DYNAMIC LOAD RESPONSE OF THE IN VITRO LUMBAR SPINE IN FLEXION

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ABSTRACT

The biomechanical response of the in vitro lumbar motion segment (functional spinal unit, FSU) under a dynamic (transient) flexion-shear load was determined. The load was transferred to the specimen by a padded pendulum and simulated a flexion-distraction injury, a so called lap seat-belt injury. The load response of the specimen was measured with a force and moment transducer, and the motions were determined with high speed photography. Two series of tests were made with 10 specimens in each, with two different load pulses: one moderate pulse (mean acceleration 2.5 g, duration 150 ms) and one severe pulse (mean acceleration 8 g, duration 250 ms). The results showed that the moderate load pulse caused initial flexion-distraction injuries at a mean bending moment of 113 Nm and a mean shear force of 346 N. The maximum flexion angulation attained during the loading sequence was 14°. The severe load pulse caused evident signs of failure or total rupture of the segments at a mean bending moment of 151 Nm and a mean shear force of 481 N. The flexion angulation just before failure was 19°. A statistically significant correlation ($r>0.7$, $p<0.05$) was found between the load response and the height of the segment, the load response and the lateral disc diameter, and the load response and the bone mineral content (BMC) in the vertebrae. Comparisons were made with previous established thresholds for static flexion-shear loading. The results indicated that thresholds for initial and ultimate flexion-distraction injury respectively are in the same range for static and transient loading conditions.

INTRODUCTION

The number of spinal injuries has increased over the past decades. Traffic accidents are the cause of more than 50% of the severe cases. Fractures and dislocations of the vertebrae as well as minor spinal injuries can cause both acute and chronic pain or disability. Knowledge of the biomechanical response of the spine under traumatic, static, and transient loads would lead to a better understanding of the injury mechanisms.

In car accidents, lap seat-belts can cause lower torso injuries with complete flexion fractures or flexion-distraction injuries as a result (Greenbaum et al., 1970; Dehner, 1971). This type of injury was very common during the 60´s and early 70´s, about 24% of all fatalities in motor vehicle accidents were flexion injuries (Nahum et al., 1968). A marked large decrease in the number of lumbar spine injuries in frontal car accidents occurred when the shoulder belt was introduced. However, Denis (1983) reported that flexion-distraction injuries still account for about 6 per cent of the severe injuries of the lumbar spine. King (1984) reported that flexion fractures attribute to the improper wearing of the lap belt while involved in a frontal collision. This type of injury may also occur to passengers on the non struck side in lateral oblique car impacts.

Numerous in vitro experimental studies of lumbar spine response to pure static loads in one direction, especially in the physiological range of motion have been reported (Adams et al., 1980 and Posner et al., 1982). The biomechanical response of an injured lumbar spine segment has been investigated by, for example, Lin et al. (1978) and Miller et al. (1986). However only a few studies have taken into account the increasing dislocation during the rapid injury phase, or the fact that a combination of bending moment, shear force and compressive force often occurs in real life (e.g. in traffic accidents).

In order to increase knowledge of spinal injury patterns under traumatic static flexion-shear loads causing flexion-distraction injuries (so-called lap seat-belt injuries) from physiological movements until complete rupture, different loading and measuring techniques have been developed and static and quasi-static in vitro studies have been conducted on the lumbar spine (Osvalder et al., 1987,1990; Neumann et al., 1991,1992).

Very few experiments of dynamic traumatic response of the lumbar spine have been reported so far. Cyclic dynamic compressive loading has been used to simulate repetitive stresses that cause fatigue failure during normal daily activities (Hansson et al., 1987) and has also been used to determine biomechanical properties of the intervertebral disc (Koeller et al., 1984). Willén et al. (1984) used an instant axial dynamic force in order to produce burst or crush fractures of the thoracolumbar spine. The strain distribution over the cortical surface in a spinal segment during whole-body (compressive) impact acceleration was studied by Hakim and King (1977). They concluded that it is erroneous to correlate the possible areas of vertebral fractures to findings from pure axial compressive tests. The applied bending moment plays an important role in the strain distribution.

Spinal response to combined dynamic (transient) loads as well as to different acceleration pulses has not been solved in the literature. The purpose of the present work has been to develop methods for dynamic (transient) loading of spinal segments in vitro and to use these methods to establish thresholds for initial and ultimate flexion-distraction injuries of the lumbar spine. The motion pattern and the load response under dynamic conditions will be discussed in relation to the lumbar spine response to static flexion-shear loads.

MATERIAL AND METHODS

The experiments were made as follows:

1. Static loading of intact specimens with a physiological flexion shear load of 25 Nm and 100 N, to precondition the specimens and to determine the flexion angulation

2. Dynamic flexion shear loading (with a moderate or severe load pulse), to cause initial or ultimate flexion-distraction injuries of the specimens

3. Static loading of injured specimens with a physiological flexion shear load of 25 Nm and 100 N, to determine the flexion angulation, i.e. permanent deformation after injury

Specimens

Lumbar specimens were obtained during routine autopsies. These were divided into appropriate Functional Spinal Units (FSU) comprising two adjacent vertebral bodies, the intervertebral disc and all intervening ligaments. The FSUs were screened by X-ray and only intact specimens were used in the experiments. The Bone Mineral Content (BMC), expressed in g/cm, in the two vertebrae was determined by dual photon absorptiometry (Roos and Sköldborn, 1974). The specimens were frozen and stored at -20° C until preparation and testing.

Onethird of each upper and lower vertebrae were mounted and fixed with Plastic Padding into specially designed steel cups. The posterior mid-sagittal part of the vertebral body was mounted at the centre of the steel cup. The lower end-plate of the disc was parallel with the lower steel cup. To enable rigid fixation, six prepared pointless metal screws (diameter 5mm) were inserted through the walls of each cup to reach the bone. A screw with a pointed end was inserted into each metal screw in order to reach partially into the vertebral bone. To prevent damage to the tissues due to the exothermic process in the setting cement the specimens remained frozen until fixation. During preparation and testing the specimens were kept moist by means of 0.9% saline solution to prevent dehydration and deterioration. The preparation methods are described in more detail by Osvalder et al. (1990).

A total of 20 FSUs from 12 spines (10 male, 2 female) were tested in two series with ten specimens in each. The specimens were selected in a random order and consisted of segments from L1-L2 and L3-L4. Data regarding age, lumbar level, specimen height, anterior/posterior disc diameter and disc thickness were collected for each FSU.

Experimental set-up for dynamic testing

The experimental set-up was designed to simulate a dynamic (transient) flexion-shear load that would cause flexion-distraction injuries to the lumbar spine as described by Denis (1983). A dynamic load of combined bending moment , shear force and compressive force was transferred to the FSU in the anterior direction by a padded pendulum (Figure 1). The pendulum (20 kg) hit an impactor situated on a metal lever about 300 mm above the mid-disc plane of the FSU. Anthropometric data has shown that the location of the centre of mass for the head and torso in sitting position is about 300 mm above the L3 vertebrae (Roebuck et al., 1975). Two 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 stopped when it had transferred enough energy to the impactor so as to pretension the ligaments and move forward the system above the joint (mid-disc plane).

The shape of the load pulse (magnitude, rise time, duration) transferred to the impactor was determined by the force-deformation characteristics of the padding material attached to the pendulum face (area 40×60 mm^2) and by the impact velocity of the pendulum. After preliminary tests, two different load pulses were chosen: one moderate (series 1) and one severe (series 2). The two pulses were chosen because they caused trauma close to the thresholds for initial (moderate pulse) and ultimate (severe pulse) flexion-distraction injuries of the lumbar FSU. Earlier studies have described the mechanisms of initial and ultimate flexion-distraction injuries under static flexion-shear loads (Osvalder et al., 1990; Neumann et al., 1992). Thus comparisons could be made between static and dynamic flexion-shear loading of the lumbar spine.

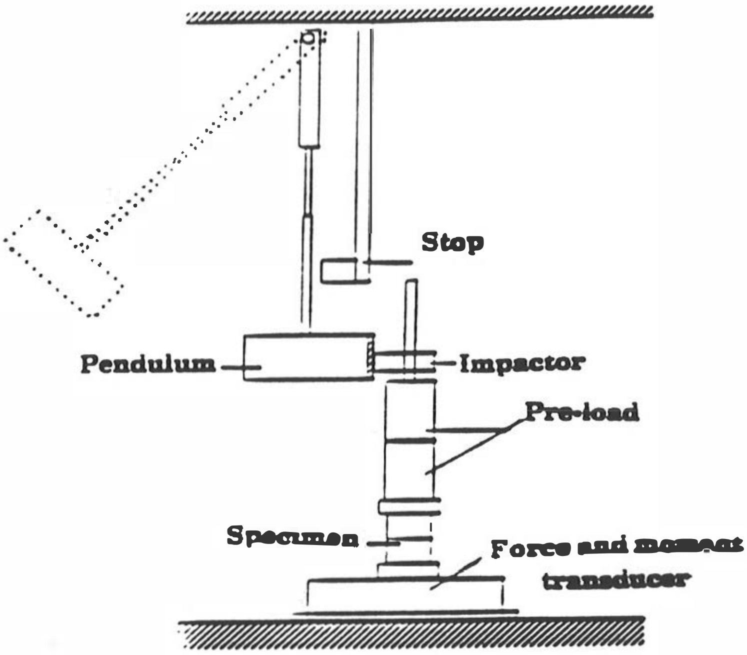

Figure 1 Experimental set-up for dynamic loading

The moderate load pulse was achieved by using 150 mm thick polyurethane padding (volume density 33 kg/m^3) on the pendulum face and an impact velocity of the pendulum of 1.6 m/s. The severe load pulse was achieved by using 40 mm thick Termolon 30 (polyethylene) on the pendulum and an impact velocity of 2.1 m/s.

The load response of the FSU was measured by means of a three dimensional force and moment transducer (AMTI MC 12-6-1000). The capacity of the transducer was 4 500 N for the force channels and 800 Nm for the moment channels. The frequency response was sufficient for dynamic measurements in the range 0-1500 Hz. The output signals from the transducer were amplified, stored, and processed by means of a microcomputer and an analysing programme (PIAS). The lower steel cup of the specimen was rigidly attached to the transducer, which was firmly attached to the ground. Thus the transducer was situated below the specimen. Therefore in order to describe the load response in the sagittal disc plane, the measured bending moment and the shear force had to be corrected. The corrected values were, however, only about 5 per cent lower than the measured values.

The acceleration pulse was measured with a one-dimensional accelerometer (Entran, range ±25 g) which was attached to the impactor. A photoelectric cell with infrared light was attached to the pendulum. The output signal from the cell was used to calculate the velocity of the pendulum just before impact as well as to trigger the force, moment and acceleration signals.

The motion behaviour of the FSU was monitored by high speed photography (500 frames per second). Small silver coloured balls (diameter 3 mm) were glued to nails (length 4 mm) which were partially inserted in the vertebral body above and under the intervertebral disc, and in the spinous processes. During the high-speed film evaluation, the position of the balls were determined by means of a digitiser and an analysing programme. Flexion angulations were calculated during the whole loading sequence and correlated to the load response.

Static physiological testing

A static flexion-shear load of 25 Nm and 100 N was applied to the specimens for four minutes before and after the dynamic loading sequence in order to compare the flexion angulation of an intact and injured specimen. The initial period of static loading also functioned so as to precondition the specimens. The static load was below the injury level for flexion distraction injuries and produced movements in the physiological range of motion (Schultz et al., 1979; Neumann et al., 1992). Two mechanical displacement gauges (accuracy ±0.01 mm) were used to measure vertical displacements of the specimens. From these measurements, flexion angulation was calculated. The experimental set-up for static flexion-shear loading is described in detail by Osvalder et al. (1990).

Definition of injury

The injury load reported was defined as the maximum bending moment and shear force that the structure could withstand before a reversal of load took place.

A detectable sign of initial injury was the obvious changes in the force and moment curves which could be linked with a change in motion pattern seen on the the high-speed film. A dip greater than 15 Nm and 50 N was the criterion for static initial flexion-shear injury (Neumann et al., 1992). If a permanent deformation was produced in the specimen during the moderate loading sequence, this was shown by increased angulation of the injured FSU compared to the intact FSU under the same static physiological flexion-shear load.

Ultimate injury occurred either when total rupture of the specimens occurred during the severe loading sequence or when evident signs of permanent deformations were shown afterwards under the static physiological flexion-shear loading (a flexion angulation >16° as described by Osvalder et al., 1990).

RESULTS

Data regarding age, dimensions and BMC of the specimens included in the two test series are summarised in Table 1. In series 1, five L1-L2 and five L3-L4 segments (from 8 different spines) were used, and in series 2, six L1-L2 and four L3-L4 (from 7 different spines) were used. Two homogeneous groups of specimens were used in the experiments. Mean(S.D.) and range for the biological parameters are presented in Table 1.

Table 1 Morphologic and demographic data of the 20 FSUs in the experiments

Series	Parameter	Mean(S.D.)	Range
	Age (Yrs)	49(9)	38-66
Series 1	Specimen height (mm)	76(7)	67-92
(N=10)	Disc thickness (mm)	13(3)	9-18
	Lateral disc diam. (mm)	57(5)	51-63
	A/P* disc diam. (mm)	40(4)	36-44
	BMC (g/cm)	3.9(0.8)	3.0-5.1
	Age (Yrs)	53(11)	38-74
Series 2	Specimen height (mm)	80(8)	68-94
(N=10)	Disc thickness (mm)	12(3)	8-16
	Lateral disc diam (mm)	59(5)	52-67
	A/P* disc diam (mm)	42(3)	39-46
	BMC (g/cm)	4.0(0.9)	2.3-5.4

* A/P =anterior posterior disc diameter

The peak acceleration value for the moderate load pulse was 5 g and the mean value approximately 2.5 g. The rise time was 30 ms and the duration 150 ms. For the severe load pulse, the peak value was 12 g, the mean value 8 g, the rise time 15 ms and the duration 250 ms. The differences between the ten load pulses in the same group were about 5 per cent.

The results from the two series are presented in Table 2. For the moderate load pulse (series 1) the mean peak value (duration >3 ms) was 113 Nm for the bending moment in flexion, and 346 N for the anterior shear force. These values represent the maximum load the specimen could withstand before a reversal of load took place and some kind of injury occurred. No specimen failed completely, but signs of injury occurred in all specimens at around 40 ms after the load was applied. This was shown by a distinct dip in the force and moment curves(>15 Nm and 50 N), which indicated that some structure had failed, but also that other structures took over its function during the remaining part of the load pulse. The mean maximum flexion angle attained was 14° and was reached at about 100 ms after impact, i.e. when the load began to decrease. Figure 2 presents a typical load-time curve for a specimen exposed to the moderate load pulse.

Table 2 Results from dynamic flexion shear loading of lumbar FSUs. Peak values >3 ms are reported.

Series	Parameter	Mean (S.D)	Range
Series 1 (N=10) Moderate pulse	Bending moment (Nm)	113(11)	97-129
	Shear force (N)	346(36)	306-420
	Max flexion angle (°)	14.1(1.4)	11.3-16.1
Series 2 (N=10) Severe pulse	Bending moment (Nm)	151(18)	128-173
	Shear force (N)	481(58)	419-568
	Max flexion angle (°)	19.3(1.5)	16.4-21.2

For the severe load pulse (series 2), overt fractures occurred in six of the ten specimens tested after 30-40 ms. The four remaining specimens showed a large dip in their moment and force curves (>75 Nm and 200 N) 60-70 ms after the load was applied, but managed to withstand the whole loading sequence (250 ms) without undergoing complete rupture. Afterwards these four specimens showed evident signs of failure (complete permanent deformations). The mean peak values (>3 ms) were 151 Nm for the bending moment and 481 N for the shear force. The mean maximum flexion angulation prior to failure was 19°. Figure 3 shows typical load-time curves for specimens exposed to the severe load pulse.

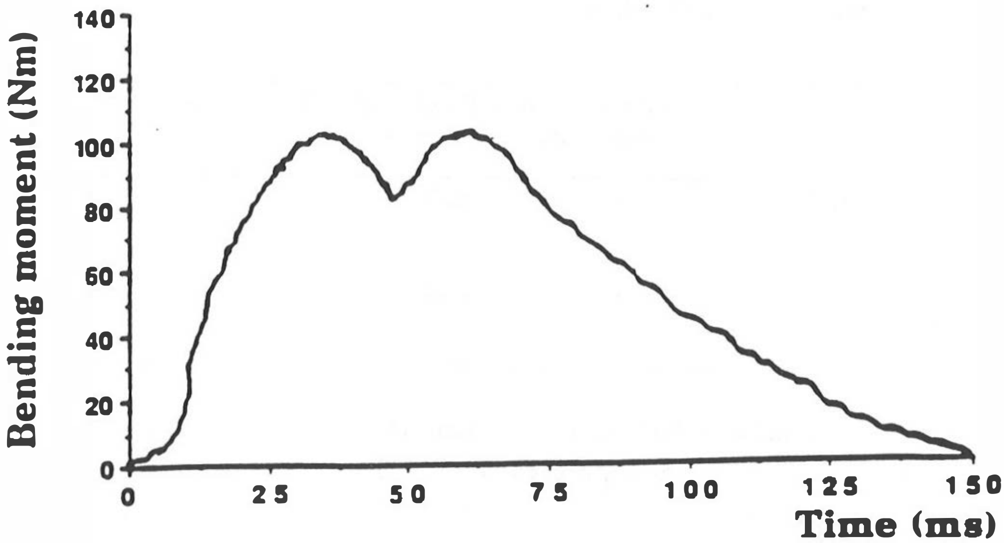

Figure 2 Bending moment versus time for a lumbar FSU exposed to the moderate dynamic load pulse.

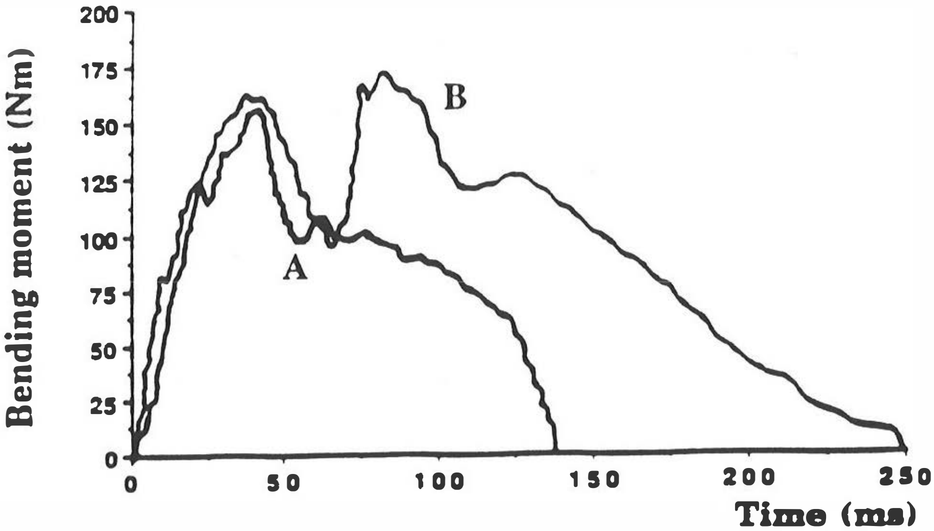

Figure 3 Bending moment versus time for typical lumbar FSUs exposed to the severe dynamic load pulse. A) overt fracture occurred after 40 ms B) injury occurred after 60 ms but the specimen managed to withstand the load during the whole sequence

The flexion angulation differed between intact and injured specimen when a static load of 25 Nm and 100 N was applied before and after the dynamic loading sequence (Table 3). For specimens exposed to the moderate load pulse, the difference in angulation between intact and injured FSU was 9°. The four specimens which were not totally ruptured during the severe load pulse showed a difference in angulation of 12°. These four specimens were able to resist the static physiological load without further deformations. These specimens had a high BMC-value (>4.0) and were among the largest specimens in size.

Table 3 Mean(S.D.) flexion angulation of intact and injured FSU under a static flexion-shear load of 100 N (25 Nm)

Series	Flexion angle(°) intact FSU	Flexion angle(°) injured FSU	Difference
Series 1 N=10	7 (2)	15(2)	8(2)
Series 2 N=4	8(2))	20(2)	12(3)

The coefficient of correlation between peak bending moment and different biological parameters related to the specimens are presented in Table 4. A coefficient of correlation r>0.7 was found between height of the FSU and the bending moment, as well as between lateral disc diameter and the bending moment. The BMC in the two adjacent vertebrae also appeared to be a good predictor of the injury load (r>0.7). The same connection was found for the shear force and the biological parameters.

Table 4 Coefficient of correlation between peak bending moment and different biological parameters related to the specimens ($p<0.05$).

Biological parameter	Coefficient of correlation (r) moderate load pulse	severe load pulse
Age	0.52	0.55
Specimen height	0.74	0.77
Disc thickness	0.55	0.51
Lateral disc diam.	0.72	0.74
A/P disc diam.	0.56	0.58
BMC	0.71	0.74

DISCUSSION

The purpose of this study was to determine the ability of the lumbar spine to resist large traumatic dynamic flexion-shear loads. An experimental set-up was designed so that a combination of bending moment, shear force and axial compression on lumbar FSUs could be applied. Thresholds for dynamic initial, and ultimate flexion-distraction injuries were determined and compared with thresholds for static flexion-distraction injuries. This was an attempt to gain further insight into spinal injury patterns and mechanisms under static and dynamic loading conditions. This information can be useful for the validation of mechanical and mathematical models of the lumbar spine as well as for the development of preventive measures that will reduce the risk of spinal injuries, particularly those caused by traffic accidents.

The experimental set-up was designed as realistically as possible to simulate a dynamic flexion-distraction type of injury of the lumbar spine, as described by Denis (1983). In this loading situation, seen, for example, in a frontal car accident, a transverse load is applied to the trunk 200-300 mm above the L1-L4 vertebrae (Roebuck et al., 1975). The lumbar spine is thereby simultaneously subjected to a combination of anterior shear force, forward bending moment and axial compression (Figure 4).

The moderate and severe load pulses were chosen to cause initial and ultimate flexion-distraction injuries of the lumbar FSUs. Rise time and duration were also chosen to imitate the load distribution of lumbar spine segments in a car collision (Cheng et al., 1979). All ten specimens exposed to the moderate load pulse showed signs of permanent deformations after the dynamic loading sequence. The severe load pulse caused total rupture or large permanent deformations of the specimens.

An experimental set-up for traumatic static flexion-shear loading was designed by Osvalder et al. (1990). The static load was applied 250±10 mm above the mid-disc plane resulting in an anterior shear force (F_S) and a bending moment in flexion (M_b) acting on the FSU. For comparisons between thresholds for static and dynamic flexion-distraction injuries it was desirable to achieve the same relationship between bending moment and shear force in the dynamic experimental set-up. After several tests with different configurations and positions of the impactor and pre-loads on the metal lever (Figure 1), the best solution was found by positioning the impactor 300 mm above the mid-disc plane.

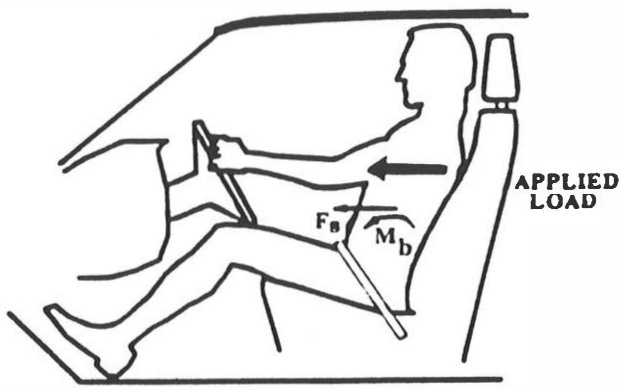

Figure 4 Loading situation for simulated flexion-shear of the lumbar spine.The lumbar spine is subjected to a combined load of anterior shear force (F_S), bending moment (M_B) and axial compression.

In the dynamic experimental set-up a compressive pre-load of 12 kg was used. Twelve kilos is a relatively low pre-load but at higher pre-loads the behaviour of the specimen tends to become unstable in the absence of muscular forces. It is very important to avoid the effect of unstable motion during mechanical testing (e.g. Goel et al., 1987). Several experiments under static loading conditions within the physiological range have been done with, and without, pre-load (e.g. Panjabi et al., 1977). To summarise these studies, the application of a pre-load increases disc compression and relaxes the ligaments. The influence of a pre-load on static traumatic loadings has not been fully investigated nor has its influence on dynamic loading.

In Table 5, thresholds for initial and ultimate flexion-distraction injuries under static (Osvalder et al, 1990; Neumann et al., 1992) and dynamic loading conditions are presented. A comparison shows that the loads causing injury may be in the same range. Because exactly the same relation between bending moment, shear force and compressive force was not achieved in the two experimental set-up, it is not quite correct to compare the magnitude of the bending moment and the shear force at the injury level for static and dynamic loadings. Furthermore, the rise-time and duration of the load pulse are parameters which may affect the load response of a specimen. These parameters should be further investigated by, for example, exposing lumbar FSUs to additional load pulses. By changing the padding material on the pendulum face and the impact velocity of the pendulum, magnitude, rise time and duration of the load-pulse are changed.

As pointed out by Hakim and King (1977) the applied bending moment plays an important role for vertebral fractures. As can be seen in Table 5, the bending moment and the flexion angulation are in the same range for static and dynamic loading conditions, both for initial and ultimate injuries. No statistically significant differences were shown on the 5% level for these parameters. For the shear force, there was a difference between static and dynamic loading conditions, which mainly depends on the difference in the experimental set-ups. After unloading, the specimens showed the same injury pattern, i.e. overt fractures in the vertebrae. This implies that the bending moment (which causes the angulation of the specimen) plays a major role in the injury mechanisms of ultimate flexion-distraction injuries of the lumbar spine, both under static and dynamic conditions. A flexion angulation of 20°, as attained in the experiments, is in clinical practice an indication of an unstable injury.

Table 5 Results from static (Neumann et al., 1992; Osvalder et al., 1990) and dynamic flexion-shear loading of lumbar FSUs. Mean(S.D.) for each parameter.

Parameter	Initial injury		Ultimate injury	
	Static	Dynamic	Static	Dynamic
Bending moment (Nm)	121(10)	113(11)	156(11)	151(18)
Shear force (N)	486(38)	346(36)	620(23)	481(58)
Flexion angulation (°)	16(1)	14(1)	20(2)	19(2)

Age and bone strength are usually parameters which correlate well. In this study the age range was wide (38-74 years), because of the obvious difficulty in collecting an appropriate number of specimens from a younger population. However, the correlation between age and load at failure was not significant in the two test series, nor was it in the static flexion-shear tests (Neumann et al., 1992).

Both the size of the FSU (i.e. lateral disc diameter and height of the segment respectively) as well as BMC in the vertebrae showed a statistically significant correlation ($p<0.05$) with the dynamic load (Table 4). The specimen exposed to the severe load pulse showed a higher correlation with BMC than did the specimens exposed to the moderate load pulse. Neumann et al. (1992) showed that the only parameter that could be connected with the injury load under static flexion-shear loading was the BMC. The static tests also showed that the coefficient of correlation was higher between BMC and the load that caused ultimate injuries, than between BMC and the load that caused initial injuries. The range for the BMC-values in the present study and in the study by Neumann et al. (1992) was 2.3-5.4 g/cm and varied within the individual spines in a random pattern independently of spinal level. These are representative values for in vivo measurements.

It is difficult to compare the present results with other dynamic experimental studies of the lumbar spine because of different experimental set-ups and loading directions. A lot of studies have been reported regarding dynamic loading of whole body cadaver spines (e.g. Mital et al., 1978; Prasad et al., 1974), animal spines (e.g. Kazarien et al., 1971) and living human volunteers (e.g. Cheng et al., 1979). Many of these experiments were made to establish tolerance levels for different parts of the body during crash tests. As far as we know, no experimental study on lumbar FSUs has been reported regarding dynamic flexion-shear loadings that can cause injuries.

CONCLUSIONS

1. The thresholds for flexion-distraction injuries may be within the same range for static and dynamic loading conditions.

2. A flexion angulation of 20° seems to be the limit for motion under a dynamic flexion-shear load.

3. The resistance to dynamic flexion-shear load is dependent on BMC in the vertebrae, the height of the FSU and its lateral disc diameter.

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