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Cervical Spine Compression Responses

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ABSTRACT

Time-varying compressive loading was applied to unembalmed human cervical spines using an MTS closed-loop hydraulic testing machine. Load programs included relaxation, cyclic loading, variable rate constant velocity loading (0.13-64 cm/sec), and constant velocity loading to failure. The failures produced were similar to those observed clinically. A generalized quasi-linear viscoelastic Maxwell-Weichert model incorporating a continuous relaxation spectrum was developed to predict the relaxation and constant velocity test responses. The fit was adequate considering the complexity of the structure involved.

WHILE INJURY STATISTICS generally attribute only 2 to 4% of serious trauma to the neck, any neck injury can have debilitating if not life-threatening consequences. The human neck not only contains vital neurologic, vascular and respiratory structures but it also provides for the support and motion of the head. From an anatomical, neurological and mechanical point of view, the neck is quite complex. Much research has been done to describe neck anatomy [17,24,37],* injury mechanisms [1,2,4,7,12,15,21,24-28,31,33,35,41], and the diagnosis [3,5,11,18,20], classification [3, 6,16,19,24,39] and treatment of neck injuries, but only a few serious efforts [5,7,32] have been made to quantitate the structural properties. Many authors have studied the viscoelastic behavior of the lumbar and thoracic discs [8-10,14,22,36,38,41] but rarely have they extended their work to include the cervical discs.

*References at end of paper.

This paper summarizes both experimental and theoretical studies aimed at quantifying and predicting the responses of the cervical spine to compression loading. Time-dependent responses (i.e. force~time and deformation~time characteristics) of the cervical spine were measured and analyzed. Peak loads and deflections, strain energies and mechanisms have been summarized for all experimentally produced failures. A model that allows prediction of these responses has been proposed. Hopefully, the results presented in this paper will be useful in further mathematical modeling, in the design of protective devices, and in the development of anthropomorphic models.

RELATED LITERATURE

Probably the earliest empirical study was Messerer's (1880) work on the mechanical properties of the vertebrae. He reported compressive breaking loads ranging from 1.47-2.16 kN (330-486 lb) for the lower cervical spine. Later, Yamada (1970), in his extensive compilation on the strength of biological tissues, provided data on the static load~deflection properties of the vertebral bodies and discs. Roaf (1960) loaded single cervical spinal units in compression, extension, flexion, lateral flexion, horizontal shear and rotation. He found that the intact disc, which failed at approximately 7.12 kN (1600 lb), was more resistant to compression than wet vertebrae, which failed at approximately 6.23 kN (1400 lb). Bauze and Ardran (1978) loaded human cadaveric cervical spines in compression and reported flexion dislocations with loads of 135-145 kg [1.32-1.42 kN (298-320 lb)]. Fielding et al. (1974) conducted shearing studies of the atlas. In all cases, the 70-180 kp [0.69-1.77 kN (154-397 lb)] force required to fracture the odontoid process was greater than the 12-180 kp [0.12-1.77 kN (26-397 lb)] force required to tear the transverse ligament. Althoff et al. (1980) described experimental fractures of the odontoid process but did not report load or deformation data. Selecki and Williams (1970) conducted an extensive study of cadaveric cervical spines loaded with a manually-operated hydraulic jack. Unfortunately, they monitored the pressure in the hydraulic line and reported their results in terms of hydraulic pressure without indicating the ram piston diameter. They were able, however, to duplicate several types of clinically observed injuries. Panjabi et al. (1975) measured rotation and translation of the upper vertebra as a function of transection of the components in single units of the cervical spine. Liu and Krieger (1978) reported load~deflection responses from axial compression tests on single cervical spinal units. Sances et al. (1982) tested isolated cadaver cervical spines in compression, tension and shear. A quasi-static compression failure was observed at a load of 645 N (145 lb) and dynamic flexion/compression failures were reported at loads ranging from 1.78-4.45 KN (400-1000 lb). Except for studies by Fielding et al., Liu and Krieger and Sances et al., all of these tests were quasi-static and most researchers recorded only the maximum load.

Studies have also been done on impacts to intact cadavers which involved the neck. Most notable of these were the works of Hodgson et al. (1980), who measured strains on the anterior surfaces of the bodies of C2-C7 and near the left facet joints of all of the cervical vertebrae of embalmed cadavers during crown impacts, and Nusholtz et al. (1981), who studied neck motions and failure mechanisms on unembalmed cadavers due to crown impacts. They both reported significant influence of spinal configuration on the spinal response and damage.

INSTRUMENTATION

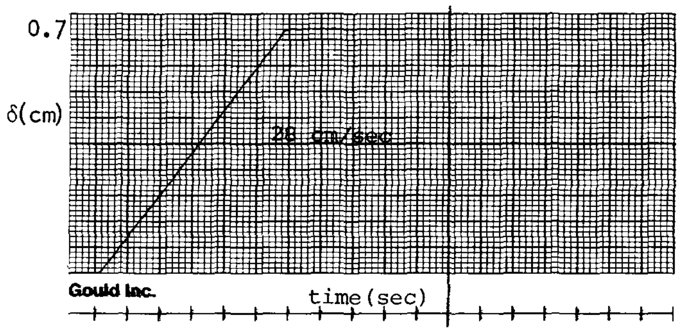
Tests were conducted with a Minneapolis Testing System (MTS) servo-controlled hydraulic testing machine which consisted of a rigid load frame, a 6000 lb ram, a temperature- and humidity-controlled environmental chamber, a 25 gpm 3000 psi hydraulic pump, two nitrogen-filled hydraulic accumulators and a displacement-controlled feedback system. Ram motion was monitored by an integral linear variable differential transformer. Load was measured by a strain gage load cell calibrated with proving rings certified by the National Bureau of Standards. Command voltages were provided by an Exact function generator. With the MTS system, we were able to apply displacements at ram speeds up to 127 cm/sec without overshoot.

Figure 1 illustrates the input displacement~time histories used in these studies. For the relaxation tests, the specimens were preloaded to 200 N and then subjected to ramp-and-hold command signals. For the cyclic modulus tests, sinusoidal waveforms were used.

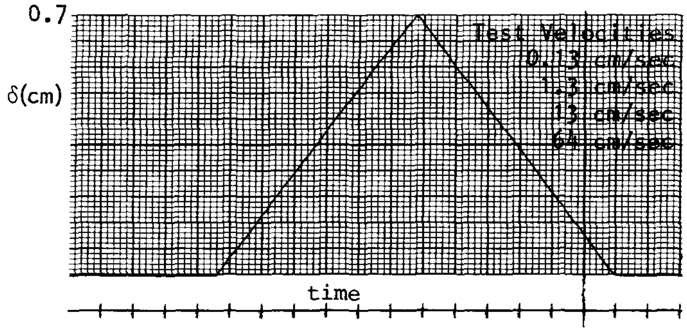
Upon completion of the cyclic modulus tests, a reference state was defined which corresponded to a 200 N preload. This preload defined a constant reference specimen length that became the initial condition for all subsequent constant velocity tests. Triangular waveforms were used for the constant velocity tests.

Specimen load~ and deformation~time histories were stored on a Tektronix 5223 Digitizing Oscilloscope and then recorded onto magnetic tape with a Tektronix 4052A Graphic Computer using the WP1310 Waveform Processing System Software.

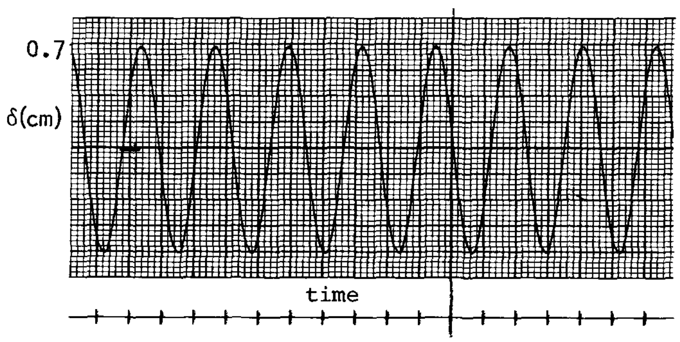
High resolution x-ray images were obtained using a Hewlett-Packard Faxitron unit before and after testing.



Relaxation Test -- Prescribed Deformation



Constant Velocity Test



Cyclic Stiffness Test 20Hz

Figure 1, Test~Deformation Time Histories

PRELIMINARY RHESUS MONKEY TESTS

Preliminary tests were performed on eight rhesus (Macaca Mulatta) monkey cervical spines in order to develop the experimental protocol. The following questions were addressed:

(1) Do changes in the mechanical properties occur as time elapses after death?

(2) Can the properly stored cervical spine be reequilibrated after non-destructive testing and tested further? That is, can the cervical spine recover from tolerable load levels or are the effects of testing irreversible? If the results of the first test are, in fact, repeatable, what period of time is necessary for full recovery prior to subsequent testing?

(3) Does freezing significantly degrade the mechanical responses?

(4) Does the cervical spine exhibit a weak enough temperature dependence between room (25°C) and body (37°C) temperatures to justify room temperature testing?

All eight rhesus monkey necks were removed during nechropsy, sprayed with calcium-buffered saline, and kept sealed in waterproof plastic bags. Relaxation tests were performed on four of the specimens within 2, 24, 48 and 60 hours post-mortem. After testing, these four specimens were stored in the refrigerator, allowed to recover for varying times and then retested. Figure 2 shows typical results. The four remaining specimens were frozen post-mortem, stored at -20°C for two months, and thawed in the refrigerator for four days prior to relaxation testing.

These preliminary rhesus monkey tests demonstrated that:

(1) Properly stored specimens tested within 60 hours post-mortem did not exhibit detectable changes in their relaxation properties. A specimen was considered to be properly stored if it was moistened with calcium-buffered saline and then kept sealed in a waterproof plastic bag throughout the experiment. This procedure was selected in order to reduce the possibility of dehydration or chemical changes which could shift the osmotic gradient. Total immersion of the specimen in saline was deemed undesirable because of the possibility of disc swelling due to fluid imbibition. Anything that could change the fluid balance of the disc and/or ligaments could change the stiffness and, perhaps, even the failure characteristics since disc and ligament stiffness influence the strain distributions.

(2) The preliminary test results have shown that even loads of tolerable levels alter the mechanical behavior of the neck. This observation may be related to the capacity of the intervertebral discs to imbibe and release fluid. A detailed discussion of the osmotic action of the disc is beyond the scope of this paper. When externally loaded, the disc exhibits a tendency to lose fluid and, when the external loads are removed, the disc exhibits a tendency to absorb fluid. It is hypothesized that the reequilibration process involves the osmotic uptake of fluid into the discs and that the reequilibrated state is the end state characterized by an osmotic balance. Figure 2 shows that a 24-hour recovery period was required between the initial test and a subsequent test in order to achieve full reequilibration and test reproducibility. During this period, the specimen was properly stored and refrigerated.

(3) For periods up to two months, freezing had no observable effect on the relaxation responses. Panjabi et al. (1975), Hirsh and Galante (1967), and Casper (1980) also reported no degradation in mechanical properties due to storage by deep freezing and thawing prior to experimentation.

(4) Tests performed at room temperature were comparable to body temperature tests. This conclusion agrees with Casper's (1980) observations for the intervertebral disc.

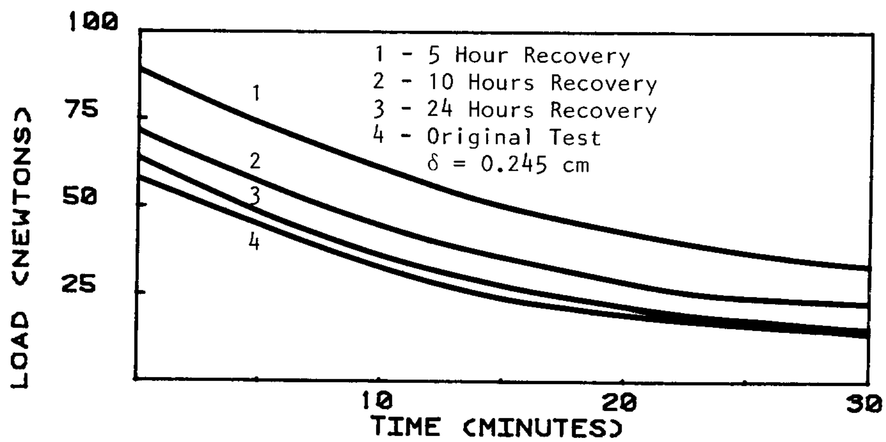


FIGURE 2, RHESUS MONKEY CERVICAL SPINE RELAXATION TESTS

EXPERIMENTAL PROTOCOL FOR HUMAN NECK TESTS

Fourteen intact unembalmed cervical spines were obtained at autopsy from cadavers. The donors, who ranged in age from 42 to 73 years, showed no evidence of cervical spine problems in their hospital records. All specimens included the base of the skull, approximately two centimeters around the foramen, at the proximal end and C5, C6, C7 or T1 at the distal end. All ligamentous structures were kept intact except the ligamentum nuchae where it attached to the base of the skull.

The specimens were sprayed with calcium-buffered saline, sealed in plastic bags until dissection and either tested on the day of removal or frozen and stored at -20°C. At the time of testing, the specimens were thawed to room temperature and allowed to fully equilibrate with their respective fluid environments. Using polyester casting resin, the ends of the specimens were cast in aluminum caps so that the caps were approximately perpendicular to the axes of the end vertebrae. During casting, the aluminum caps were cooled in a flowing water bath to minimize degradation due to the heat of polymerization. Figure 3 illustrates the lordotic curve configuration of the specimens after casting. Next, the initial values of α and L_O were determined. α was measured with an adjustable protractor. L_O was measured with vernier calipers. A moment M_O, measured with a spring scale operating on a known moment arm, was then applied to make the end caps parallel and the cervical vertebrae approximately vertically aligned. This moment varied from 5 to 30 N-m. The specimen was placed in the test fixture.

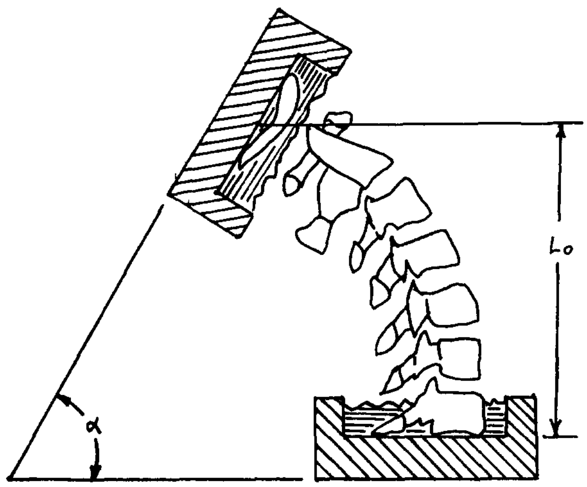


FIGURE 3. SPECIMEN WITH END CAPS

The test fixture (Figure 4), which consisted of a dovetail slide driven by a precision micrometer lead screw, allowed movement of the distal end of the specimen with respect to the proximal end. A displacement h in the anterior or posterior direction was applied with the slide and the lead screw in order to obtain the desired degree of flexion or extension. All specimens were x-rayed before testing in order to document the initial configuration. Finally, the test fixture was installed in the testing machine.

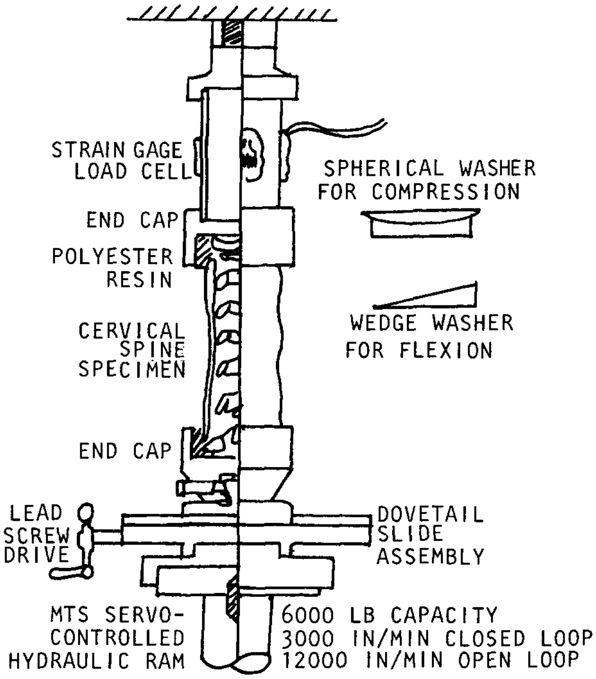


Figure 4, Test Fixture

The following tests were performed at room temperature:

(1) fully equilibrated relaxation test, δ=0.7 cm
(2) cyclic modulus test, δ=0.7 cm, 20 Hz, 150 cycles
(3) mechanically stabilized relaxation test, δ=0.7 cm
(4) constant velocity tests, δ=0.7 cm, ram speed = 0.13, 1.3, 13, and 64 cm/sec
(5) constant velocity load-to-failure test, ram speed ≈ 64 cm/sec.

With the computerized data collection system, tests #2 - #5 were completed in less than one hour.

After testing, the specimens were x-rayed in order to document the final configuration. Next, they were dissected. Failed ligaments and bones were noted and photographed. Critical dimensions were measured and recorded in Table 1.

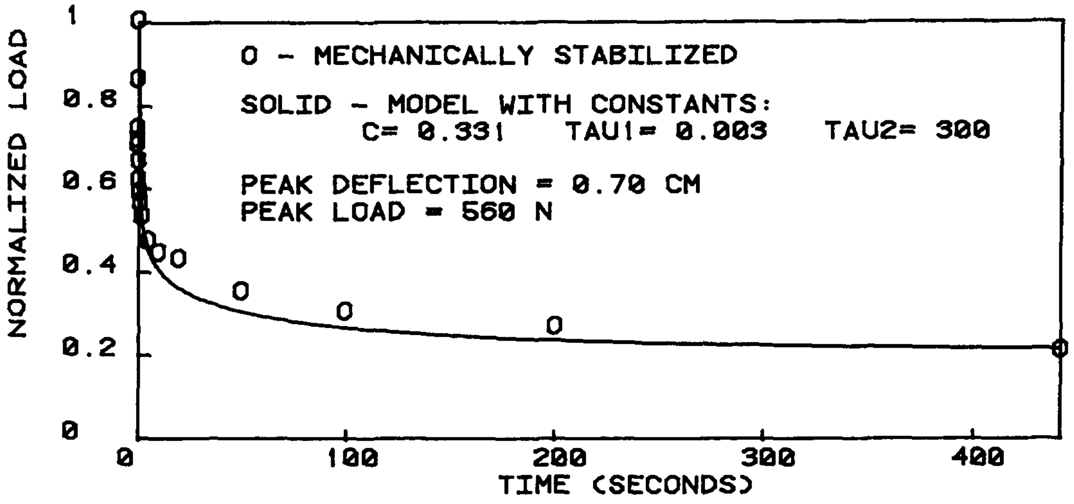


FIGURE 5, TYPICAL RELAXATION TEST FOR HUMAN CERVICAL SPINE A80-384

RELAXATION TESTS

The relaxation tests were performed by applying a ramp displacement of 0.7 cm in 25 msec followed by a constant displacement of 0.7 cm for 5 min. The load-time histories were monitored and recorded. Figure 5 shows a typical relaxation test for a human cervical spine.

A variable rate of load relaxation was demonstrated. Initially, for constant deformations, the load decay was extremely rapid. Thereafter, the load decayed at a much slower rate. This observation renders a standard lumped parameter viscoelastic model with a single dominant long-term time constant a poor predictor of neck behavior. Instead, a generalized Maxwell-Weichert model is proposed since it incorporates an ensemble of decay mechanisms and associated time constants.

CYCLIC MODULUS TESTS

By definition, the cyclic modulus G_c is given by:

$$G_c = \frac{P_c}{L_c}$$

where P_c is the load amplitude and L_c is the deformation amplitude. A sinusoidally-varying compressive displacement of 0.7 cm peak-to-peak amplitude at 20 Hz was applied for 150 cycles to the fully-reequilibrated specimen. The load-time history was monitored and recorded. Figure 6 shows a typical plot of G_c vs. deformation cycles.

Preconditioning behavior was demonstrated. When a specimen was subjected to a repeated deformation history about a fixed length, there was a decrease in the cyclic modulus G_c as the number of deformation cycles

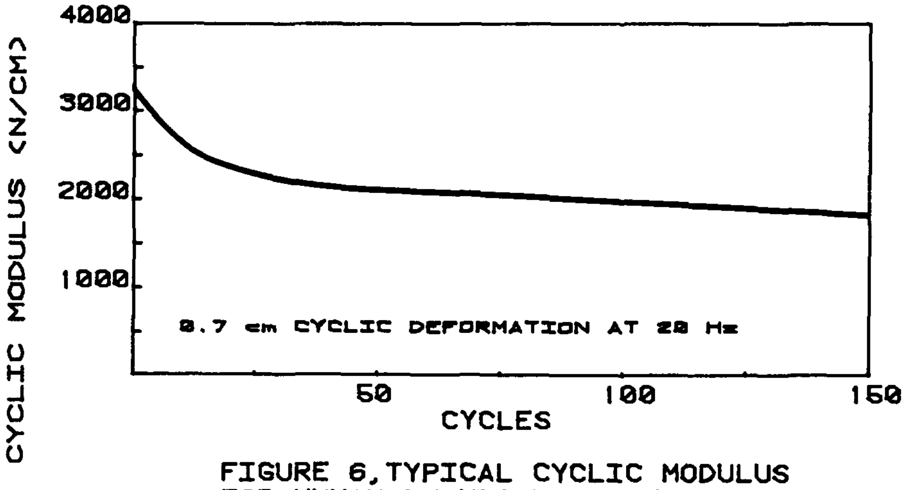


FIGURE 6, TYPICAL CYCLIC MODULUS FOR HUMAN CERVICAL SPINE

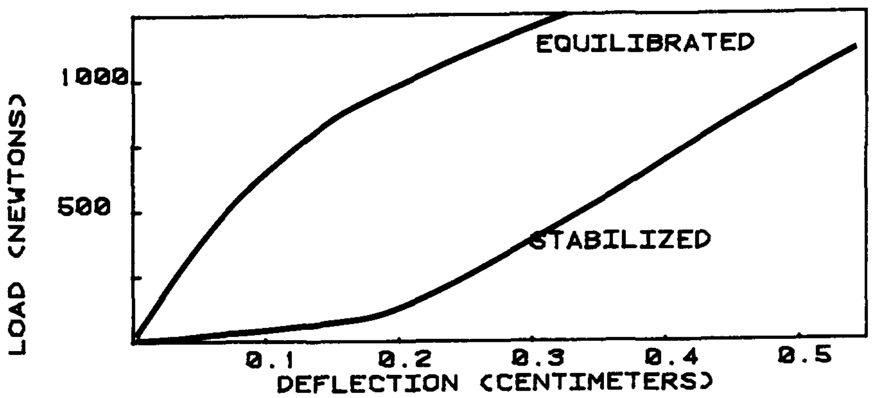


FIGURE 7. TYPICAL ELASTIC RESPONSE FOR EQUILIBRATED AND MECHANICALLY STABILIZED HUMAN CERVICAL SPINE

increased. The initial cycle was representative of the elastic response of the fully reequilibrated cervical spine. Eventually, a steady-state was reached, which we defined as the mechanically stabilized state, where G_C approached a constant value and the load-deflection response was repeatable. The cyclic modulus of the mechanically stabilized state ranged from 45% to 55% of the modulus of the reequilibrated state.

Figure 7 illustates, for a typical relaxation test, the difference between the fully equilibrated elastic response and the mechanically stabilized elastic response.

These cyclic modulus tests demonstrated the influence of the previous load history and the osmotic state of the cervical intervertebral discs on the mechanical response.

VARIABLE RATE CONSTANT VELOCITY TESTS

Figure 8 shows typical results for a human cervical spine in a mechanically stabilized state. The deformation rate was varied by a factor of 500. The stiffness ranged from 1285 to 2250 N/cm, less than a two times increase. The indicated points are measured data.

Deformation rate sensitivity is common in viscous, viscoelastic and plastic materials. For example, the ultimate strength and stiffness of compact bone increases with increasing strain rate [23]. Cancellous bone is sensitive to strain rate to a lesser degree [40]. The intervertebral disc is also sensitive to strain rate to a lesser degree [10].

The human cervical spine exhibits a dependence on deformation rate. This experimental result is consistent with a generalized quasi-linear viscoelastic Maxwell-Weichert model which incorporates a continuous spectrum of relaxation mechanisms and predicts a more distributed deformation rate sensitivity than the standard lumped parameter viscoelastic models.

CONSTANT VELOCITY LOAD-TO-FAILURE TESTS

The last test performed on each mechanically stabilized specimen was the constant velocity load-to-failure test. Ram velocity was nominally 64 cm/sec. By moving the base of the specimen one centimeter in the anterior or posterior direction via the slide and lead screw, the classical extension, compression and flexion injuries were produced.

Table 1 summarizes the type of failure, the maximum load and deflection, and the strain energy or area under the loading portion of the load-deflection curve. Figures 9 through 19 show representative curves.

The following four failure mechanisms were observed as the specimens buckled:

EXTENSION/COMPRESSION - As the body, discs and facet joints resisted the load, the posterior elements were compressed and, as failure of the disc and end plates occurred, the cervical spine extended in a forward buckling mode. Specimen A80-339 failed in this way with rupture of the anterior longitudinal ligament and distraction of the anterior section of the disc between C4 and C5. This occurred with a one centimeter posterior eccentricity.

JEFFERSON FRACTURES - In the clinical literature [18], the common etiology of a fracture of the atlas is a direct blow to the top of the head. In these tests, the experimentally produced atlas fractures, which were usually bilateral and symmetrical, involved the anterior and posterior arches. This was probably due to the compressive force driving the articular condyles outward and bending the arches.

A fourth-order polynomial was used to fit the high rate initial loading curve of the relaxation test. The polynomial used was

$$F^e = 26 + 2600\delta - 5360\delta^2 + 18900\delta^3 - 14700\delta^4, \; \delta < 0.76 \text{ cm}$$

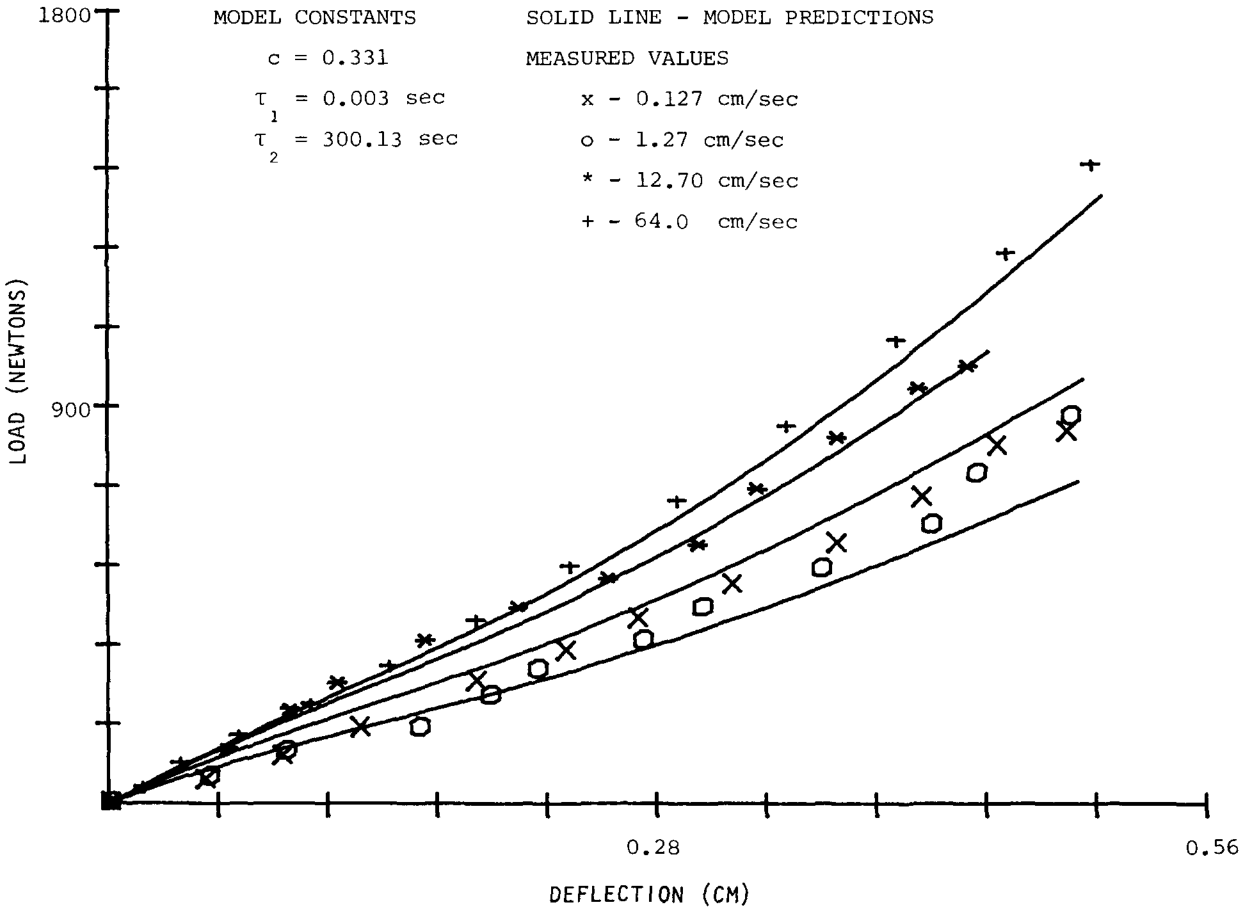


Figure 8, Strain Rate Sensitivity of Human Cervical Spine A80-384

BURST FRACTURES - Comminuted vertical fractures through the vertebral body produced fragmentation of the centrum into a number of large pieces. There were no obvious areas of compressed cancellous bone. Analysis of x-rays taken before and after each test indicated that the specimens that burst were slightly flexed to straight while the specimens that sustained the Jefferson fractures were slightly extended to straight. The burst fractures required larger forces and strain energies than the Jefferson fractures. The load-deflection diagram exhibited a characteristic M-shape or twin peak. Specimen A80-384 showed multiple spikes in the first peak which may be related to the multiple fracturing process.

ANTERIOR WEDGING - The addition of a small flexing moment arm (h < 1 cm) using the test fixture resulted in compression and fracture of the anterior section of the vertebral body. The addition of a slightly larger moment arm (h = 1 cm) produced buckling rearward. Pieces of the cortical shell were displaced in a random pattern. End plate failure occurred and the intervertebral disc was disrupted. However, the amount of displacement applied to the specimen did not result in large anterior dislocation or rupture of the anterior longitudinal ligament. By careful alignment and adjustment of the slide-positioning device, we were able to produce fractures similar to those observed clinically. But, after fourteen tests, we had the distinct impression that one or two centimeters forward or backward, right or left, made a tremendous difference in the outcome. Perhaps, this is the reason there is such a wide range of responses to cervical spine compression in the relevent literature [1,2,4,15,21,24-28,31,33,35,41].

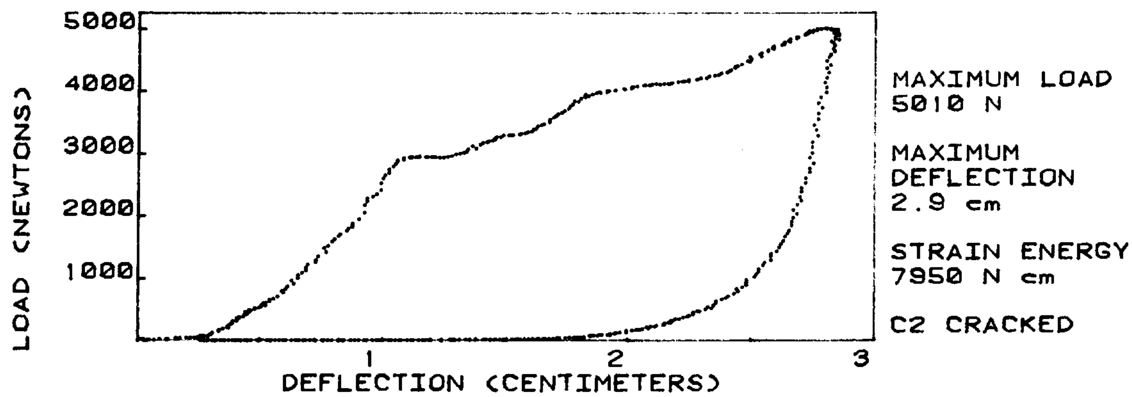


FIGURE 9, A80-269 FIRST FAILURE

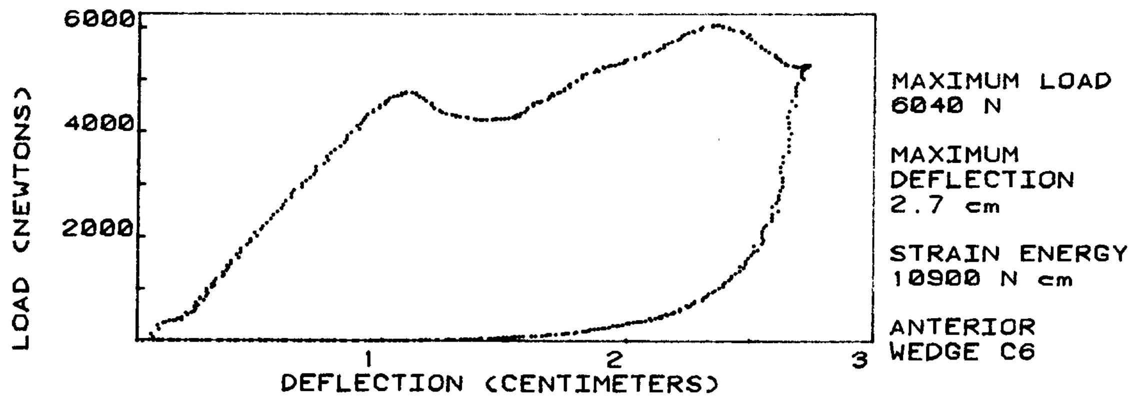
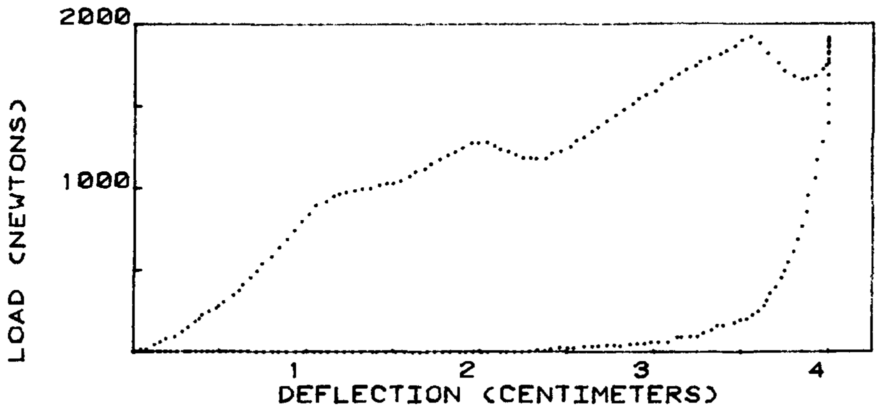


FIGURE 10, A80-269 SECOND FAILURE

TABLE 1

Specimen No.	Age (years) Sex	Description	L_o (cm)	α_o (°)	Ram Velocity (cm/sec)	Failure Mode	C5 Area (cm^2)	Max. Load (N)	Max. Deflection (cm)	Strain Energy (N-cm)
A79-409	58M	B.O.S.* to T2	23	65	50	Jefferson Fr.	5.71	3560	3.0	7470
A79-415	37M	B.O.S. to T1	21	57	50	Compression C5	5.98	5340	3.0	12800
A79-419	49F	B.O.S. to T2	22	63	50	Compression C4&C5	4.29	4860	3.0	10300
A79-423	52M	B.O.S. to T1	19	60	50	Jefferson Fr.	6.17	4190	3.0	7920
A79-431	46M	B.O.S. to T1	20	59	50	Anterior Wedge C5	6.30	4720	3.0	9340
A80-289	70M	B.O.S. to C7	14	60	54	C2 Cracked	5.43	5010	2.9	7950
Retest		C3 to C7	9.1	10	57	Anterior Wedge C6		6040	2.7	10900
A80-339	62F	B.O.S. to T1	21	63	84	Extension Failure	3.51	1930	4.0	4480
A80-352	62M	B.O.S. to C6	9.2	65	55	Jefferson Fr.	6.58	3120	3.0	5740
A80-357	46F	B.O.S. to C6	9.5	55	56	Jefferson Fr.	3.71	960	2.9	1800
A80-364	41M	B.O.S. to C6	12	55	45	C1&C2 Fractured	5.62	5270	2.5	8550
A80-368	77M	B.O.S. to C6 C3,4,5 Bodies Fused	11	45	57	C1 Fractured	5.77	3650	2.7	6350
A80-384	64F	B.O.S. to C7	16	47	92	C2 Fractured	4.38	4060	4.5	12300
Retest		C3 to C7	10	30	77	Burst C4 and Anterior Wedge C4&C5		6840	3.5	15500
A83-26	44M	C2 to T2	13	60	88	Burst Fracture C3,C4&C5	5.45	5470	4.4	15600
A83-42	63F	B.O.S. to C6	11	45	87	Burst Fracture C3&C6	3.28	3000	2.8	5550

*B.O.S. = Base of Skull



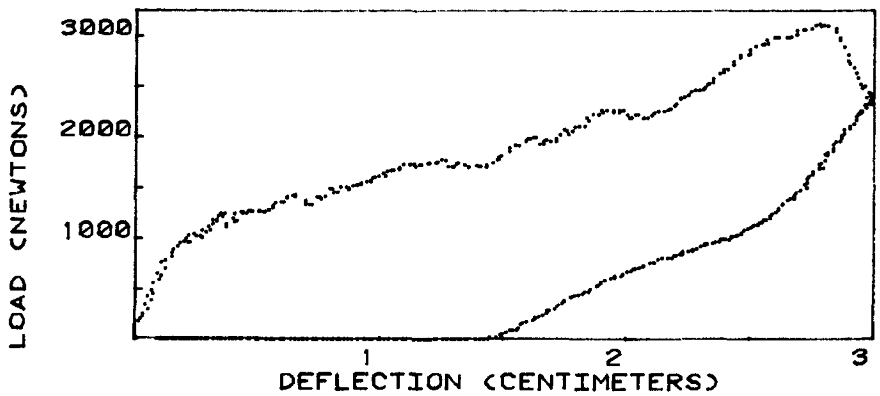
MAXIMUM LOAD
1930 N

MAXIMUM DEFLECTION
4.0 cm

STRAIN ENERGY
4460 N cm

EXTENSION FAILURE
RUPTURED ANTERIOR LONG. LIGAMENT

FIGURE 11. A60-339 FAILURE TEST



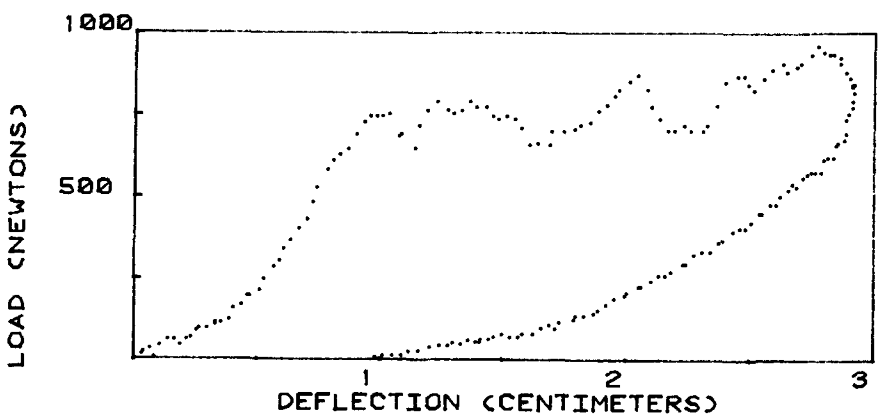
MAXIMUM LOAD
3120 N

MAXIMUM DEFLECTION
3.0 cm

STRAIN ENERGY
5740 N cm

JEFFERSON FRACTURE

FIGURE 12. A60-352 FAILURE TEST



MAXIMUM LOAD
960 N

MAXIMUM DEFLECTION
2.9 cm

STRAIN ENERGY
1600 N cm

JEFFERSON FRACTURE

FIGURE 13. A60-357 FAILURE TEST

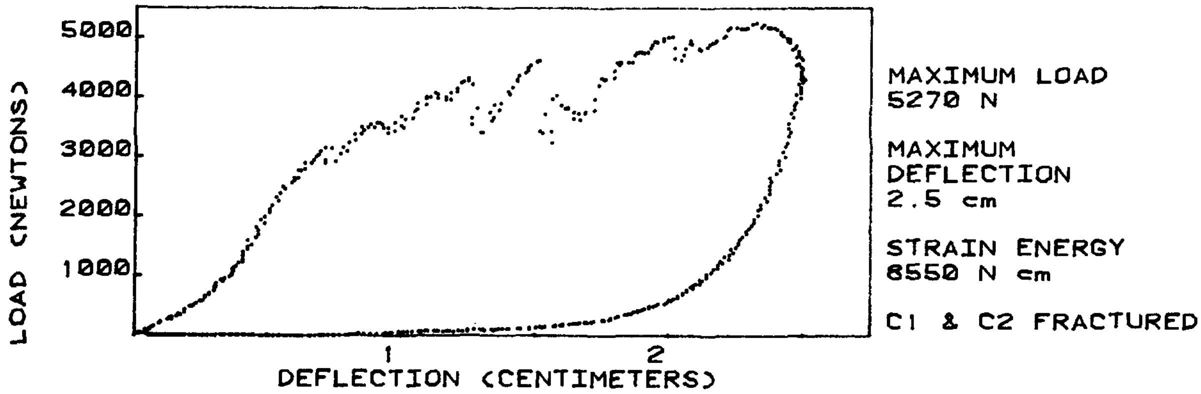


FIGURE 14. A60-364 FAILURE TEST

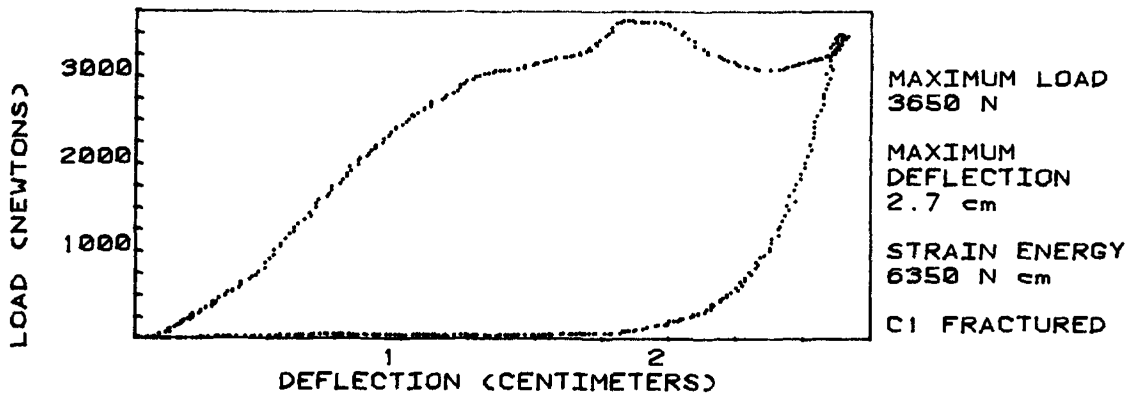


FIGURE 15. A60-368 FAILURE TEST

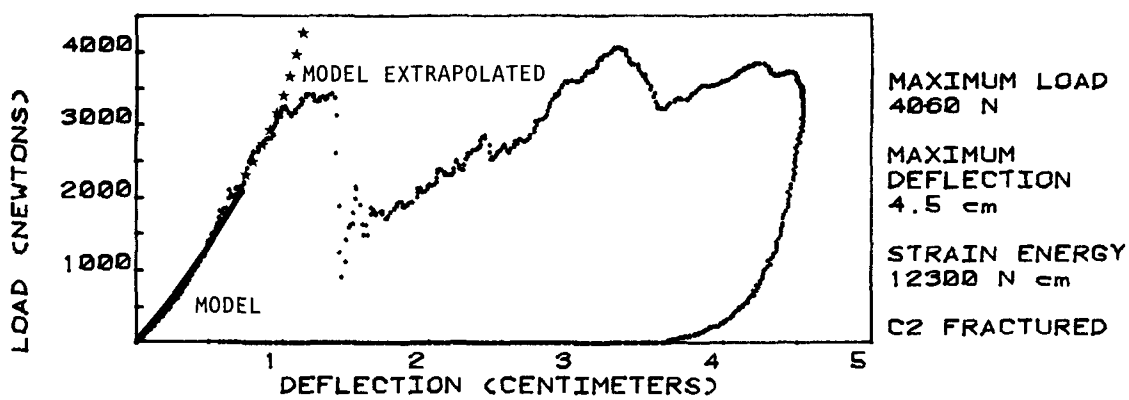
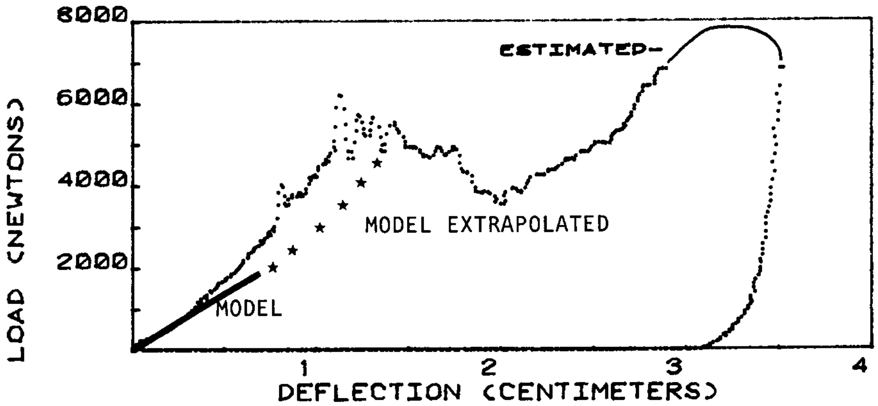


FIGURE 16. A60-364 FIRST FAILURE



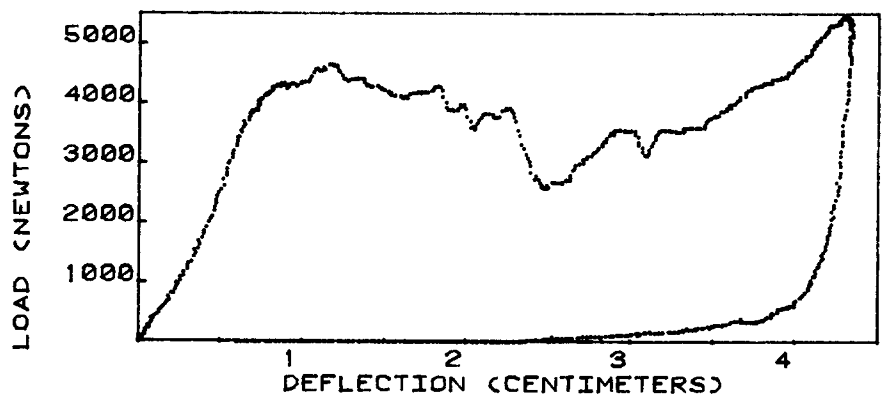
MAXIMUM LOAD
> 6640 N

MAXIMUM
DEFLECTION
3.5 cm

STRAIN ENERGY
> 15500 N cm

BURST FRACTURE C4
ANTERIOR WEDGE C4&C5

FIGURE 17, A80-384 SECOND FAILURE



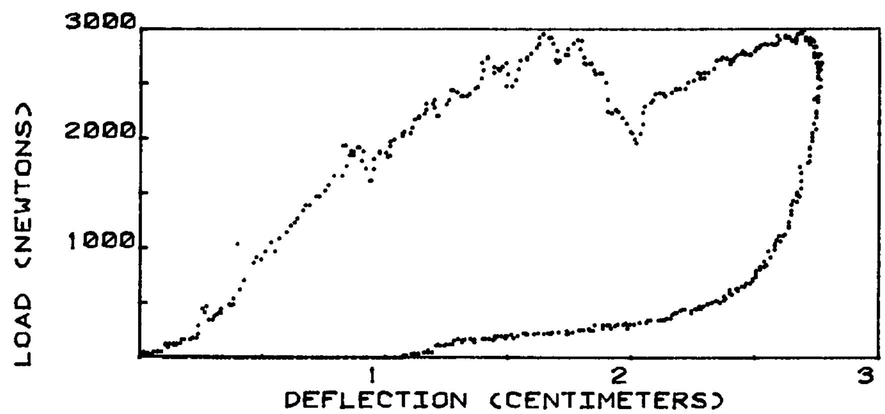
MAXIMUM LOAD
5470 N

MAXIMUM
DEFLECTION
4.4 cm

STRAIN ENERGY
15600 N cm

BURST FRACTURE
C3, C4 & C5

FIGURE 18, A83-26 FAILURE TEST



MAXIMUM LOAD
3000 N

MAXIMUM
DEFLECTION
2.8 cm

STRAIN ENERGY
5550 N cm

BURST FRACTURE
C3 & C6

FIGURE 19, A83-42 FAILURE TEST

MODEL

We have attempted to develop a model that would allow the prediction of the compressive load-deformation responses of the human neck for arbitrary deformation-time histories. A variety of standard linear viscoelastic models were tried and proven inadequate. These included the Maxwell, Kelvin, three-parameter solid and four-parameter solid models. Both the presence of rapid initial load decay for fixed deformations and the fact that the hysteresis loop and the load- deformation response fail to exhibit the strong strain rate dependence predicted by standard lumped parameter viscoelastic models suggest that the behavior is governed by a broad distribution of relaxation times. The model developed is based on the quasi-linear viscoelastic constitutive law hypothesized by Fung (1972). Determination of model constants procedes in the manner used by Pinto and Patitucci (1980) for cardiac muscle, by Casper and McElhaney (1980) for the intervertebral disc, and by Sauren and Rousseau (1983).

In a mechanical sense, we may view this continuous spectrum as arising from a generalized Maxwell-Weichert model. The reduced relaxation function, $Y_r(t)$, may be written as

$$Y_r(t) = \frac{E_\infty + \int_0^\infty E(\tau)e^{-t/\tau}\, d\tau}{E_\infty + \int_0^\infty E(\tau)d\tau} \tag{1}$$

Defining $H(\tau)$, the relaxation time distribution function, as

$$H(\tau) = \tau E(\tau) \tag{2}$$

and substituting into equation (1), we find, after some rearrangement:

$$Y_r(t) = \frac{1 + \frac{1}{E_\infty}\int_0^\infty H(\tau)e^{-t/\tau}\, d(\ln\tau)}{1 + \frac{1}{E_\infty}\int_0^\infty H(\tau)d(\ln\tau)} \tag{3}$$

$H(\tau)$ may be approximated from experimental data as the negative slope of the relaxation modulus vs. logarithmic time plot. This slope is roughly equal to a constant, C, over a major portion of the time domain. Therefore,

$$H(\tau) = -C \quad ; \tau_1 < \tau < \tau_2 \tag{4}$$

$$H(\tau) = 0 \quad ; \tau < \tau_1,\ \tau > \tau_2$$

and the relaxation spectrum becomes:

$$E(\tau) = -C/\tau \quad ; \tau_1 < \tau < \tau_2 \tag{5}$$

$$E(\tau) = 0 \quad ; \tau < \tau_1,\ \tau > \tau_2$$

Defining C* as

$$C^* = -C/E_\infty \tag{6}$$

and substituting equations (5) and (6) into equation (3), we find:

$$Y_r(t) = \frac{1 + C^*\left(E_1(t/\tau_2) - E_1(t/\tau_1)\right)}{1 + C^*\ln(\tau_2/\tau_1)} \tag{7}$$

where $E_1(t/\tau)$ is the exponential integral function.

The relaxation spectrum approximation can be incorporated in the hereditary integral representation to allow the prediction of load-deformation behavior at various deformation rates. Employing $Y_r(t)$ in the quasi-linear viscoelastic representation, we find:

$$F(\delta,t) = \int_0^t Y_r\,(t-\tau)\,\frac{dF^e(\delta(\tau))}{d\delta}\,\frac{d\delta(\tau)}{d\tau}\,d\tau \tag{8}$$

where $F(\delta,t)$ is the force as a function of deformation and time, $dF^e/d\delta$ is the slope of the 'elastic' load-deformation curve, and $d\delta(\tau)/d\tau$ is the change in deformation with time.

In the integral representation, $Y_r(t-\tau)$ is obtained from a relaxation test on a mechanically stabilized neck and $d\delta(\tau)/d\tau$ is the deformation rate. For short test times (less than 0.025 sec for full displacement), the load-deformation curves for the neck undergo rapid compaction. Instantaneous deformation of the neck is impossible to achieve in a physical sense. Therefore, the elastic load-deformation curve is estimated from the 0.025 sec-full stroke load-deformation curve.

For integral calculations, the pseudo-elastic load-deflection curves were computer-fitted to a power series in δ. Little additional information was gained by extending the power series beyond a 4th-order expansion.

MODEL PREDICTIONS

The loading responses of the mechanically stabilized constant velocity tests were predicted using the model constants calculated from the corresponding mechanically stabilized relaxation test data. The theoretical results, plotted as solid lines, are compared to the experimental results, indicated as points, in Figures 8, 14, and 16. The measured constant velocity responses are predicted with reasonable accuracy within the load range of the relaxation tests upon which they were based. However, the prediction of failure loads is beyond the scope of this model. They must, therefore, be determined empirically. We are currently addressing the question of the validity of extrapolating this model beyond the load range of the corresponding relaxation test and for predicting unloading responses. This work is ongoing.

SUMMARY

In the engineering disciplines, a designer starts with a basic building material and shapes it into a structure with specified load and deformation responses. These load and deformation responses are defined as the structural properties. The structural properties are determined by the size, shape, configuration and material of which a structure is composed. In contrast, the material properties are independent of the structure or shape of the material under consideration. Since the human body exists, it exhibits load and deformation responses which determine its injury potential in traumatic environments. Knowledge of the properties of the material of which the human body is composed is useful in so far as it leads to a better understanding of these structural properties.

The structural properties of the cervical spine have been investigated, with particular emphasis on the quantification and prediction of the time-dependent responses to dynamic compression loading.

A key aspect of this research has been the development of an experimental protocol that produces accurate and repeatable test results and is biomechanically significant. The rhesus monkey tests established the validity of delayed post-mortem testing, specimen freezing, and room temperature testing. Results of these tests and the cyclic modulus tests on the human neck have led to procedures for proper specimen storage and the definitions of reequilibrated and mechanically stabilized states. These procedures and definitions are significant in that they have ensured test accuracy and reproducibility.

Relaxation tests were performed on equilibrated, reequilibrated, and mechanically stabilized specimens. An initially rapid and subsequently slow load decay pattern was observed. The dynamic variable rate constant velocity tests were performed on mechanically stabilized specimens. Results indicated a demonstrable deformation rate dependence. Dynamic load-to-failure tests were also performed on mechanically stabilized specimens. A discussion of failure modes, a summary data table, and load-deflection curves indicated that the failures produced were similar to those observed clinically. It was found that small eccentricities (±1cm) in the load axis could change the buckling mode from posterior to anterior.

A mathematical model with constants established from a relaxation test was developed. This model predicts with reasonable accuracy the specimen behavior in constant velocity tests at different rates within the load range of instantaneous elastic response data. Extrapolation beyond this range is probably not justified. We are also currently exploring the model's predictive ability for variable rate loading and unloading tests.

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