrated. Force-deflection representation of fracture, as noted in Figure 9, demonstrates a dramatic decrease in force for increased deflection. Hence, some arbitrary per cent change in stiffness or elastic modulus may have to be used to determine the rate dependence of failure loads.

DISCUSSION

Static strengths of spine segments determined from preparations similar to ours have been reported by Ruff[2] and Brown, et al.[7] The majority of these reports have concentrated on breaking strengths. However, Perey[6] and Brown, et al.[7] have published stiffness figures and force-deflection curves respectively under static loads. Their data is of the same order of magnitude as our own at the lowest strain rates.

McElhaney, et al.[10, 11] have ascertained the effect of rate of deflection on bone properties. Rectangular specimens of cortical bone from femurs 4 x 4 x 6 mm in size were tested in a constant strain rate machine. An increasing stiffness logarithmically related to strain rate was demonstrated.

McElhaney's[11] figures for elastic moduli for bone cortex are two orders of magnitude larger than those reported here. This can, of course, be explained by the very thin cortex surrounding vertebral bodies[12] and the disc structure,[8] which we have found to be only 1/3 as stiff as the centra.

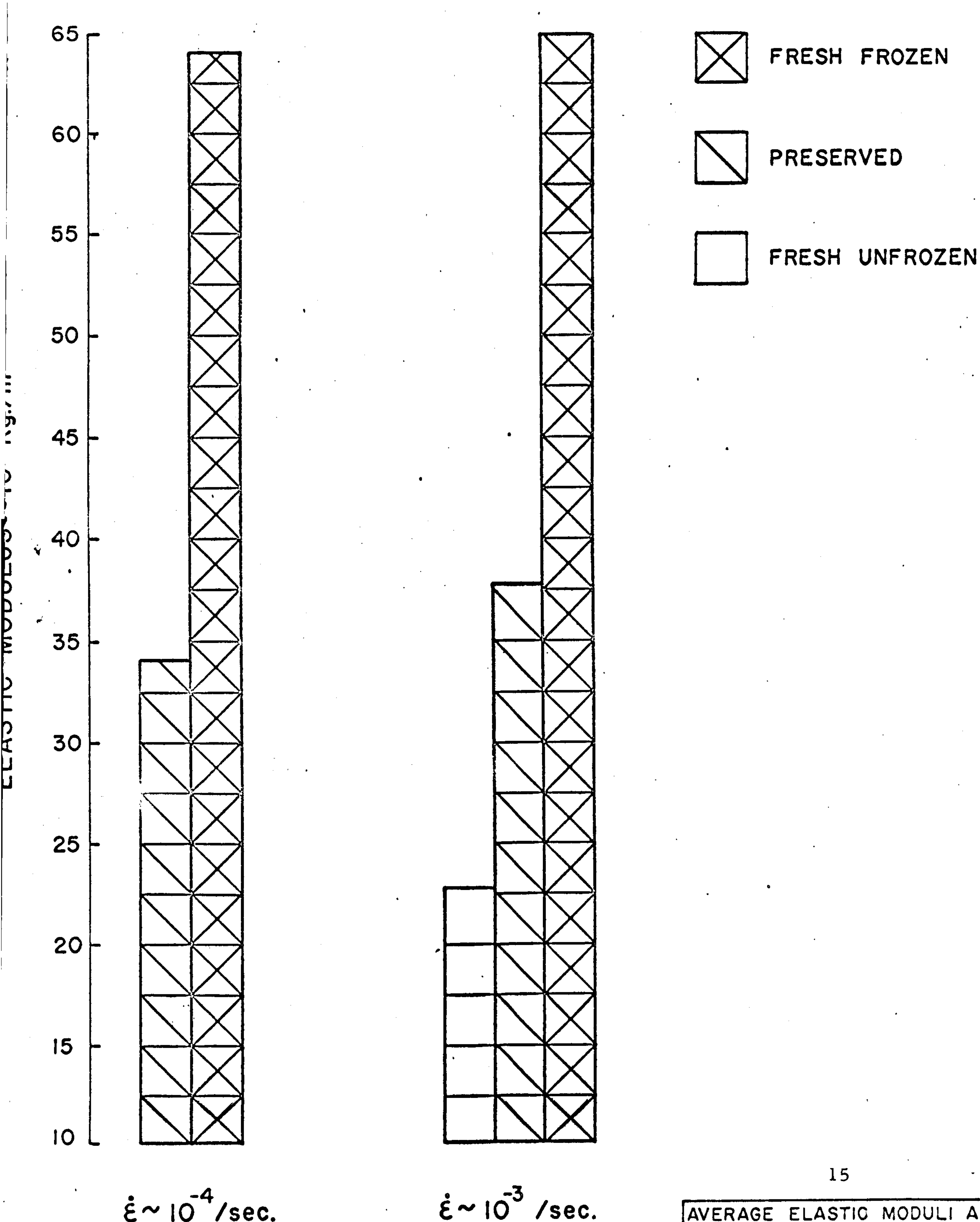

Figure 7

AVERAGE ELASTIC MODULI AS A FUNCTION OF METHOD OF PREPARATION.		
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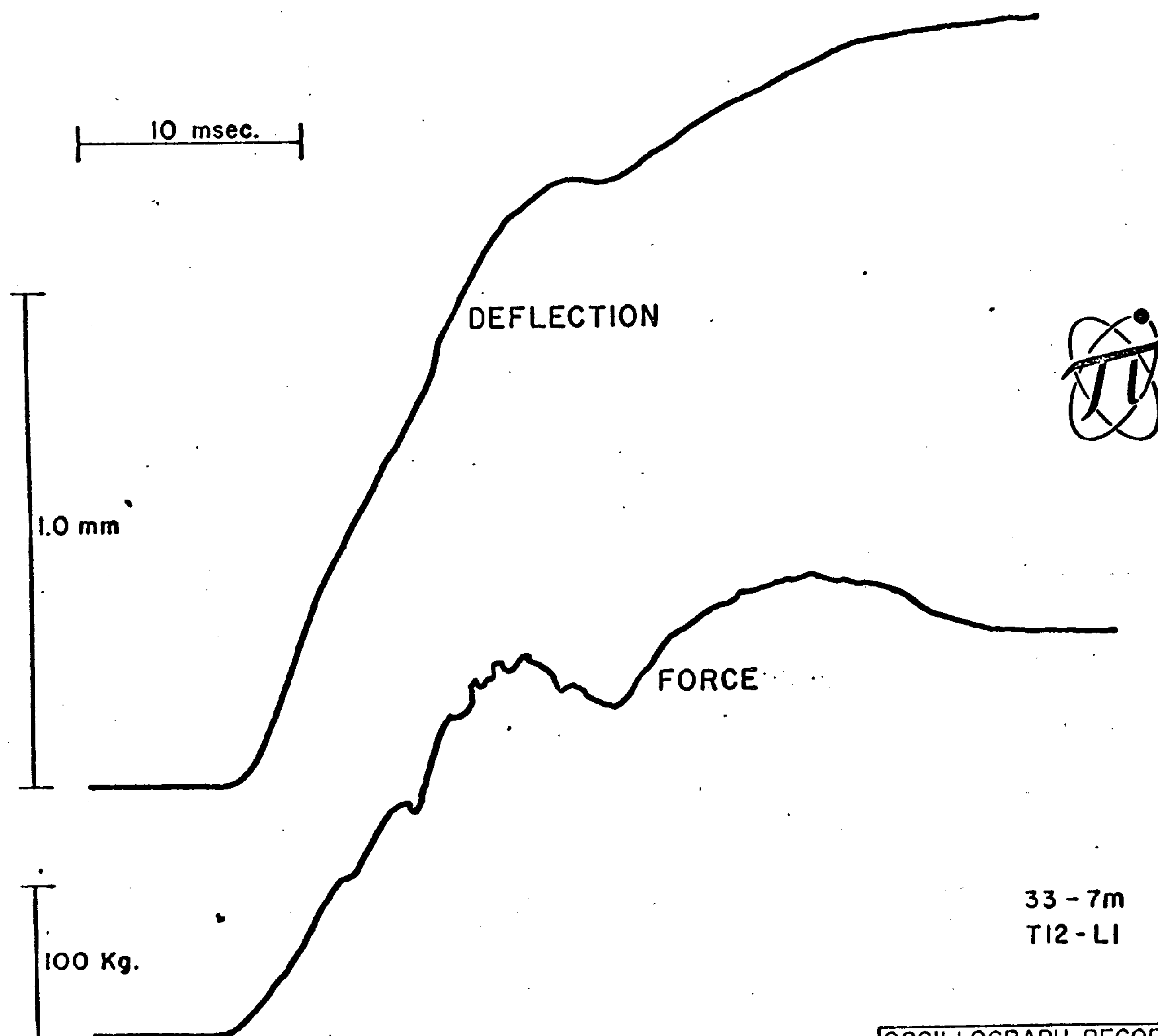

Figure 8

OSCILLOGRAPH RECORDING OF FORCE AND DEFLECTION WITH COMMON TIME BASE.		
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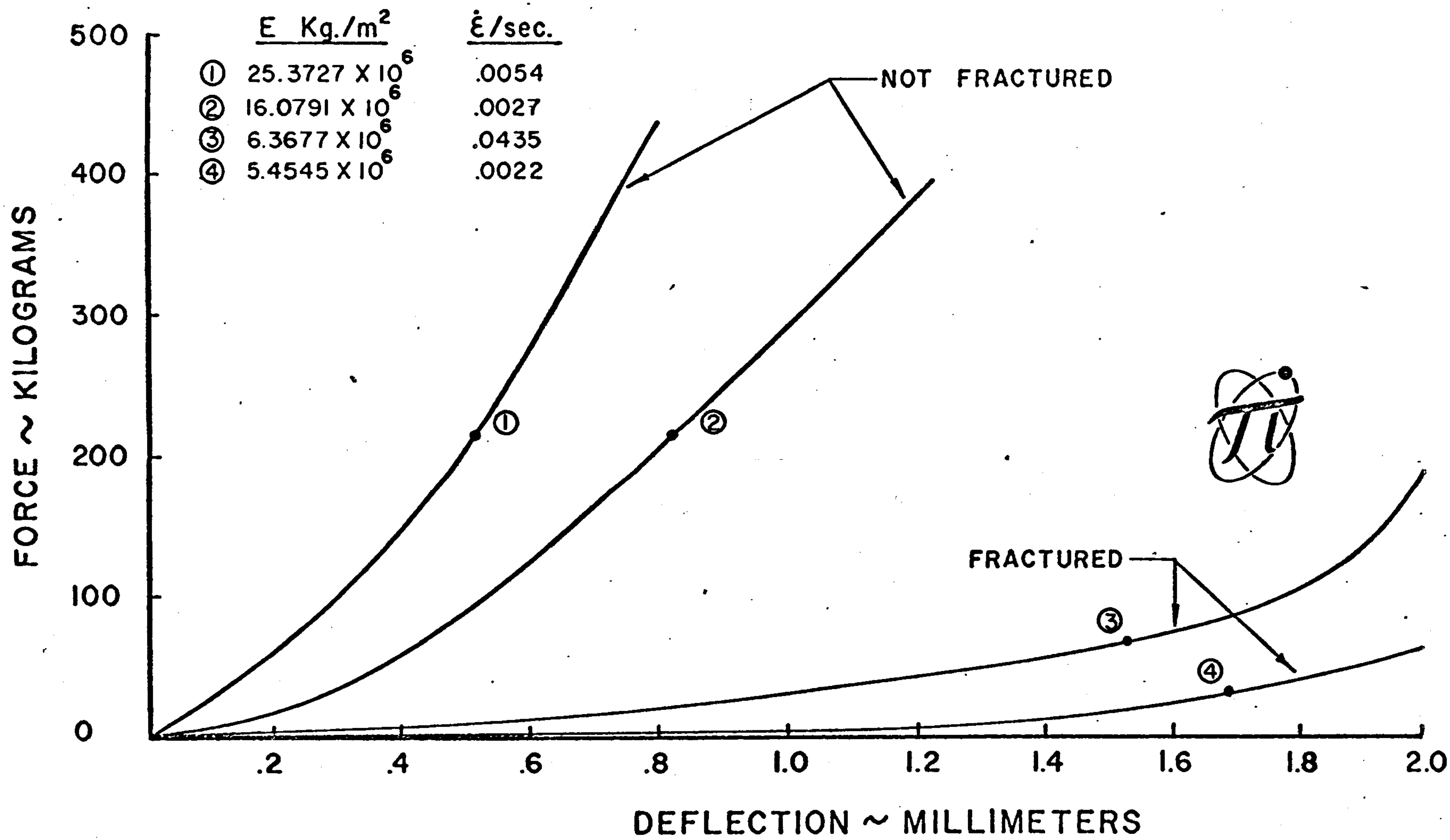

Figure 9

FORCE DEFLECTION CURVES FRESH UNFROZEN VERTEBRAE (T12-L1)		
DATE: 11-7-67	FOR: L.S.H.	DWG. NO: A-00159-31

RECOMMENDATIONS

In order that a spinal model, such as that reported by Toth,[5] be used to the best advantage, continued tests of this sort will be necessary. Normalization of force-deflection data for areas and initial lengths may make earlier predictive results available from the model, since the mechanical characteristics of other segments of the spine could then be obtained from anthropometric data. These would then be substantiated by actual measurements as refinement of the model continues.

Testing of this model using an instrumented dummy and entire cadavers is planned.[13] Since this study involves preserved material, reliable information concerning the differences in response in fresh and formalin-preserved spinal segments is vital.[14, 15] Most importantly, experiments with whole cadavers will allow meaningful assembly of the isolated spinal segments on mathematical and experimental grounds.

From a comparison of actual measurements of deflections and loads in the cadaver and those predicted by the model, successive improvement of the model should result. As a final goal, this model must adequately predict individual segmental loads in situ when subjected to arbitrary input waveshapes.

CONCLUSIONS

A preparation of two human vertebral bones and a single intact intervertebral disc shows a definite rate dependence when subjected to dynamic loading. A considerable variation in elastic moduli during a given test with essentially constant strain rate gives indication of nonlinear response. A statistically reliable determination of these characteristics for fresh and formalin-preserved bones is essential to the construction of a useful model of spinal response to acceleration.

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SECTION II

BONE ASH AND MOISTURE DETERMINATIONS ON ISOLATED VERTEBRAE

INTRODUCTION

Many factors such as age and health of an individual, dietary intake and hormonal control are known to affect the strength of bones. Recent studies by Bell, et al.[1] have related the bone ash content to the mechanical strength. These studies have demonstrated that the breaking strength of a vertebra depends on the amount of osseous tissue in it. Although the relationship is not linear, it is demonstrated that the breaking strength is proportional to the osseous tissue content of the bone.

METHODS AND MATERIALS

Moisture content and bone ash determinations were made on five T12-L1 vertebral segments. Sagittal sections of 3 - 5 mm width were cut from the middle of the body of the segments, thus each determination represents both T12 and L1 bone. Determinations were made on the segments after dynamic testing per Section I of this report. Segments 33-7M, 34-7M and 35-7M were frozen between dynamic testing and moisture and ash determinations. The possible consequences of this procedure are discussed later.

RESULTS

Table 1 of this section summarizes the results of the afore mentioned determinations.

DISCUSSION

Bone contains from 14 to 44 per cent water depending upon the type of bone and its location in the body. In our case, where we are working with sagittal sections of vertebrae, thus more porous bone, the moisture content

TABLE 1: MOISTURE CONTENT AND BONE ASH CONTENT OF SAGITTAL SECTIONS OF ISOLATED VERTEBRAE (T12-L1)

Sample No.	Sample Wt.	Moisture Wt.	% Moisture	Average % Moisture	Ash Wt.	% Ash	Average % Ash
31-7M	6.1900	2.6122	42.2		2.1380	34.5	
	5.9476	2.5127	42.2	42.8	2.0314	34.2	34.3
	6.1583	2.7048	43.9		2.1010	34.1	
32-7M	9.6870	4.4577	46.0	45.6	3.1941	33.0	
	8.1970	3.8849	47.4		2.5158	30.7	32.1
	7.9962	3.4709	43.4		2.6174	32.7	
33-7M	10.2017	5.1954	50.9	52.1	1.9757	19.4	
	10.2131	5.3459	52.3		1.8770	18.4	18.3
	6.5296	3.4635	53.0		1.1150	17.1	
34-7M	9.8180	4.3420	44.2	44.3	1.6030	16.3	
	10.1195	4.4376	43.9		1.6284	16.1	16.2
	7.4234	3.3167	44.7		1.2009	16.2	
35-7M	8.1621	4.4544	54.6	55.5	1.5524	19.0	
	7.4311	4.1628	56.0		1.4273	19.2	18.8
	8.0667	4.5189	56.0		1.4718	18.2	

can be considered to be much higher. Table 1 contains data on the moisture content of sagittal sections of isolated vertebrae (T12 and L1).

The moisture content of the sections are a little high. This could be due to the type of section taken from the sample. Samples 33-7M, 34-7M and 35-7M were cut while frozen which may have allowed for condensation of some moistures on the samples, thus allowing for an indicated elevation in the moisture content. Future samples will be cut from fresh vertebrae immediately after testing on the Detroit Testing Machine and moisture content determined. This will prevent alterations in moisture due to storage in the frozen state and minimize the alterations when preparing the individual samples.

Bone ash content in normal, dry, de-fatted bones range from 50-70 per cent with a 30 - 70 per cent organic matter. Bone ash in our experiments was determined on a wet basis and the bone tissue had not been de-fatted; therefore, our values are a little low.

The bone ash results appearing in Table 1 may be converted to dry weight basis; however, this would not appreciably affect the range. In the future, as in the moisture determinations, the ash content will be from vertebrae which have not been frozen and stored. This should eliminate any effect of the freezing or condensation of moisture on the sample.

At this time the sample size is too small to correlate bone ash content with the strength of the vertebrae tested.

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[1] Bell, G. H., Dunbar, O., Beck, J. S. and Gibb, A.: Variations in strength of vertebrae with age and their relationship to osteoporosis. Calc. Tiss. Res., 1:75, 1967.

SECTION III

The following is an annotated bibliography which adds to that reported in the Final Report - Phase I, Studies on Vertebral Injuries Sustained During Ejection - by Technology Incorporated for Contract NONR-4675(00), sponsored by the Physiology Branch, Biological Sciences Division, Office of Naval Research.

1. Bell, G.H., Dunbar, O., Beck, J.S. and Gibb, A.: Variations in strength of vertebrae with age and their relation to osteoporosis. Calc. Tiss. Res. 1:75-86, 1967.

   Fourth and fifth lumbar vertebrae and bone specimens from the iliac crests were obtained at post mortem examinations of human subjects, ages ranging from 26 to 86 years. The iliac crest samples were used in the assessment of trabecular density. The lumbar vertebrae, after removal of spinous process and pedicles, were compressed in a testing machine to mechanical failure. Breaking stress (load per unit area at failure), strain (percentage deformation) at failure, and relative ash content (ash per unit volume) were the results obtained. It was found that with increasing age, both the breaking stress and the relative ash content declined.

2. Currey, J. D.: Differences in the tensile strength of bone of different histological types. J. Anat., 93:87-95, 1959.

   A study using oxen femur to relate the strength and weakness of bone to the absence or presence of Haversian systems. The bone samples were chilled and machined and polished into dumbell-shaped pieces. After machining, samples were placed in saline for 24 hours. Machined and saline soaked samples were then tested for ultimate tensile strength. Conclusions of such testing were that immature Haversian systems have large central cavities which reduce the actual amount of bone substance present per unit volume, and newly formed Haversian systems are not as fully mineralized as, and therefore presumably weaker than, the surrounding primary bone.

3. Dempster, W. T. and Coleman, R. F.: Tensile strength of bone along and across the grain. J. Appl. Physiol., 16:355, 1961.

   Tibia specimens from museum cadavers were subjected to direct tension or bending until failure. Specimens were .75 in. x .75 in. x 108 in. thick. Both wet and dry specimens were tested. It was concluded that

the weaker structural elements are the cement lines surrounding the osteones and the planes between the lamellar of the Haversian systems. Differences in fibrous tearing at certain fracture edges suggest that bone is weaker parallel to, rather than transverse with, the predominant orientation of the collagenous fibers. Ultimate psi data is given for all testing conditions.

4. Evans, F. G.: Methods of studying the biomechanical significance of bone form. Am. J. of Phys. Anthropol., Vol. II, 1953.

Many engineering terms, such as force, energy, tensile, compressive, etc. are defined in the proper context. Many technics are reviewed which demonstrate how the biomechanics of bone may be determined and the factors influencing the determinations.

5. Evans, F. G. and Lebow, M.: The strength of human compact bone as revealed by engineering technics. Am. J. Surg., 1952.

Bone samples were removed from the long bones of the lower extremities. The tested bone size was 0.150 x .090 inches. A 5,000 lb. capacity materials testing machine was used to appy the force. Drying the specimens increased the average tensile strength but reduced the per cent elongation under tension. Wet samples had a higher per cent elongation under tension. Dry samples produced a straight-line stress-strain diagram, whereas wet samples produced a curved stress-strain plot. The ultimate tensile strength for femurs was 12,000 lb/in.^{2} with slightly higher values for tibia and fibula.

6. Furst, W.: The force required to crush vertebrae: Its probable mechanical relation to the postmetrazol fracture. Presented before the Philadelphia Neurological Society, College of Physicians and Surgeons, Philadelphia, Pennsylvania, February 23, 1940.

The fifth thoracic vertebra was removed and placed in formalin. Prior to testing, the intervertebral cartilage was removed and the test specimen dried. Axial compression testing was done using an electrically controlled press capable of measuring pressure variations within five pounds. With a pressure of 250 - 275 pounds cracking sounds occurred, and at 750 - 800 pounds the vertebra cracked apart transversely. Emphasis is placed on the necessity for maintaining spinal hyperextension in preventing post metrazol vertebral fractures and intervertebral disc complications.

7. Hardy, G. W., Lissner, H. R., Webster, J. E. and Gurdjian, E. S.: Repeated loading tests of the lumbar spine. Surgical Forum, 9:690-695, 1958.

Fresh and embalmed cadaver lumbar vertebrae, usually 5, were submitted to axial compression and transverse bending tests. Loads were applied by a modified commercial fatigue testing machine at a rate of 120 cpm. The posterior elements were removed on all but two of the test specimens. After preparation, the specimens were refrigerated and tested 5 days - 5 weeks later. Freezing seemed to have no effect on the results. The loads were applied at given total cycles until fatigue. Two segments in which the posterior elements were left intact appeared to be stronger. Data of loads applied and total cycles is presented.

8. Henzel, J. H., Brinkley, J. W. and Mohr, G. C.: Acceleration profile associated with thoracic vertebral compression. J. Trauma, 6(6):756-766, 1966.

Report of a case of vertebral compressions, T4 - T5, occurring as the result of an impact exposure during which the acceleration input to the subject was accurately measured. Description of the seat and restraint system used and impact profile associated with T4 - T5 vertebral body compression. The significance of the information presented in this report is that acceleration vectors and force magnitudes were recorded in an instance in which vertebral structural tolerance was exceeded.

9. Lease, G. O'D. and Evans, F. G.: Strength of human metatarsal bones under repetative loading. J. Appl. Physiol., 14(1):49-51, 1959.

The fatigue life of 51 intact human metatarsals from eight adult males and three females were tested in a Sonntag Flexure Fatigue machine. Bones were tested wet and dry. The repetitions to failure with a 15 lb. load, the maximum applied varied from 1,000 to 10,297,000 for the dry specimens, and from 150,000 to 13,908,000 for the wet ones. The greatest variation between the minimum and maximum number of repetitions to failure was 10,287,000 and 10,966,000 repetitions for the dry and wet specimens, respectively. Metatarsals II and III had the longest fatigue life when wet, while metatarsals IV and V had the greatest fatigue life when dry. No consistent relations were found between the fatigue life of the bones and their size or the age of the individual from whom they were obtained.

10. McElhaney, J.H.: Dynamic response of bone and muscle tissue. J. Appl. Physiol., 21(4):1231-1236, 1966.

The mechanical response of bone and muscle tissue to impacts of varying velocity was studied. Fresh bovine femur, embalmed human femur, and bovine muscle tissue surrounding the femur were subjects of the testing. Results of this experiment are presented in the form of stress-strain curves at various strain rates. Average values of properties obtained from these curves are given in terms of strain rate (/sec), ultimate compressive strength (psi), energy absorption capacity (lb/in^3), elastic modulus (psi), maximum strain to failure (%), and Poisson's Ratio (elastic/dried). A critical velocity was noted for bone in the neighborhood corresponding to a strain rate of 1/sec. A stress, strain, strain-rate surface representation of the data is suggested and the similarities between the dynamic response of bone, nylon, and aluminum noted.

11. McElhaney, J., Fogle, J., Byars, E. and Weaver, G.: Effect of embalming on the mechanical properties of beef bone. J. Appl. Physiol., 19(6):1234-1236, 1964.

Beef femur specimens were wet sanded into test specimens. Both fresh and embalmed bones were tested by compressive forces. Fresh bones were found to have a mean ultimate compressive strength of 19,360 psi and a mean elastic modulus of 4.18×10^6 psi. Embalmed bones revealed a mean ultimate compressive strength of 16,990 psi and a mean elastic modulus of 3.92×10^6 psi.

12. Nickerson, J.L. and Drazic, M.: Young's modulus and breaking strength of body tissues. Technical Documentary Report No. AMRL-TDR-64-23, Biophysics Laboratory, Wright-Patterson AFB, Ohio, 1964.

This report describes the results of the measurement of Young's Modulus, the breaking strength and breaking index

(breaking index = $\frac{\text{breaking strength}}{\text{Young's Modulus}}$) in the stretching of 23 different tissues or segments of tissues from dogs and humans. Tissues used in testing were esophagus, stomach, artery, and intestine. Young's Modulus values ranged from 5.7×10^6 dynes/cm^2 for the aortic arch to 110×10^6 dynes/cm^2 for the skin of the abdomen. Breaking strength for the same tissues ranged from 3.3 to 44.4 Kg/cm^2 respectively.

13. Ross, D: A method for the production of increased compression strength of bone: An experimental study. Brit. J. Surg., 20:337-342, 1932-1933.

Tibia from dogs were tested before and after resection. The bones were potted in plaster-of-paris and axial compressive forces were applied by a 10, 000 lb. capacity Olsen testing machine. Bones that had been resectioned and new bone allowed to grow showed a higher compressive strength than normal bone.

14. Saville, P. D. and Smith, R.: Bone density, breaking force and leg muscle mass as functions of weight in bipedal rats. Am. J. Physical Anthropol., 25:35-39, 1966.

Rats made to develop without fore-limbs were used to supply rear leg femurs for test. Rear leg muscle mass was found to be larger in test rats than in normal rats. Increased muscle mass was associated with increasing femur density. Increased density increased the force required to break the bone. They concluded that weight bearing influences bone density and breaking force through muscle mass.

15. Schraer, H., Sear, W. J. and Schraer, R.: Changes in bone mass and density in living rats during the manipulation of calcium intake. Arch. Biochem. and Biophysics, 100:393-398, 1963.

Rats maintained on high calcium diets tended to have an increased bone density over control subjects receiving regular diets.

16. Sedlin, E. D.: A rheologic model for cortical bone. Dept. of Orthopaedic Surg., Univ. of Göteborg, Göteborg, Sweden, 1965.

Physical properties of bone were studied under controlled laboratory conditions in which the effects of variables in the methods of preparation, methods of storage, methods of testing, ages of subjects, sources of samples and their microanatomy were examined. Subject of the investigation was 663 samples of human femoral cortex obtained from 43 subjects at autopsy. From this information it was possible to obtain a model for the deformation of cortical bone using rheologic symbology and methodology.

17. Smith, R. W. and Keiper, D. A.: Dynamic measurement of viscoelastic properties of bone. Am. J. Med. Elect., 4(4): 156-160, 1965.

Specimens were taken from human rib, ilium, tibia and clavicle and placed in 50% alcohol. Specimens were then wetted with Ringer's solution and ground to specific size. Elastic modulus, viscous modulus, stiffness rate and bone density were measured. Values for elastic modulus of each specimen was given in dynes/cm^2.

18. Steindler, A.: Physical properties of bone. Arch. Physical Therapy, 17:336-345, 1936.

A review of the literature on the physical properties of bone prior to 1936 was made.

19. Trotter, M. and Peterson, R.R.: Ashweight in precent of their dry fat-free weight. The Anat. Rec., 123:341-358, 1955.

Bone specimens were taken from embalmed cadavers. All specimens were degreased with acetone prior to ashing. In six complete skeletons the total percentage ash weights varied from 64.8% - 66.9%.

20. Vose, G.P. and Kubula, A.L.: Bone strength - Its relationship to x-ray determined ash content. Human Biol., 31:261-270, 1959.

Embalmed human femurs were dried at room temperature for 30 days. Four were soaked in Ringer's solution prior to testing. X-ray determined ash content vs. breaking stress showed the factors to be related in an exponential manner. No correlation between strain/unit length and breaking stress by bending was found.

21. Walmsley, R. and Smith, J.W.: Variations in bone structure and the value of Young's Modulus. J. Anat., 91:603, 1957.

Study of the variation of Young's Modulus values using the radius of the horse and the metacarpus of the ox. The central four-fifths of the cortex of these shafts of these bones were cut longitudinally into a number of rods which were then trimmed to a uniform rectangular cross-section. This investigation attempted to correlate the variations in Young's Modulus with certain histological features of the bones.

22. Weaver, J.K.: The microscopic hardness of bone. J. Bone and Joint Surg., 48A(1):273-288, 1966.

Fresh autopsy bone specimens were acquired and frozen. Microscopic sections were ground for testing from cortical bone. Microscopic hardness was found to be a reliable measure of the degree of mineralization. Little variation in hardness of the same bone taken from standard sites in different individuals.

23. Weiss, L.: Static loading of the mandible. Oral Surg., Oral Med. and Oral Pathol., 19:253-262, 1965.

Two hundred and twenty (220) embalmed human mandibles had loads applied to the symphysis at the rate of 100 lbs/minute. Stress coat lacquer was applied to the bone. The most frequently fractured areas were in the subcondylar region, the region of the angle of the mandible and the region of the mental foramen. These regions were found to be under the greatest tensile strain.