DYNAMIC LOAD RESPONSE OF THE LUMBAR SPINE IN FLEXION

Anna-Lisa Osvalder* and Pavel Neumann

*Department of Injury Prevention, Chalmers University of Technology,
S-412 96 GÖTEBORG, Sweden
Department of Orthopaedic Surgery, Sahlgren Hospital,
S-413 45 GÖTEBORG, Sweden

ABSTRACT

The biomechanical responses of 48 lumbar motion segments (consisting of two adjacent vertebrae, the intervertebral disc and all intervening ligaments) exposed to loads similar to those in frontal car accidents were determined by means of a new method of applying dynamic (transient) flexion-shear loads. The applied loads caused flexion-distraction injuries in the specimens, so called lap seat-belt injuries. Good repeatability and high accuracy were achieved with the dynamic load application method.

The peak values of the applied load pulses in the present experiments varied between 5-12 g, with a rise time between 5-30 ms and a duration between 150-250 ms. The specimens could withstand loads up to 225 Nm and 720 N in flexion before obvious fractures occurred. Signs of injury in the bony structures were observed at lower load levels. The energy absorption of the specimens varied between 15-35 J. The tensile force affecting the posterior structures were 3-5 kN and the deformation rate of the posterior ligaments during the loading sequence varied between 0.5-2.6 m/s.

The results showed that the magnitude of the applied load pulse and the loading rate determined the degree and severity of spinal injury. The duration of the load pulse did not affect the load and injury response. The specimens could withstand higher loads and absorb more energy when the loading rate was increased, but the deformations at injury were smaller when the loading rate was high. There is thus an indication of viscoelastic behaviour in the specimens. The biological parameters bone mineral content, anterior-posterior length and height of the specimen showed high correlations with the dynamic load response of the lumbar spine.

INTRODUCTION

Very few studies on the biomechanical response of the lumbar motion segment (functional spinal unit, FSU), under dynamic loading conditions, have been reported in the literature so far. Cyclic dynamic compressive loading has been used by Liu et al. (1983) and Hansson et al. (1987) to determine fatigue response of the lumbar spine. Willén et al. (1984) and Fredrickson et al. (1992) produced crush fractures of the thoracolumbar spine during instant dynamic compressive loading. The lumbar spine response to combined dynamic loads in different directions have not been adequately studied so far, nor have the effects of different shapes of the applied load pulse. Studies with human cadavers or volunteers have been carried out to establish tolerance levels for different parts of the spine during crash tests (e.g. Vulcan et al. 1970; Begeman et al., 1973; Prasad et al., 1974 ; Mital et al., 1978; Cheng et al., 1979; Pintar et al., 1989; Yoganandan et al., 1990), but there are great differences in magnitude, direction, composition, duration, and pulse shape of the applied load as well as differences in the biological material used.

Neumann et al. (1992-b) reported a significant strain rate dependency of the lumbar, anterior longitudinal, ligament-bone complex when loaded at low (0.1 mm/s), medium (4 mm/s) and high (200 mm/s) distraction rates. Hukins et al. (1990) found an increase in stiffness with increasing strain rate for lumbar spinal ligaments. Effects of strain rate on strength and failure characteristics of the ligaments in living animals were reported by Noyes (1974), Crowninshield and Pope (1976), and Woo et al. (1981). Hayes (1983) showed that a bone sample, which was rapidly loaded had greater ultimate strength and absorbed more energy than a sample which was loaded more slowly. Carter and Hayes (1976) studied the influence of strain rate on bone compressive strength, and the influence of loading rate on the mechanical properties of vertebral bone was discussed by Panjabi et al. (1973) among others. However, in the case of impact loading of the spine in traffic accidents, the bone and ligaments are exposed to much higher loading rates than reported in these studies.

In car accidents, lap seat-belts can cause flexion-distraction injuries of the lower torso (Smith and Kaufer, 1969; Greenbaum et al., 1970; Dehner, 1971; Denis 1983; Gertzbein and Court-Brown, 1988). These injuries seem to be due to the extreme hyperflexion of the torso around the lap seat-belt, which immobilises the hips. Most of these injuries are localised to the L3-L4 motion segment. In order to increase knowledge of spinal injury patterns, from physiological movements until complete disrupture, static and dynamic methods have been developed for in vitro studies of the lumbar spine response to simulated flexion-distraction injuries (lap seat-belt injuries) (Osvalder et al., 1990,1992; Neumann et al., 1992-d).

The purpose of the present work was to submit lumbar FSUs to different dynamic load pulses (similar to those in frontal car accidents causing flexion-distraction injuries), an thus to determine the load response and deformations of the lumbar spine under these loads.

MATERIAL AND METHODS

Experimental procedure

The experiments were done as follows:

1. Static loading of an intact specimen with a non-injurious physiological load in flexion of 25 Nm and 100 N, so as to precondition the ligaments and to determine the flexion angulation under load;
2. Dynamic flexion loading of the specimen with a moderate, medium or a severe load pulse;
3. Static loading of the specimen once again (25 Nm and 100 N), so as to determine the flexion angulation of the potentially injured specimen, i.e. the possible residual permanent deformation.

Specimens

The experiments were carried out on lumbar functional spinal units (FSU) consisting of two adjacent vertebrae, all the intervening ligaments and the intervertebral disc (muscles excluded). The bone mineral content (BMC), expressed in g/cm, was determined in the two adjacent vertebrae with dual photon absorptiometry (Roos and Sköldborn, 1974). Each FSU was rigidly mounted with stainless-steel screws into aluminium cups and fixed with a two component cement (Plastic Padding). The preparation methods are described in more detail by Osvalder et al. (1990) and (1992).

A total of 48 lumbar FSUs (L1-L2; L3-L4) were tested in six groups. Data regarding sex, age, lumbar level, and bone mineral content were collected for each FSU. Height and anterior-posterior length of the specimen as well as anterior-posterior and lateral disc diameter were measured using a pair of vernier callipers (accuracy 0.1 mm). The different groups of specimens used in the experiments were chosen to be as homogeneous as possible. The bone mineral content varied within the individual spines in a random pattern independently of spinal level. The age of the specimens was set to be as low as possible. Morphological data of the specimens are presented in Table 1.

Table 1 Morphological data of the FSUs used in the experiments. Mean (S.D.) and range are presented for each parameter.

Load pulse	MODERATE		MEDIUM		SEVERE	
	1	2	1	2	1	2
No. of specimens	(N=10)	(N=8)	(N=5)	(N=5)	(N=10)	(N=10)
Age (Yrs)	49(9)	58(15)	49(10)	50(11)	56(12)	58(8)
Range	38-66	34-75	35-58	34-60	38-74	46-79
BMC (g/cm)	3.9(0.7)	3.9(0.6)	3.8(0.6)	3.8(0.4)	3.6(0.9)	3.8(0.9)
Range	3.0-5.1	2.9-5.0	3.3-4.7	3.3-4.3	2.3-5.4	2.4-5.2
Specimen height (mm)	76(7)	80(8)	80(5)	80(4)	80(8)	79(5)
Range	67-92	69-90	73-85	76-85	68-94	73-85
Specimen AP length (mm)	90(3)	89(2)	89(4)	90(1)	89(5)	88(3)
Range	86-94	84-92	85-93	88-92	81-91	83-93
Lateral disc diam. (mm)	57(5)	58(4)	57(3)	56(3)	59(5)	58(5)
Range	51-63	52-63	53-61	52-60	52-67	52-67
AP disc diam. (mm)	40(4)	40(2)	41(2)	40(2)	42(3)	42(3)
Range	36-44	36-42	39-43	38-42	39-46	36-46

Dynamic test set-up

A dynamic (transient) load of combined bending moment, shear force and compressive force (defined as flexion-shear) was transmitted to each FSU in the anterior direction by means of a padded pendulum (Figure 1). The pendulum hit an interface situated on a metal lever about 300 mm above the mid-disc plane of the FSU. Cylindrical weights (12 kg) were attached to the metal lever and acted as a compressive pre-load on the specimen as recommended by Goel et al. (1987). The padded pendulum was arrested by a mechanical stop when it had transferred enough energy to the interface to move forward the whole system (the upper vertebra, the upper cup, and the metal lever with pre-loads and interface) above the joint (the intervertebral disc). The test set-up for dynamic loading is described and discussed in detail by Osvalder et al. (1992).

The shape of the load pulse transmitted to the interface was determined by the force-deformation characteristics of the padding materials (polyurethane padding and polyethylene) attached to the pendulum face and by the impact velocity of the pendulum. After preliminary tests, three test series with six different load pulses were chosen: Moderate 1 and 2, Medium 1 and 2, and Severe 1 and 2 (Table 2). The difference between the load pulses in the three test series was the magnitude of acceleration. The difference between the two load pulses in each series was the loading rate. In these experiments the loading rate was defined as the peak acceleration value divided by the rise time for the acceleration.

The load response of each FSU was measured by means of a three dimensional force and moment transducer (AMTI MC 12-6-1000) situated under the specimen. In order to describe the load response in the sagittal disc plane, the measured bending moment and shear force had to be corrected. The linear acceleration was measured with a uni-directional accelerometer (Entran, range ±25 g) which was attached to the interface. The output signals from the transducers were amplified and stored, and then processed by means of a Macintosh II computer and Lab-VIEW application software (National Instruments). The motion behaviour of the FSU was monitored by high speed photography (500 frames per second). Small silver coloured balls were glued to nails which were partially inserted in the vertebral body above and under the anterior and posterior part of the disc and in the spinous processes. During the high-speed film evaluation, the positions of the balls were determined with an accuracy of ±0.1 mm by means of a digitiser and an analysing program. The horizontal displacement was defined by the distance that the centre of the upper vertebral body had moved, in the anterior horizontal direction, relative to the lower fixed vertebra. The vertical displacement was given by the distance which the posterior part of the upper vertebra had moved relative to its initial position. The flexion angulation for the entire FSU was defined by its rotation in the sagittal plane (Figure 2). The energy absorption was estimated from the load-deformation curve for bending moment versus flexion angulation (in radians).

Static physiological testing

A static flexion-shear load (25 Nm/100 N) was applied to the specimens for four minutes, before and after the dynamic loading sequence. The static load was chosen to produce movements not exceeding the normal physiological range of motion for the lumbar FSU (Schultz et al., 1979; Panjabi et al., 1982; Neumann et al., 1992-c). Since the bending moment had shown to play the major role in causing the flexion-distraction injuries, as reflected by the rotation of the specimen in the forward direction, the flexion angulation of the specimen was chosen to be compared before and after the dynamic loading sequence, to get an indication of a potential instability in the segment. Mechanical displacement gauges were used to measure the vertical displacements of the specimens and thereby the flexion angulation was calculated with an accuracy of ±0.1°. The experimental set-up for static flexion-shear loading is described in detail by Osvalder et al. (1990).

After the subsequent static physiological loading of the potentially injured specimens, each specimen was deep frozen with carbon dioxide in the loaded position, before being stored at -20°C. Radiographs, including sagittal projections on X-ray as well as on computed tomography, were then obtained on the frozen specimens, in the flexed position, to investigate if any fractures or signs of ruptures had occurred in the structures.

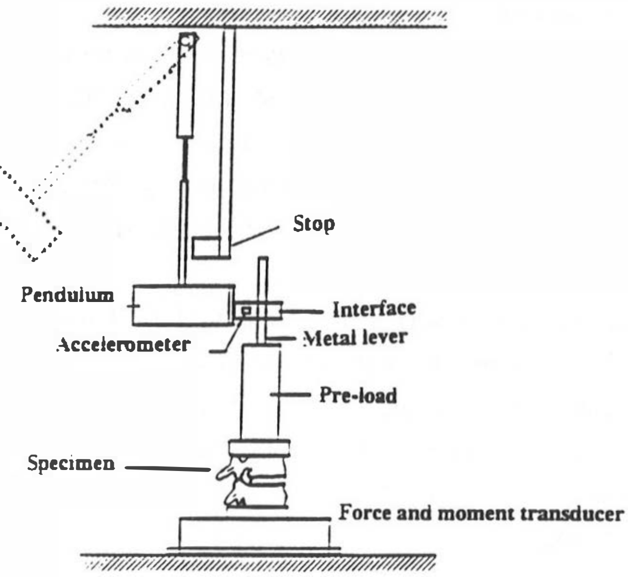

Figure 1 Experimental set-up for dynamic flexion-shear loading.

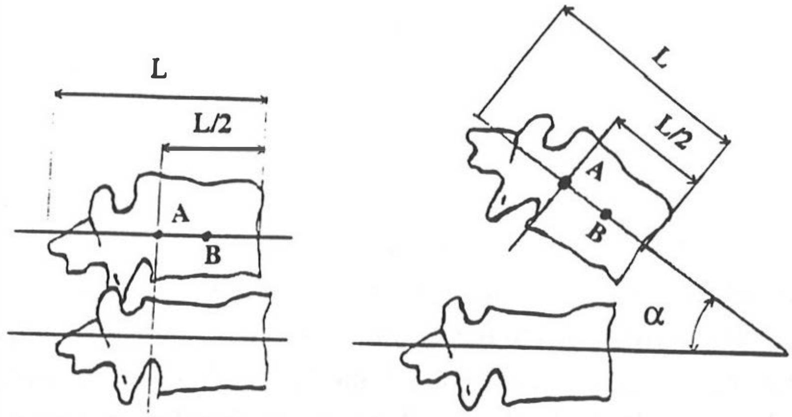

Figure 2 Initial and flexed position of a FSU. L: anterior-posterior length, A: determination of vertical displacement, B; determination of horizontal displacement, α -flexion angulation

Table 2 Description of the dynamic load pulses. Loading rate is defined by the peak acceleration divided by the rise time, and expressed in g/ms (1 g=9.81 m/s^2).

Load pulse	Series 1 MODERATE 1	2	Series 2 MEDIUM 1	2	Series 3 SEVERE 1	2
No. of specimens	(N=10)	(N=8)	(N=5)	(N=5)	(N=10)	(N=10)
Peak acceleration (g)	5	5	8	8	12	12
Rise time (ms)	30	15	30	15	15	5
Duration (ms)	150	140	200	180	250	230
Loading rate (g/ms)	0.17	0.33	0.27	0.53	0.80	2.40

RESULTS

The differences in magnitude, rise time, and duration respectively, between the load pulses in the same group were less than 5%. The moderate and severe load pulses resulted in biomechanical findings close to the static threshold of flexion-distraction injury and ultimate injury to the FSU. The medium load pulses resulted in biomechanical findings in between the results from the moderate and severe load pulses. According to these findings the degree of failure in the specimens varied for the three test series, i.e. the same kind of injuries had not occurred. In each series there was no difference between the two groups regarding the injury signs found in the specimens. The severity of the injury was increased when the magnitude and the loading rate of the applied load pulse was increased.

The results from the three test series are presented in Table 3. The peak values for the bending moment and shear force (in the sagittal disc plane) represent the maximum load the specimen could withstand before an obvious change (a distinct dip) was noticed on the moment and force curves, which could be linked with increased displacements and flexion angulation seen on the the high-speed film. In general, the specimens exposed to the highest loading rates in each test series, withstood higher loads and absorbed more energy before injury occurred.

The reported flexion angulation and the horizontal and vertical displacements (Table 3) represent the maximum values attained before the dips in the moment and force curves occur. In general the maximum deformations before injury occurred were smaller for the specimens exposed to the highest loading rates in each test series. During the subsequent static physiological loading (25 Nm, 100 N), all specimens showed increased angulations compared to those measured before the dynamic loading sequences. This was a sign of that residual, permanent deformations had occurred (Table 4). The flexion angulation of the injured specimen almost corresponds to the maximum flexion angulation attained during the dynamic loading sequence for all groups (Tables 3 and 4). The high-speed photography did not operate satisfactorily for moderate load pulse 2 and it was therefore not possible to report values for the flexion angulation, horizontal and vertical displacements, and energy absorption for these experiments.

The distinct dip always occurred at the same time on the moment-time and force-time curves. For all tests the dip occurs early in the loading sequence (10-60 ms after the application of the load) (Table 3). The duration of the load pulse does thus not affect when the injury dip occurs. For the moderate, medium, and severe load pulses the injury dip occurs at about 15 ms, 10 ms, and 5 ms respectively, after the steady state has been reached for the load, i.e. during this time period the peak value of the load was constant. In each test series there was no difference between the two groups regarding the time period the specimens could withstand the maximum load without being injured, in spite of the loading rate being different. When the magnitude of the applied load was increased the time period the specimen could withstand the maximum load without being injured was decreased.

Table 3 Results from dynamic flexion-shear loading of lumbar FSUs. Mean (S.D.) and range are reported for each parameter.

Load pulse	MODERATE 1	MODERATE 2	MEDIUM 1	MEDIUM 2	SEVERE 1	SEVERE 2
Rise time (ms)	30	15	30	15	15	5
No. of specimens	(N=10)	(N=8)	(N=5)	(N=5)	(N=10)	(N=10)
Bending Moment (Nm)	140(11)	148(15)	159(10)	171(10)	185(15)	207(13)
Range	120-160	122-175	145-175	155-185	150-215	175-225
Shear Force (N)	430(36)	455(39)	475(40)	495(42)	600(45)	660(29)
Range	385-530	395-540	415-545	430-575	525-715	605-720
Time first dip (ms)	40-60	25-45	40-55	20-40	20-25	10-15
Flexion Angle (°)	14.1(1.6)	-	16.2(1.2)	15.2(1.1)	19.1(1.3)	18.0(1.2)
Range	11.3-16.1	-	13.9-16.8	13.3-16.2	16.3-21.2	15.1-21.0
Vertical displ. (mm)	14.1(1.2)	-	15.4(1.0)	14.8(0.9)	17.5(1.5)	17.0(1.2)
Range	12.7-15.4	-	14.0-16.2	13.1-16.6	15.8-20.6	15.1-20.1
Horizontal displ. (mm)	6.1(1.0)	-	7.2(1.1)	6.9(1.1)	8.1(1.1)	7.4(1.0)
Range	4.9-8.3	-	6.1-8.5	5.5-8.4	6.9-10.1	6.2-9.6
Energy (J)	14.7(2.6)	-	19.3(2.0)	21.3(1.9)	28.0(3.0)	34.4(3.1)
Range	11.9-17.8	-	16.9-21.6	18.9-23.5	24.5-32.0	30.0-37.8

Table 4 Mean (S.D.) flexion angulation of intact and injured FSU under a static physiological flexion-shear load of 25 Nm and 100 N.

Load pulse	MODERATE 1	MODERATE 2	MEDIUM 1	MEDIUM 2	SEVERE 1	SEVERE 2
No. of specimens	(N=10)	(N=8)	(N=5)	(N=5)	(N=5)	(N=4)
Flexion angle (°) **Intact FSU**	6.4(1.4)	6.6(1.5)	6.7(1.1)	7.4(1.2)	7.4(1.3)	6.8(1.5)
Flexion angle (°) **Injured FSU**	14.3(1.5)	14.2(1.4)	16.4(1.3)	15.4(1.0)	19.4(1.2)	18.5(1.1)

Moderate load pulse

The mean peak bending moment was 140 Nm for moderate load 1 and 148 Nm for moderate load 2. The mean peak shear force was 430 N and 455 N, respectively. For these loads no specimen failed completely, but signs of injury occurred in all specimens. This was shown by a distinct dip in the moment and force curves which indicated that some substantial structure had failed (Figure 3). The injury dip occurred at about 40-60 ms after the load was applied for moderate load pulse 1 and at about 25-45 ms for moderate load pulse 2. All the FSUs in the two moderate groups showed permanent deformations during the subsequent static physiological loading by a flexion angulation of 14° compared to 7° for the intact FSUs (Table 4). No signs of disrupture were noticed on the radiographs. There was no statistically significant difference on the 5% level (t-test) between the two moderate groups regarding load response (maximum bending moment and shear force). Neither was any statistically significant difference shown between the two moderate groups regarding the flexion angulation from the subsequent statical physiological loading.

Medium load pulse

The mean peak bending moment was 159 Nm for medium load 1 and 171 Nm for medium load 2. The mean peak shear force was 475 N and 495 N, respectively. For the medium load pulses 1 and 2 no specimen failed completely but signs of injury occurred in all specimens as reflected by the distinct dip in the moment and force curves. The injury dip occurred at about 40-55 ms after the load was applied for medium load pulse 1 and at about 20-40 ms for medium load pulse 2 (Figure 4). During the subsequent static physiological loading all the FSUs showed a flexion angulation of 15-16° compared to 7° for the intact FSUs (Table 4). Evident signs of disrupture were noticed on the radiographs in one of the specimens from medium group 1 and in two of the specimens from medium group 2. There was no statistically significant difference on the 5% level (t-test) between the two medium groups regarding load response (maximum bending moment and shear force), flexion angulation, horizontal and vertical deformation. However, a significant difference ($p<0.05$) was shown in energy absorption between the two groups. No statistically significant difference was shown between the two medium groups regarding the flexion angulation from the subsequent statical physiological loading. However, only five specimens were tested in each group.

Severe load pulse

The mean peak bending moment was 185 Nm for severe load 1 and 207 Nm for severe load 2. The mean peak shear force was 600 N and 660 N, respectively. Obvious fractures occurred in five specimens exposed to severe load pulse 1 (after 20-25 ms) and in six specimens exposed to severe load pulse 2 (after 10-15 ms). For these specimens the flexion angulation just before failure was ≥19°. For these totally injured specimens the subsequent statical physiological testing could not be performed. The remaining 9 specimens showed large dips in their moment and force curves, but managed to withstand the whole loading sequences without undergoing complete failure. For these specimens evident signs of disrupture were noticed on the radiographs. Subsequent static physiological loading of these specimens demonstrated clear signs of disrupture in the posterior elements by a flexion angulation of 18-19° compared to 7° for the intact FSUs (Table 4). Figure 5 shows typical load-time curves for specimens exposed to the severe load pulses. There was a statistically significant difference (t-test) between the two groups regarding load response ($p<0.05$) (maximum bending moment and peak shear force) and energy absorption ($p<0.01$). There was also a significant difference between the two groups ($p<0.05$) regarding maximum flexion angulation and maximum horizontal displacement before the injury dip occurred during the dynamic loading sequence.

Influence of biological parameters

For each group a linear regression analysis showed that the highest correlations were found between the load response and the bone mineral content, the anterior-posterior length of the specimen and specimen height. Spinal level, age and sex did not explain the variation in the results.

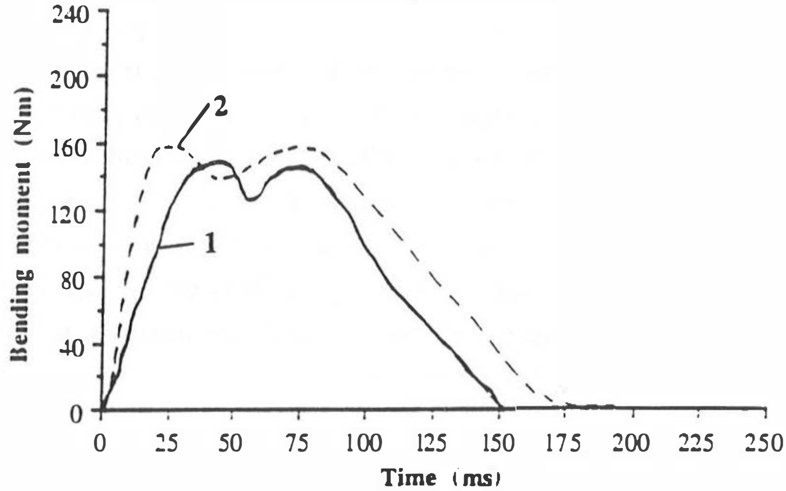

Figure 3 Bending moment versus time for lumbar FSUs exposed to the moderate load pulses 1 and 2.

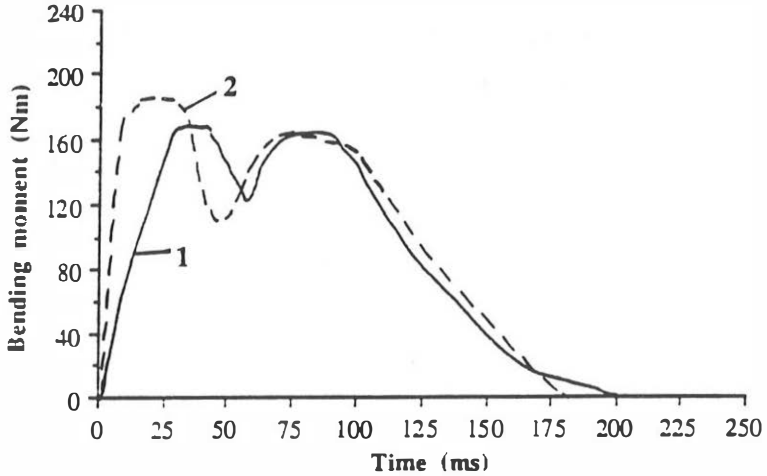

Figure 4 Bending moment versus time for lumbar FSUs exposed to the medium load pulses 1 and 2.

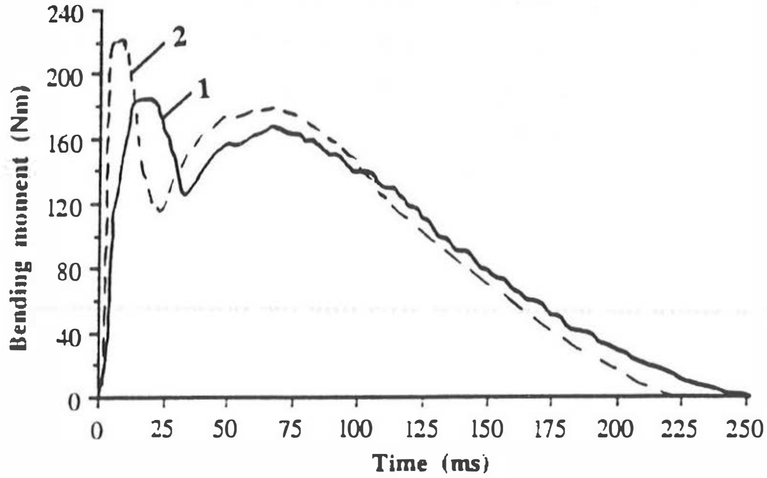

Figure 5 Bending moment versus time for lumbar FSUs exposed to the severe load pulses 1 and 2. Evident signs of rupture occurred in the posterior elements, but the specimens managed to withstand the load during the whole loading sequence

DISCUSSION

The purpose of this study was to expose lumbar FSUs to different dynamic (transient) load pulses and thereby investigate the load response and deformations of the specimens under combined flexion-shear loads. The different dynamic load pulses used, caused injuries of different degrees to the segments, varying from subtotal flexion-distraction injuries (residual permanent deformations) until complete failure. An attempt to gain further insight into spinal injury patterns under dynamic loading conditions has been made. The biomechanical results and the explanation of the biological variation of the spine obtained would lead to a better understanding of the injury mechanisms. This information could be useful in the validation of mechanical and mathematical models of the spine. This would also help the development of preventive measures that would reduce the risk and minimise the severity of spinal injuries, particularly those caused by traffic accidents.

The experimental set-up used in this study was designed to simulate a dynamic flexion-distraction type of injury to the lumbar spine in vitro. Magnitude, rise time and duration of the load pulses used should imitate the load distribution of the individual lumbar spine segments during frontal impacts. Variation of the parameters was based upon the results from sled tests, with human cadavers and volunteers, during frontal impacts undergoing deceleration levels of the sled from 5-15 g for cadavers and 2-8 g for volunteers (Begeman et al., 1973; Mital et al, 1978; Cheng et al., 1979).

The moderate, medium, and severe load pulses caused different degrees of trauma, from initial subtotal injury until total failure of the specimens. All 18 specimens exposed to the moderate and medium load pulses showed residual permanent deformations., but no obvious fractures were shown in the vertebrae. The severe load pulses caused obvious fractures in 11 of the 20 FSUs tested, i.e. these specimens failed completely during the loading sequences and could therefore not be exposed to the subsequent static physiological loading. The evaluation of the radiographs showed evident signs of disrupture in the posterior elements in the remaining 9 FSUs exposed to the severe load pulses. It is difficult to compare the results of this study with other dynamic experimental studies of the lumbar spine because of different experimental set-ups and loading directions. As far as we know, no experimental study on the lumbar spine has been reported regarding dynamic flexion-shear loading. The static load response of the FSU at the threshold of flexion-distraction injury and ultimate injury have been reported by Osvalder et al. (1990) and Neumann et al. (1992-d).

In the present study the tensile force affecting the posterior structures of the FSU just before failure was estimated by the maximum bending moment at the center of rotation (situated in the posterior part of the lower vertebra, close to the disc) divided by the distance between the centre of rotation and the centre of the posterior structures. This estimation resulted in a tensile force of 3-5 kN affecting the posterior structures, which is higher than the reported tensile force of about 2.8 kN for static ultimate flexion-shear loading of the lumbar FSUs (Osvalder et al. 1990).

The results in this study indicated that the load response and the energy absorption of the FSUs increased with increasing loading rate. The results also indicated that the deformations at injury were less when the loading rate was increased. This might be an indication of viscoelastic behaviour of the different structures of the FSU (bone, ligaments, disc). Viscoelastic properties have also experimentally been demonstrated for different other biological structures or whole body parts (e.g. Viano and Lau, 1988; White and Panjabi, 1990; Neumann et al., 1992-b). Viano and Lau (1988) presented human impact tolerances for compression and showed that at a velocity of deformation of about 2 m/s (differs somewhat depending on the biological structure) the 'viscous injury zone' starts. Their studies demonstrated an interdependence between the velocity of deformation and the compression of the body on injury risk.

In order to get an indication of the deformation rate of the posterior structures of the FSUs during the dynamic loading, the mean deformation rate was estimated by the vertical deformation produced in the posterior structures of the FSU before injury occurred, divided by the duration for the deformation. The estimation resulted in deformation rates between 0.5-2.6 m/s, the highest rates were found for the severe load pulses. These rates are much higher, especially for the severe load pulses, than the deformation rates used in dynamic experiments of different structures (bone, ligaments) of the lumbar spine.

The significant differences between the two severe load pulses regarding load and deformation could be explained by that the viscoelastic properties for the FSU are more pronounced during the severe dynamic loading. During the severe loading sequences the estimated deformation rates for the posterior structures of the FSUs was 1.9-2.6 m/s. For the moderate and medium load pulses, the elastic properties of the FSU may be more pronounced than the viscoelastic properties (deformation rates 0.5-1.5 m/s). Thereby the load response of the FSU during the moderate and medium dynamic loading was more like the static case. The duration of the load pulse does not influence the load and injury response of the FSU in these experiments. The first major injury sign (demonstrated by a distinct dip in the force and moment curves) always occurred very early in the loading sequence, 10-60 ms after the load was applied, i.e. 5-15 ms after the steady state has been reached for the load. In all experiments only one distinct injury dip was shown on the moment and force curves.

The results indicate that the magnitude of the load pulse, together with loading rate seem to be the parameters which primarily determines the degree and the severity of the flexion-distraction injuries. This information could be useful, for example, for the validation of mathematical and mechanical models of the spine. When preventive measures for the spine during impacts are developed, it is important to decrease the peak value of the impact load as well as decrease the velocity of the deformation of the different structures of the spine, which could reduce the risk and minimise the severity of spinal injuries.

As pointed out by Schultz et al. (1979) among others, individual differences in cadaver segment behaviour are large. The specimens used in the different test series in the present study were chosen to be as homogeneous as possible. The mean value for the age was around 55 years, but the age range was wide (38-74 years). The three morphological parameters bone mineral content, anterior-posterior length and height of the specimen showed high correlations with the load response for all groups. It seems that the bone mineral content in the vertebrae could explain much of the variation in the load response in both the moderate and severe test series. The size of the segment obviously plays an important role for the variation in the load response. It should be pointed out that there is an interdependence between the anterior-posterior length of the FSU and the bone mineral content as determined by means of dual photon absorptiometry (Roos, 1975), and there is also an interdependency between the height and the anterior-posterior length of the FSU. The bone mineral content has explained much of the variability of the results from experiments carried out on different tissue components of the lumbar FSU (bone, ligament, disc) during static compressive loading (Hansson et al., 1980; Keller et al., 1987; Granhed et al., 1989). Significant correlation was found between the bone mineral content in the adjacent vertebrae and the structural properties of the lumbar, anterior, longitudinal ligament-bone complex during quasi-static and dynamic tensile testing (Neumann et al. 1992-a and 1992-b). The results of these studies indicate that, in the FSU, not only the bone but all the connective tissue components should be considered as an entity subjected to fractions of the same external load.

CONCLUSIONS

1. During dynamic flexion-shear loading the lumbar FSU can withstand higher loads and absorb more energy before an injury occurs when the loading rate is high. The deformations at injury tend to be smaller when the loading rate is high. This is an indication of viscoelastic behaviour of the FSU.

2. The magnitude of the load pulse together with loading rate primarily influences the degree and the severity of flexion-distraction injury to the lumbar spine.

3. The duration of the load pulse does not affect the load response of the FSU in these experiments. Injury always occurred 10-60 ms after that the load was applied.

4. The response of the lumbar FSU to a dynamic flexion-shear load is dependent on the bone mineral content in the vertebrae and the size of the segment.

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