Structural Considerations of the Human Vertebral Column under $+G_Z$ Impact Acceleration

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Among the major limitations on manned aerospace vehicular—and escape—system designs are the structural limits of the human body. One of the lower limits is the strength of the vertebral body under $+G_z$ (eyeballs down) impact acceleration. When the vertebral column is considered as a structural member, a finite limit on nonfracturing accelerations can be specified, as has previously been the case. An hypothesis as to the mechanism of fracture, which suggests an approach capable of raising the limit, and experimental evidence in support of the hypothesis are presented. A crude device based upon the approach was designed and tested experimentally on cadaveric exposures to $+G_z$ acceleration. A statistically significant increase in the level of acceleration required to cause fracture was measured.

Theoretical Background

DURING ejections of aviators from jet aircraft, a considerable number of vertebral fractures occur. These are not due to blunt trauma to the vertebrae, nor to striking the aircraft or ground, but appear to occur during or slightly subsequent to initial application of vertical $+G_z$ (eyeballs down)[1] impact forces to the vertebral column.¶ Several hypotheses have been advanced to explain these fractures. Most of the proposed explanations concern simply peak acceleration values, and peak rates of onset of acceleration, and view the vertebral column as a single structure or series of single structures having a physical failure point which, when exceeded, causes structural failure of one or more vertebrae.

It is hypothesized by others, however, that such structural failure limits can be increased by altering body positioning. In this view, opening the angle that the longitudinal axis of the torso forms with the vertical axis of the unrestrained pelvis, by means of increasing the seat angle, will prevent fracture, by arranging the impact vector of the ejection thruster so that it does not occur normal to the superior or inferior surface of the thoraco-lumbar vertebral bodies.

The theory is hereby proposed that one of the major causes of ejection vertebral fracture is the dynamic reaction of the vertebral column under $+G_z$ (eyeballs down) impact acceleration in the presence of improper restraint; that is, there are certain movements of the individual vertebral bodies under $+G_z$ impact acceleration that cause the characteristic ejection vertebral fracture. If this theory is correct and if these motions are prevented, the fractures would therefore be expected to occur only at markedly higher levels.

This may be explained by considering the vertebral column as a series of spring-mass systems, with the intervertebral disks serving as springs, the vertebral bodies (and body segments that they support) as individual masses, and the anterior and posterior interspinous ligaments as spring limiters (on tension only).

Figure 1 demonstrates an idealized vertebral segment consisting of three vertebrae acting as masses and two intervertebral disks acting as springs. If this segment acted as a simple spring-mass system, an impact acceleration in the vector normal to the superior or inferior surface of the vertebral bodies and passing through the center of mass should result in plateau compression fracture(s), as noted in Fig. 2, if the compression failure limit of the superior surface of the vertebral body is exceeded. (Acceleration vector direction is shown in the figures by an arrow). Such fractures are rarely seen in ejection vertebral fractures, however.

If an acceleration is applied with the vector either before or behind the center of mass, anterior or posterior wedge compression fractures, respectively, should result if the failure limit of the superior surface of the vertebral body is exceeded due to anterior or posterior bending of the multipple spring-mass system, as illustrated in Fig. 3. However, recent examination of eighty ejection vertebral fracture cases[2] showed that almost all were anterior compression fractures, as demonstrated in

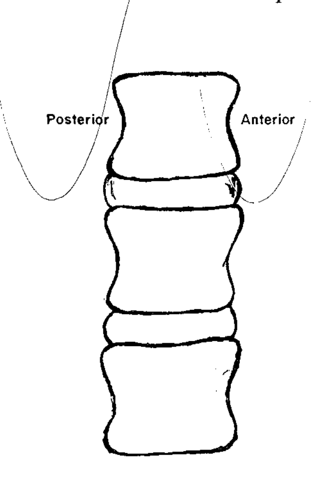

Fig. 1 Diagram of a typical thoraco-lumbar spinal segment.

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§ The theoretical analysis of the problem of ejection vertebral fracture, hypothesis of mechanism of injury, and proposed means for preventing such injuries were originated by the coprincipal investigator (C.L.E.)

¶ Acceleration terminology used in this paper is that adopted as AGARD standard. Thus, $+G_z$ refers to eyeballs down acceleration exerted on the man or cadaver.

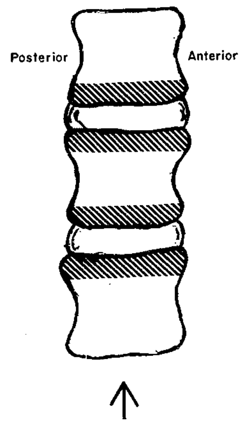

Fig. 2 Diagram of a spinal segment showing areas affected by plateau fractures.

Fig. 4; only one posterior compression fracture was found.** One possible explanation is that the posterior compression of the system is subject to some spring-limiting mechanism whereas anterior compression is not. Examination of the anatomy reveals this to be the case.

The posterior compression limiter (or spring limiter) is seen to be the articular facets of the vertebrae, held together by ligaments, as demonstrated in Fig. 5, that serve as a posterior hinge for adjacent vertebrae. As can be seen in Fig. 6, this hinge allows anterior vertebral lips to touch but prevents any contact of the posterior vertebral lips. Thus an additional spring-limiter system other than the anterior and posterior interspinous ligaments is acting.

An hypothesis is therefore presented: posterior compression of the vertebral column in the thoraco-lumbar area is limited by the articular facets of the vertebrae while anterior compression is not limited. If this hypothesis is true, a means of preventing anterior vertebral compression fracture is suggested.

If the vertebral column could be forcibly restrained during application of impact acceleration in a position of relative hyperextension as in Fig. 7, anterior compression would be limited. Therefore, to cause a compression fracture, the spinous processes connecting the articular facets to the vertebral body would have to be fractured or the ligaments that bind the articular facets together would have to be torn. The forces required to cause fractures of the spinous processes with restraint in a position of hyperextension as in Fig. 8, would be increased quite considerably over those required to cause anterior vertebral fracture while restrained in the conventional erect manner. Thus the vertebral fracture threshold limit would be markedly increased.

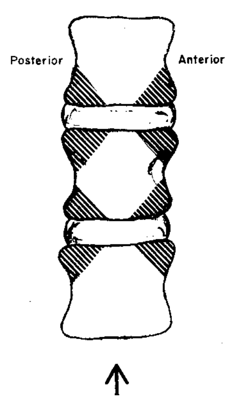

Fig. 3 Diagram of a spinal segment showing areas affected by compression fractures.

** Autopsies performed on ejection fatalities rarely include vertebral body examinations. The conclusion was based upon examination of the x-ray report in each instance of those confirmed by x-ray as having suffered a vertebral fracture and survived. Those fatally injured at time of ejection were not included.

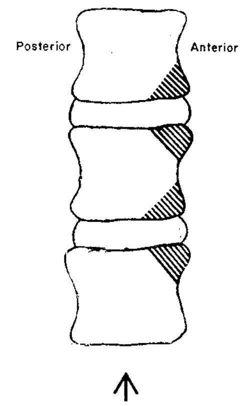

Fig. 4 Diagram of a spinal segment showing areas affected by ejection vertebral fracture.

This theory is supported by Vulcan and King[3] who showed that the vertebral column was subjected to significantly high bending strains, and that these strains were highest along the anterior aspects of the vertebral bodies. The cause of these strains was attributed to eccentricity of location of center of gravity of head and torso, with respect to that of the vertebral column, resulting in forward rotation of head and torso during $+G_z$ acceleration.

Apparatus and Procedure††

Cadavers

Medical records for each cadaver were screened, and cadavers were rejected if: 1) they were over 70 yr of age at time of death; 2) the cause of death was thought to exert any considerable effect on the strength of the vertebral body; 3) the specimen was over 180 lb or under 120 lb; 4) x-rays showed any evidence of prior vertebral fracture, abnormal curvatures, congenital malformations, or an unacceptably high degree of arthritis or decalcification. Sex selection was not made. The remaining cadavers were then prepared for experimental exposure. Data concerning the cadavers used in these experiments are listed in Table 1.

Selected cadavers were subjected to abdominal evisceration to facilitate instrumenting vertebral bodies and maintaining the integrity of the electrical connections under the experimental acceleration. This procedure lightened the loads that normally would be exerted on the vertebral column during acceleration. However, the experimental design dictated that each cadaver act as its own control by being subjected, with identical instrumentation, to a series of incremental accelerations. The cadavers were then limbered up by kneading and

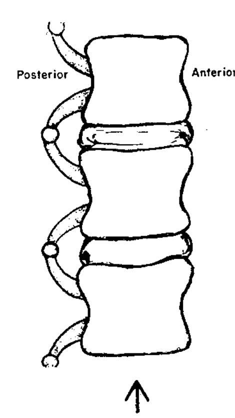

Fig. 5 Diagram of an erect spinal segment showing the relative position of the articular facets and spinous processes.

†† The experiments designed to prove or disprove the hypothesis as previously outlined were performed under the direction and supervision of the coprincipal investigator (A.I.K.).

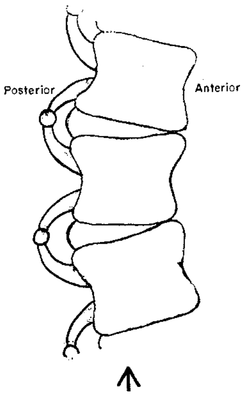

Fig. 6 Diagram of a spinal segment showing relative positions of the vertebral bodies during flexion.

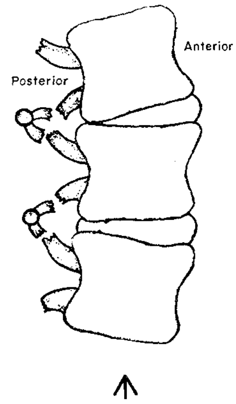

Fig. 8 Diagram of a spinal segment showing the mechanism required to permit posterior compression fractures of the vertebral bodies.

manipulation of the vertebral column to lessen the effects of the embalming process and rigor mortis per se.

The vertebrae to be instrumented were identified by examination of the vertebral column within the abdominal cavity. The crest of the ilium was identified, and this served as a marker for the superior border of L4 by identifying the marked posterior curving of L5 and by location of the lowest rib and its attachment to T12. By these means, L1 was definitely located and served as the basic anatomical reference thereafter.

Each vertebral body to be instrumented was cleaned by cutting and scraping off the ligament around the site to be instrumented, taking care not to injure the bone and to ensure the maximum possible continuity of the ligaments. The exposed bone surfaces were cleaned and dried with acetone prior to mounting the instrumentation.

Accelerator System

All experiments were performed on a vertical accelerator housed in an eight-story elevator shaft of the School of Medicine at Wayne State University. The sled has an aircraft ejection seat mounted directly above the piston. The angle between seat pan and seat back is 90°. Accelerator line of thrust is parallel to the seat back. The cadaver was positioned in the seat prior to firing. The acceleration pulse was induced with the sled at rest at the bottom of the shaft; stroke length is 8 ft. At completion of the input acceleration pulse, braking was initiated and deceleration completed within about 40 ft. The acceleration pulse was approximately trapezoidal in shape, the rate of onset and the magnitude of the plateau being variable. Details of the accelerator have been described by Partick.[4]

Instrumentation and Data Train Systems

Foil-type strain gages, 0.125 in. in length, were bonded to the cleaned surfaces using Eastman 910® adhesive and catalyst. Strain gages were installed on the anterior surface of the vertebral body in the midsagittal plane, and lateral gages were mounted in bilaterally symmetrical pairs coplanar in the coronal plane of T12, L2 and L4 on cadavers 1 and 2. Cadaver 2 also had an anterior gage on L1 and two posterior gages on L2. On cadavers 3–10 the only gages installed were anterior ones on each vertebra from T11–L4. On cadavers 11 and 12 only an anterior gage on L4 was installed, in order to preserve the integrity of the anterior spinous ligament, as a control. All gages were applied with their sensitive axes parallel to the vertical axis of the vertebral body, and all leads were presoldered to the gage terminals. An attempt was made to install these gages near the neutral axis of the vertebra so that their output indicated predominantly axial compression.

The gages were tested, and a minimum of 500 MΩ of resistance to ground was obtained to ensure long-lasting and noisefree outputs. After testing for proper operation, the installed gages were coated with Gagekote #3® to insure insulation from body or embalming fluids. The leads from the gages were attached by sutures to an adjacent body part to prevent accidental removal during exposure to experimental conditions.

After the cadaver was positioned in the accelerator chair, the leads from the lateral strain gages were connected to form diagonally opposite arms of a four-arm Wheatstone bridge; the other two arms were 121-Ω high-stability precision resistors. This configuration resulted in summing of the output of the left and right gages to eliminate the effects of lateral bending of the vertebral column. The anterior gage formed a two-arm bridge with a 121-Ω resistor, with the other two arms being provided by the bridge balance unit.

The following gage identification symbols were used: the prefix A denoted a gage on the anterior surface of T12. The prefix D denoted gages mounted on the lateral surface of the vertebra. Strain values of DL2 represented the average output of the two gages on the lateral surface of L2 on either side of AL2. Posterior gages were denoted by the prefix DD.

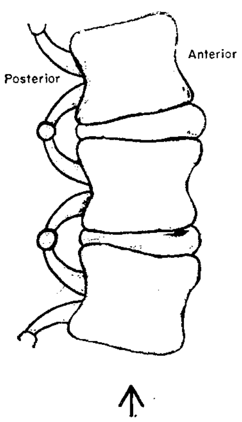

Fig. 7 Diagram of a spinal segment showing relative positions of the vertebral bodies during extension.

Table 1 Data concerning experimental cadavers

Cadaver no.	Ref. no.	Age at death	Cause of death
1	1584	61	Unknown
2	1615	63	Subhepatic abscess
3	1634	61	Carbon monoxide asphyxia
4	002	63	Congestive heart failure
5	061	57	Cirrhosis of the liver
6	1665	49	Tuberculosis
7	930	60	Carcinoma of the tongue
8	017	54	Unknown
9	062	56	Pneumonia
10	095	64	Pneumonia, hemophilia
11	125	62	Unknown
12	127	64	Cardiovascular atherosclerosis

Sled acceleration was measured by a $\pm 50\ g$ strain gage accelerometer (Statham A-6-50®). Total shoulder harness load from shoulder to seat back was measured by a 1000 lb load cell. On certain runs, seat pan load was measured by a 6000 lb strain gage load cell.

Outputs from the transducers went to bridge balance and carrier amplifier units (Heiland®), and real-time hard-copy write-out was obtained from a light beam recorder (Visicorder®). Three volts were constantly applied across each gage; the outputs were adjusted independently of the gage voltage.

Hyperextension Device System

The hyperextension devices consisted of wooden blocks, 6 in. × 4 in. × 2¼ in., that were fastened to the seat back, and height of the blocks from the seat pan was adjustable. The long axis of the block was placed horizontally against the seat back so that the block's smallest dimension was between cadaver L1 and seat back.

Placement of the blocks was determined by previous experiments carried out to determine the optimal location, size and shape.[5,6] Based on those findings, the centerline of each block was placed opposite the body of L1 in almost all cases. Due to the difficulties inherent in performing x-rays on a metal seat positioned over an elevator shaft, a wooden mock-up seat was used to determine the location of L1. While L1 was known to be instrumented, and this could be detected easily on x-ray, the posterior exterior anatomical localization of L1 was necessary in order to place the hyperextension device. This was accomplished by using radio-opaque markers on the wooden chair, in the midsagittal plane. The x-ray showed the position of the external markers and the position of L1 relative to them. The cadaver was then placed in the ejection seat, the geometry duplicated by means of identical markers, and the centerline of the block placed next to the centerline of L1. In the case of cadavers 11 and 12 only 14 was gaged and L1 was identified by this method.

Restraint System

The restraint system consisted of an automotive lap belt under a regular aircraft lap belt and shoulder harness, with leg straps. The wrists were tied together and loosely anchored to the seat base without tension by means of a single rope run through an eye bolt in the seat base to prevent flailing of extremities. A rope was also looped around the aircraft lap belt and through the eye bolt to serve as an inverted V. This step was taken so that no load would be transmitted to the arm rest, and to prevent vertical displacement of the lap belt during pretensioning of the shoulder harness and during the experimental acceleration. The lap belt was always snugly tightened, and the shoulder harness was then preloaded to a 20 lb tension before each run except for those runs in the flexed mode (vide infra). Although the head was unrestrained, its initial position was kept approximately vertical by means of masking tape that broke once the head started rotating but did not have sufficient breaking strength to serve as a motion limiter under acceleration.

Method

Instrumented cadavers restrained in the seat were subjected to $+G_z$ impact accelerations in 3 g–4 g (peak) increments, keeping the rate of onset constant at roughly the rate of onset of the standard ejection seat (approx 300 g(sec–500 g/sec). The input acceleration pulse was trapezoidal, with peak plateau amplitudes varying from 5.5 g to 24.5 g. Duration of the input pulse varied from approx 140 m/sec to 300 m/sec and was dependent upon the peak acceleration, since total stroke length was 8 ft. After each impact exposure, the cadaver was x-rayed again, and if no fracture was noted, the next incremental exposure was given. In some cases, the x-rays were developed after the completion of several runs. If fracture occurred in one of these runs, the data from all postfracture runs were discarded. The end point for each cadaver was fracture of a vertebral body demonstrable by x-ray.

At each peak acceleration level, the cadaver was tested in one or more of the following spinal modes: 1) the entire thoraco-lumbar segment was forced into moderate hyper-extension and maintained in this position by the restraint system and hyperextension device—the extended mode; 2) the vertebral column was allowed to assume the normal erect configuration of a seated cadaver, with a lap belt and shoulder harness restraint system—the erect mode; and 3) the cadaver was restrained in a seated position by a lap belt only. In the absence of a shoulder harness, hyperflexion of the vertebral segment was permitted—the flexed mode. On some cadavers anterior strain gages were used on all vertebral bodies from T11 through L4, but on controls only L4 was instrumented to determine the effect of instrumenting the vertebrae on fracture level.

The effect of evisceration on fracture level and measured strain was tested on the two cadavers on which only L4 was instrumented and which were not eviscerated. Retention of the abdominal viscera can only affect the fracture level adversely since some of this weight must be borne by each spinal segment. To determine the effect of the experimental acceleration as a cause of fracture, precise fracture timing relative to onset of sled acceleration by examination of vertebral body strain curves was used. From postrun x-ray studies alone it was not possible to determine whether a detected fracture was due to the experimental acceleration, or to the braking deceleration which may be considered as only a necessary experimental evil and not a factor in the experiment.

Result and Discussion

A total of 75 runs were performed on 12 cadavers. The data may be considered under three different aspects: a) peak sled accelerations and spinal mode at fracture for each cadaver; b) anatomical and x-ray fracture data for each cadaver; and c) strain-time curves for the instrumented vertebrae of each cadaver.

Peak Sled Accelerations

The experimental design and resulting acceleration levels at fracture as well as the spinal mode in which fracture occurred are listed in Table 2. A summary of average peak sled acceleration required to cause fracture for all modes and average age at death of those cadavers is contained in Table 3. The extended mode shows roughly a 50% increase in the peak sled acceleration required to cause fracture over the level for the erect mode, and a 100% increase over the flexed mode. Yet the seat angle, the line of thrust relative to the pelvis, and the cadaveric back structure itself were all unchanged, and the instrumentation and cadaveric preparation were identical in the majority of the runs.

Reference to Table 2 reveals that the peak acceleration at fracture for cadaver 11 in the extended mode is unknown since no fracture occurred, despite peak sled acceleration of 24.5 g. Yet no cadaver run in either the erect or flexed mode escaped fracture. More importantly, examination of the time of fracture (determined from the strain gages) shows that all erect and flexed mode fractures occurred within the duration of the sled experimental acceleration pulse, while none of the extension mode fractures definitely occurred during this period, and only one (cadaver 12) could have occurred during experimental acceleration. Instead, examination of Table 2 shows that for cadavers 9 and 10, run in the extended mode, the vertebral fractures occurred after the end of sled acceleration and thus well into the period of the braking deceleration pulse; and, therefore, these present a different problem. The deceleration pulse,

Table 2 Fracture levels and spinal modes

Cadaver no.	Rate of onset (g/sec)	Peak accl. (g)	Duration (m/sec)	Time after onset (m/sec)	Mode at fracture	No. of modes tested	Fractured vertebra
1	246	11.0	195	116	erect	2	T12
2	320	7.5	238	138	erect	2	T10
3	375	5.5	300	130	erect	2	T11
4	Data lost	14.0	Data lost	110[a]	erect	2	T9
5	418	14.0	208	60–140	erect	2	T11
6	830	7.0	263	120	flexed	3	L2
7	162	11.0	248	85	flexed	3	T11
8	333	9.0	245	195	flexed	3	L2 & L4
9	267	14.0	206	390	extended	1	Tension fracture on superior end plate of L1
10	483	20.0	170	290	extended	1	Tension fracture on superior end plate of T12 & L2
11	280	24.5	140	no fracture	extended	1	No fracture
12	333	12.5	178	not known	extended	1	T8

which was necessarily used for these experiments due to location of the accelerator device, does not occur during operational ejection seat use. Therefore, a fracture due to the deceleration event is only an experimental artifact with regard to the input acceleration of interest. The deceleration fractures that occurred to cadavers 9 and 10 apparently were due to the disruption of the anterior vertebral ligament incidental to instrumentation of the anterior vertebral bodies T11 through L4, which altered the vertebral column dynamic response on deceleration. The time of occurrence of the other fracture in the extended mode (cadaver 12) is not known, and, therefore, has been arbitrarily assigned to the group due to the acceleration pulse for conservative analysis of the data. The difference between the average fracture levels in the erect and flexed modes as compared with the extended mode, therefore, could possibly be more significant if the reverse assignment were made.

A t test was performed for the fracture g levels between the various spinal modes and the results are given in Table 4. The differences in g level between the extended mode and the other two modes were found to be statistically significant ($P = 0.05$). The null hypothesis was rejected despite conservative estimates made for the fracture level in the extended mode.

When the fracture g levels in the various modes were analyzed, it was not possible to use paired sets of data; so, the appropriate values of t were obtained from the equation for unpaired data with unequal samples. In particular, for extended mode the four fracture levels were obtained under slightly different conditions, in that in two of the cases the anterior ligament was left intact while it was disrupted in the other two. Similarly, when the cadaver was not eviscerated, the fracture level would be expected to decrease. Therefore, the actual difference and the probability that it did not occur by chance can only be higher than those given here.

Comparison of the fracture levels obtained in this series (Table 3) must be made with those reported by Ruff.[7] Table VI-8 from his chapter shows maximum tolerance of the individula vertebrae for T12-L1 tested approximately in the erect mode to be 24.5 g and 23.0 g, respectively, and minimum tolerance for the same vertebrae as 18.6 g and 18.2 g, respectively. Yet the findings in the present study indicate an average tolerance in the erect mode of 10.4 $g \pm 3.79$ g and in the extended mode of $17.5g \pm 5.55g$. It is believed that this seeming inconsistency can be resolved on the basis of age. Ruff's specimens were obtained at least in part from accident victims. It is presumed that those accident victims were youthful, whereas the average age of cadavers in the erect and extended modes of the present study was 61 yr. Figure 7 of the study by McElhaney and Roberts[8] shows that strength of the vertebral body in the sixth decade of life is approximately half that in the second decade. Empirical data from aircraft accidents indicate that the majority of individuals suffering vertebral fracture are in their twenties.

If this relationship holds true, therefore, the average fracture level for the erect mode at age 20, which is the one of interest extrapolated from the present study, would be roughly $20g$–$25g$ and for the extended mode would be $35g$–$44g$. Since cadaveric bone is not so strong as living human bone or fresh cadaveric bone, the comparison becomes potentially even more meaningful.

Anatomical and X-Ray Fracture Data

Roentgenograms of the fractures occurring to various cadaveric subjects showed anterior compression fractures for cadaver 7 and cadaver 8, both in the flexed mode; for cadaver 5 and cadaver 2 both were in the erect mode. The fractures that occurred in the extended mode to cadavers 9 and 10 were different from those in the erect flexed modes, were somewhat unexpected from previous pathological data, and were apparently due to tension on the superior end plate of the vertebral body by the attachment of the intervertebral disk and/or ligament to the end plate. If the hypothesis being tested were true, no fractures would occur in the posterior vertebral structures. This was verified since there was no roentgenographic evidence of damage to the posterior structures of the vertebrae for any experimental acceleration exposure.

Table 3 Summary of peak acceleration values at fracture in the three spinal modes

	Fracture level (g)	No. of cadavers	Average age (yrs)
Extended	17.75 ± 5.55	4	61.5
Erect	10.4 ± 3.79	5	61.0
Flexed	9.0 ± 2.00	3	54.3

Table 4 Student's t test of fracture g-levels between the spinal modes

Modes	Sample size	t	P
Extended and erect	9	2.36	0.05
Extended and flexed	7	2.56	0.05
Erect and flexed	8	0.58	>0.50

Strain-Time Histories of Cadaveric Vertebrae under $+G_z$ Impact Acceleration

The strain-gage data were used not only for determination of peak strain but also to determine temporal relationships between strain measured on different vertebrae on the same run and comparisons of strain patterns on the same vertebrae of a cadaver run at the same input acceleration pulse but with a different mode. Strain measurements made on the anterior surface of the vertebral body in the erect mode showed that it was in compression throughout the acceleration pulse in most cases. However, tension did develop during several runs on cadavers 4 and 5. Tensile strains were all below 700 microstrain. Strain measurements made in the same place in the extended mode, however, showed development of tension in the vertebral body during the initial 60 m/sec of the acceleration pulse, generally less than 300 microstrain.

The typical fracture to be expected in the two modes on the basis of these measurements would thus be considerably different but predictable. In the erect mode an anterior compression fracture would be expected; this was demonstrated by x-ray. In the extended mode a tension type of fracture with separation of the end plate would be expected, and this was also demonstrated by x-ray.

The effectiveness of the hyperextension device can be demonstrated by the determination of the reduction in compressive strain resulting from its use. Table 5 is a listing of the average percentage reduction in strain for paired spinal modes from the various vertebrae that were gaged anteriorly. The over-all percentage reduction is also given.

There were 32 runs on eight cadavers in which strain data for the erect and extended modes at the same g level were available, which is a number large enough to permit tests for statistical significance. A paired t test was carried out for each vertebra for the erect vs extended mode. The difference in strain was computed from which the value of t and the probability, P, that the null hypothesis holds, were obtained for each g level. Table 6 lists this information for all six vertebrae for the bimodal comparison. The number of observations at each acceleration was limited due to fracture at relatively low g levels and to the different gaging patterns used on two of the cadavers. The t test was carried out only when there were three or more pairs of data. In general, the differences were significant for the lumbar vertebrae ($P = 0.05$). For a T12, the reduction in strain was still acceptable, with $P = 0.06$. However, for a T11 the statistical significance of reduction in strain due to use of the hyperextension device was not demonstrated. There was also a consistent reduction in strain in the lateral gages, averaging 24.7% for 17 sets of data from six pairs of runs in the erect vs extended mode. A reduction of 7.9% occurred for a pair of posterior gages placed on L4 in cadaver 2.

A limited comparison could also be made for runs in the flexed and extended mode and those in the flexed and erect mode, and average percentage strain reduction for each vertebra is listed in Table 5. The over-all reduction between the flexed and extended mode was 58.7%, while that between the flexed and erect mode was 33.1%. The numbers of tests were not sufficiently great for these mode comparisons to be tested for statistical significance, however.

The reduction in anterior strain resulting from use of the hyperextension device can be attributed only to a redistribution of the eccentric compressive load on the vertebrae. Since the lateral gages also indicate a reduction, it is possible that some of this load was being transmitted via the articular facets. The exact mechanism of the observed reduction is yet to be explored, but the posterior structures of the vertebrae did not sustain any damage as a result of the use of the hyperextension device.

Table 5 Summary of average percentage reduction in strain between listed vertebrae in listed modes for anterior gages

	Over-all reduction %	AT11	AT12	AL1	AL2	AL3	AL4
Erect vs extended	44.4	20.1	32.8	40.6	67.7	65.3	43.4
Flexed vs extended	58.7	41.4	41.6	58.5	69.1	82.2	67.7
Flexed vs erect	33.1	77.5	23.7	35.2	46.5	31.4	34.0

One possible reason for reduction of strain in the extended mode is transmission of a considerable portion of the load measured in the erect mode to the seat frame via the hyperextension device used in the extended mode. This was evaluated by making successive runs, one in the erect and one in the extended mode, on the same cadaver at the same acceleration level, using the seat pan load cell measurement in the erect mode as a control.

Results of these runs showed a peak seat load cell value of 599 lb for the erect mode and 592 lb for the extended mode. This indicates that the hyperextension device does not support appreciable vertical loads and that the observed reduction in strain is due to a redistribution of load within the column; that is, there is load transmission by the articular facets.

Conclusions

It is concluded that: 1) the hypothesis proposed has been supported by the data collected and presented; no data were observed to indicate that it was incorrect; and it is possible, for these particular data, to reject the null hypothesis at an acceptable level of statistical significance.

2) Moderate forced hyperextension of the cadaveric vertibral column in the area of L1 by a 6 in. × 4 in. × 2¼ in. wooden block necessitates an increase of 50% in the peak sled acceleration level required to cause anterior compression fracture of the lumbar vertebrae over that required in the erect mode, and the difference is statistically significant.

3) The lumbar anterior vertebral body strain in the extended mode is markedly reduced by such a positioning device below that experienced in the erect mode, and the reduction in strain is statistically significant.

4) No posterior vertebral fractures resulted from any of these experiments.

5) There is a preferred position, therefore, of the lumbar portion of the vertebral column during exposure to $+G_z$ acceleration from any source. This position can be achieved artificially by forcibly restraining the shoulder and pelvis of a cadaver to a rigid seat back and forcibly hyperextending the lumbar vertebral column in the area of L1.

Table 6 Paired t test of strain data

	Vertebra																	
g	AT11			AT12			AL1			AL2			AL3			AL4		
Level	n	t	P	n	t	P	n	t	P	n	t	P	n	t	P	n	t	P
4.5	5	1.63	0.18	7	2.29	0.06	5	3.63	0.02	7	8.03	<.01	5	2.88	0.04	5	3.87	0.02
7.5	...	...	...	...	...	...	...	...	...	3	4.48	0.05	...	...	...	3	6.49	0.02
9.0	3	5.86	0.03	3	19.22	<.01	3	6.97	0.04	3	5.74	0.03	3	3.08	0.09	...	...	...

6) This series of experiments, therefore, has considerable implications both for ejection-seat design and restraint-systems design for any human being subjected to $+G_z$ impact acceleration.

7) It would appear from the evidence presented here that the internal structure of the vertebral column can be so arranged by restraint devices that it can withstand considerably greater loads without fracture than the same vertebral column not so restrained. This implies that the orientation of each vertebral body relative to those adjacent determines in part the sled peak-acceleration value in the $+G_z$ vector at which fracture occurs. Therefore, the characterization of the orientation of the entire vertebral column relative to the applied acceleration vector by a single direction is inadequate to explain the vertebral fracture threshold limits determined experimentally.

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