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Posture and the compressive strength of the lumbar spine

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Summary

The effect of posture on spinal compressive strength was examined in a series of three experiments on cadaveric material. Lumbar 'motion segments', consisting of two vertebrae and the intervening disc and ligaments, were compressed while positioned in various angles of flexion and extension. In the first experiment load sharing between the disc, the apophyseal joint surfaces, and the intervertebral ligaments was inferred from measurements of intradiscal pressure (IDP). Results showed that extension caused the apophyseal joints to become load-bearing, and damage could occur at compressive loads as low as 500 N. Flexion angles greater than about 75% of the full range of flexion (as defined by the posterior ligaments) generated high tensile forces in these ligaments, and caused substantial increases in IDP. The optimum range for resisting compression therefore appeared to be 0–75% flexion. The second experiment compared the distribution of compressive stress within the disc at the endpoints of this range, and showed that at 0% flexion high stress concentrations occur in the posterior annulus of many discs, whereas an even distribution of stress was usually found at 75% flexion. However, the third experiment showed that there was no significant difference in the compressive strength of motion segments positioned in 0% and 75% flexion. A comparison of the range of flexion/extension movements *in vivo* and *in vitro* led us to conclude that in life a position of moderate flexion is to be preferred when the lumbar spine is subjected to high compressive forces.

Relevance

The experiment suggests that the normal lumbar lordosis should be flattened during manual handling to avoid injury to the osteoligamentous lumbar spine.

Key words: Posture, lumbar spine, mechanical loading, injury, intervertebral disc

Introduction

The close association between lumbar disc prolapse and the lifting of heavy weights[1] shows how important it is to lift in a correct manner. The usual advice is to 'bend the knees' and 'keep the back straight'. Bending at the knees can help to reduce the distance between the body and the weight to be lifted, and this in turn reduces the compressive force acting on the spine, where the term 'compressive' refers to the force acting perpendicular to the mid-plane of the disc. The second recommendation is rather confusing because it is intended to mean 'preserve the natural lumbar lordosis' as opposed to flexing (flattening) the lumbar spine[2]. It is not necessary to preserve a lordosis in order to minimize the compressive force on the spine, and indeed the benefits of doing so are not self-evident.

Measurements of intradiscal pressure in various sitting postures suggest that a lumbar lordosis can reduce the hydrostatic pressure in the nucleus pulposus[3]. Compressive failure of a lumbar motion segment occurs in the vertebral body end-plate[4–6] presumably as a result of high pressure in the nucleus pulposus, and so it is generally assumed that a lordosis can protect the spine against compressive injury. However, the protective action of a lumbar lordosis has never been demonstrated at the high load levels encountered during heavy labour, and there are

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reasons to suppose that it might not then operate[7]. Also, the possible benefit of a lordosis must be offset against the known disadvantages of increased apophyseal joint loading[8–10] and increased compressive loading of the posterior annulus compared to the anterior annulus[11,12].

On the other hand, lifting heavy weights with a flexed lumbar spine may also be hazardous. Excessive flexion can sprain the ligaments of the neural arch[13] and also stretch the posterior annulus to the extent that the disc becomes vulnerable to posterior prolapse[14–17].

In actual practice, most people flex the lumbar spine when lifting objects from the ground, even when they bend their knees[18–21]. Natural habits can sometimes be improved upon, but it would be unwise to advocate an unnatural lordotic lifting technique unless its presumed advantages can be demonstrated clearly. The purpose of the present experiment was to compare the effect of lordotic and flexed postures on the ability of the osteoligamentous lumbar spine to resist high compressive forces. In this way we hoped to establish the range of flexion and extension in which the lumbar spine has its optimum resistance to compression.

Methods

Cadaveric material

Nineteen lumbar spines aged between 19 and 74 years were collected at routine necropsy from subjects who had no history of spinal injury or prolonged bed-rest. Spines were stored in sealed plastic bags at −17°C for up to 3 months before use. They were then thawed at 3°C for 12 h, and dissected into 29 'motion segments' consisting of two vertebrae and the intervening disc and ligaments (details in Table 1). If two motion segments were obtained from the same lumbar spine, one would be tested immediately, and the other vacuum-sealed in a plastic bag, stored overnight at 3°C, and tested on the following day. During the testing procedures specimens were covered in thin polythene film to minimize water loss.

Mechanical testing

Each motion segment was secured in two cups of mildly exothermic dental stone. Screws and hooks inserted into the spinous processes and into the superior and inferior articular processes ensured that no movement could occur between bone and stone. A computer-controlled hydraulic materials testing machine (Dartec Ltd, Stourbridge, UK) was used to apply compressive, bending and shear loads to the specimen, as shown in Figure 1. A combination of compression, bending, and shear can be used to simulate the vector sum of all gravitational and muscle forces acting on the spine in the sagittal plane *in vivo*[13]. The height of the rear roller shown in Figure 1 was adjustable, so that the specimen

Table 1. Details of the 29 motion segments used in these experiments. Disc degeneration was scored on a scale of 1 (no degeneration) to 4 (gross degeneration). Specimens 1–14 were used in the matched-pair study

	Specimen details				Limit of flexion		Limit of extension
Number	Sex, age	Level	Disc degeneration	Creep height loss (mm)	BM (Nm)	Angle (°)	Angle (°)
1	M, 74	L_{4-5}	4	1.1	45.7	11.6	
2	M, 74	L_{2-3}	4	2.0	—	6.0	
3	M, 64	L_{3-4}	3	1.2	67.4	12.1	
4	M, 64	L_{1-2}	3	0.9	43.6	6.5	
5	M, 53	L_{4-5}	3	0.9	—	15.0	
6	M, 53	L_{2-3}	3	1.0	36.4	11.0	
7	M, 30	L_{2-3}	2	0.9	65.6	11.2	
8	M, 30	L_{4-5}	2	1.3	—	17.0	
9	F, 67	L_{2-3}	3	1.3	36.6	9.2	
10	F, 67	L_{4-5}	3	2.0	41.9	10.5	
11	F, 38	L_{2-3}	2	1.0	44.4	10.7	
12	F, 38	L_{4-5}	2	1.2	48.7	14.2	
13	F, 42	L_{4-5}	2	1.1	—	17.0	
14	F, 42	L_{2-3}	2	1.0	52.4	10.4	
15	M, 36	L_{2-3}	2	0.9	77.1	12.0	−8
16	M, 52	L_{2-3}	2	1.0	59.5	9.8	−3
17	M, 52	L_{4-5}	2	1.0	51.9	12.5	−4
18	M, 69	L_{2-3}	3	—	59.8	11.2	
19	M, 19	L_{2-3}	1	1.0	59.1	14.2	
20	M, 40	L_{4-5}	2	1.0	69.8	18.1	−4
21	M, 40	L_{2-3}	2	0.9	124.3	15.1	−6
22	M, 49	L_{2-3}	2	0.8	93.4	14.3	−4
23	M, 49	L_{4-5}	2	0.8	66.4	12.9	−4
24	M, 52	L_{2-3}	2	1.1	51.2	12.0	−6
25	M, 19	L_{2-3}	1	1.4	68.9	14.4	−4
26	F, 46	L_{2-3}	2	1.5	67.4	10.9	−5
27	M, 27	L_{1-2}	4	1.4	44.0	8.9	−4
28	F, 31	L_{4-5}	2	1.6	74.0	20.9	−3
29	M, 34	L_{4-5}	2	0.9	81.3	19.6	−4
Mean					61.2	12.7	
STD					19.6	3.5	

could be compressed while wedged at any angle of flexion or extension.

All motion segments were then creep loaded in order to reduce the water content of the intervertebral disc and bring it to some point in the presumed physiological range[22,24]. A compressive force of 1800 N was applied for 2 h, with the specimen held in 0° of flexion (i.e. in the neutral position for an unloaded motion segment). A graph of specimen height loss against time was plotted so that any damage to the specimen would be revealed by a discontinuity in the graph.

The full range of flexion and extension movement for each motion segment was then established, as follows. The rear roller shown in Figure 1 was removed and a combination of bending and compression applied to the specimen in 2.0 s loading/unloading cycles. The relative proportions of compression and bending were determined by the position of the front roller, which was usually 30 mm anterior to the geometric centre of the disc at the start of loading. The following were sampled at 250 Hz and stored on a microcomputer: vertical compressive force acting on the 20 kN load cell; vertical movement of the ram, measured by a linear variable displacement transducer (LVDT) mounted permanently on the ram; and specimen flexion angle, measured using an electrogoniometer mounted on the axis of the front roller. The bending moment acting about the geometrical centre of the disc was calculated from the force and flexion values, as described previously[18]. Loading cycles were repeated at higher and higher loads while graphs of bending moment against flexion angle were plotted. Eventually, the loading curve would show a distinct reduction in gradient at high load, indicating that the elastic limit had just been exceeded. This was confirmed by repeating the loading cycle and noting a small residual deformation and increased hysteresis in the second curve. The flexion angle at the elastic limit was taken to represent the 'range of flexion' of that motion segment. A similar technique was used on some specimens to establish the range of extension movement.

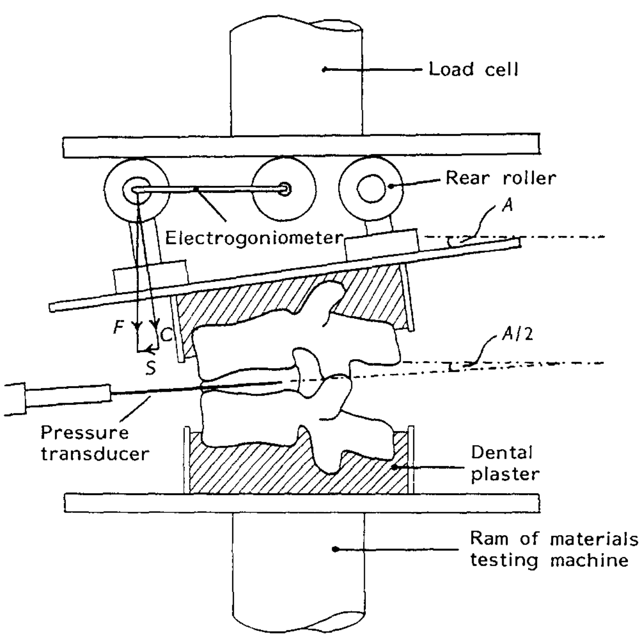

Figure 1. Apparatus used to apply bending and compression to the specimens. The height of the rear roller is adjustable, allowing the specimen to be flexed to various angles of flexion and extension. The low-friction bearings of the rollers transmit only a vertical force F between load cell and specimen. The compressive force C is defined as $C = F \times \cos(A/2)$ where A = flexion angle.

Experiment 1: Posture and load sharing between disc, ligaments, and apophyseal joints

Exhaustive tests were performed on nine motion segments (numbered 15, and 20–27 in Table 1). Each specimen was positioned in 0° of flexion, and a compressive force of 3000 N or 4000 N applied in a 4.0 s loading/unloading cycle. A pressure transducer mounted in the side of a 1.3 mm diameter needle[12] was used to measure the horizontal component of compressive stress acting in the centre of the nucleus pulposus. It is well established that this stress is isotropic, and so the term 'intradiscal pressure' or IDP will be used for these measurements. IDP was sampled at 125 Hz during each loading cycle, and a graph was plotted of IDP against applied compressive force. This procedure was repeated at 2° intervals of flexion and extension, starting in full flexion and ending in full extension. The whole series of IDP/force graphs was then repeated after the supraspinous, interspinous, and capsular ligaments, and the ligamentum flavum had been sectioned with a scalpel, and changes in IDP were attributed to tensile forces acting in the ligaments before they were cut. A third series of graphs was obtained after the apophyseal joints had been removed by means of two horizontal saw-cuts, and changes in IDP attributed to compressive forces acting on the facet surfaces before they were cut.

Some technical failures with the pressure transducer led to some incomplete data sets in this experiment.

Experiment 2: Posture and the distribution of stress in the intervertebral disc

The purpose of Experiment 1 was to establish the range of flexion and extension within which the stresses acting in the apophyseal joints and the intervertebral ligaments were low. The results (described below) suggested that this range was from 0 to 75% of flexion, where '0% flexion' represents an unloaded motion segment, and '100% flexion' is defined by the elastic limit of the intervertebral ligaments. Experiment 2 then compared the distribution of compressive stress within the disc at the end-points of this range, using the technique of 'stress profilometry'[12].

Briefly, each motion segment was positioned in 0% or 75% flexion and subjected to a compressive force of 500 N for a period of 20 s. During this time, a 1.3-mm diameter needle containing a side-mounted pressure transducer was pulled through the disc along its sagittal midline, from the posterior to anterior margins, and the component of compressive stress acting perpendicular to the transducer was sampled at a rate of 25 Hz. The needle could be rotated about its long axis so that vertical and horizontal stress could be measured in successive tests. The stress data were analysed digitally and the following parameters calculated for each pair of horizontal and vertical profiles: mean pressure in the nucleus pulposus (IDP), and peak horizontal or vertical stress in the anterior annulus and posterior annulus.

'Matched pairs' of motion segments from seven spines were used for these tests. Stress profiles were obtained, as described above, with each motion segment wedged in 0% and then 75% flexion, and with compressive loads of 500 N and 2000 N. Additional profiles were obtained from eight other motion segments at various flexion and extension angles spanning the whole range of movement.

Experiment 3: Posture and compressive strength

The same seven matched pairs of motion segments used in Experiment 2 were used in this experiment. One motion segment from each spine was compressed to failure at 0% flexion, and the other at 75% flexion, using the apparatus shown in Figure 1. The compressive force was increased at a rate of 2.5 mm/s until failure was marked by the gradient of the force–deformation curve falling to zero (or decreasing markedly if this did not occur). In order to avoid bias in the results, half of the specimens tested at 0% flexion were higher in the lumbar spine than the specimen tested at 75%. Similarly, half of them were tested on the second day. Values of compressive strength were corrected to account for the average increase of 13% between adjacent motion segments in the lumbar spine (calculated from the data of Brinckmann[4]). This ensured that the strengths of 'flexed' and 'lordotic' specimens would be comparable. After testing, the disc was excised from the motion segment. Its cross-sectional area was measured, and its level of degeneration scored according to the criteria described by Galante[23]. The vertebral endplates were photographed.

Statistical analysis

Paired *t*-tests were used to compare differences between the specimens tested at 0% and 75% flexion. Elsewhere, independent *t*-tests were used. The level of significance was set at 5%.

Results

After 2 h of creep loading, most specimens were continuing to lose height, but at a reduced rate, and none showed any sign of having been damaged during the test. The height lost by each specimen is shown in Table 1.

The range of flexion of each motion segment tested is shown in Table 1, together with the bending moment required to reach this limit. Some missing values were caused by instrument failures.

Experiment 1: Posture and load sharing between disc, ligaments, and apophyseal joints

Typical results are shown in Figures 2 and 3 and a summary is given in Table 2. Figure 2 shows that the IDP in an intact motion segment was constant for much of the range of motion, but increased as the limit of flexion was approached, and decreased in extension. On flexion, the increase can mostly be attributed to tension acting in the stretched posterior ligaments, because the increase was slight after these ligaments had been cut (Figure 2). Similarly, the reduced IDP in extension was caused primarily by the bony surfaces of the apophyseal joints resisting a proportion of the applied compressive force, since the reduction in IDP was not seen after these joints had been removed (Figure 3). Similar arguments can be used to show that the posterior ligaments resisted a proportion of the compressive force acting on the motion segment in extension (Figure 2) and that the facet surfaces played no part in resisting compression in the flexed posture (Figure 3). This latter figure actually shows a decrease in IDP in flexion after removal of the facet surfaces: this may be because not all of the ligaments were completely sectioned in the preceding test.

A similar pattern of results was obtained for all nine motion segments tested, but differences in the range of flexion between specimens meant that average values could only be computed for '100% flexion', '75% flexion' and '50% flexion' rather than for specific angles (Table 2). The range of extension was less variable so the angles −2° and −4° were retained. The average values show that the dependence of IDP on flexion and extension angle becomes less marked as the compressive force increases from 500 N to 3000 N. At high loads the IDP is substantially increased at 100% flexion but not at 75% flexion. The lower three rows of Table 2 refer to isolated discs, without any neural arch, and they show that when the compressive force is low, IDP is considerably increased in full flexion, presumably because the stretched posterior annulus prestresses the

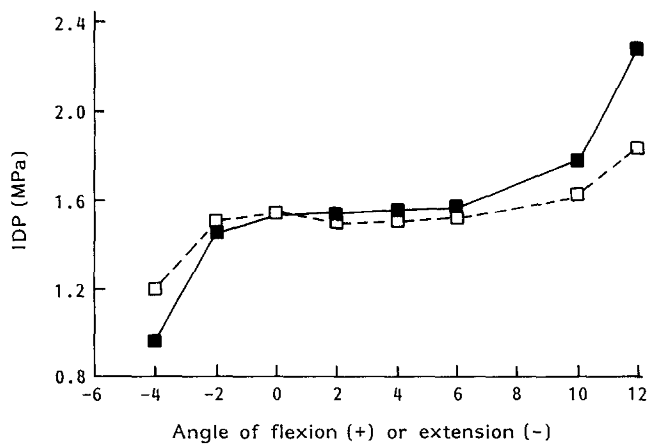

Figure 2. Effect of flexion and extension on IDP for a typical specimen (male aged 49, level L_{2-3}). The effect of ligament tension on IDP is revealed by the difference between the two graphs. Compressive force = 2000 N. ■, Intact; □, ligaments cut.

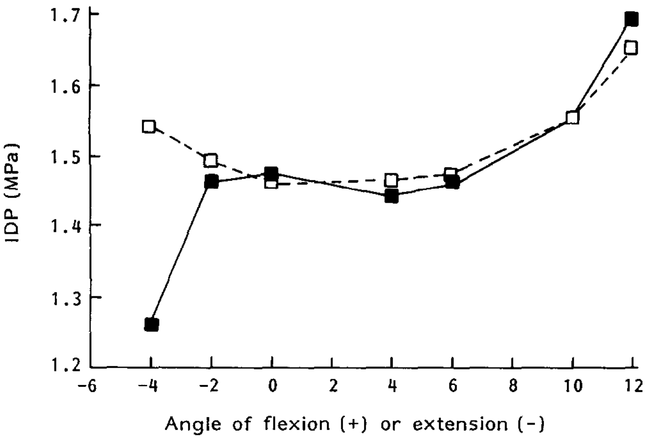

Figure 3. Effect of flexion and extension on IDP for a typical specimen (M, 49; L_{4-5}). The effect of the facet surfaces on IDP is revealed by the difference between the two graphs. Compressive force = 2000 N. ■, Ligaments cut; □, facets cut.

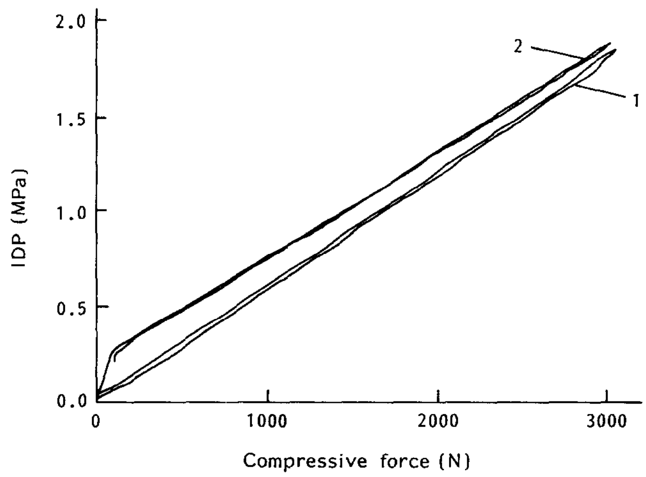

Figure 4. Linear relationship between compressive force and IDP is shown for a motion segment in 0° (curve 1) and 12° of flexion (curve 2) (M, 40; L_{4-5}). The increase in IDP associated with flexion becomes less marked at high loads.

nucleus. At high loads, however, the IDP is practically independent of flexion or extension angle.

A more complete picture of the relationship between IDP and applied compressive force is shown by Figure 4. At low loads ligament tension ensures that the IDP is higher in flexion than in the neutral position (0% flexion) but the difference is much reduced at higher loads. Graphs of IDP against applied compressive force yielded an unexpected result: in 2° and 4° of extension the graphs were non-linear and irregular when the load was increasing, but linear when the load was decreasing (Figure 5). Subsequent graphs followed the unloading part of the curve. Evidently, during the application of the compressive force, the resistance of some structure became impaired, so that an increased load was thrown on to the intervertebral disc. Hence during the unloading phase the IDP was increased for the same value of compressive force. This phenomenon was observed in all seven of the motion segments for which full data were obtained. The damaged structure could not be identified, but the scale of the injury could be estimated from the reduction in compressive force associated with the same IDP value before and after injury (Figure 5). It may be assumed that this reduction reflects the impairment in the damaged structure's resistance to compression at 2° or 4° of extension.

Experiment 2: Posture and the distribution of stress in the intervertebral disc

Stress profiles showed consistent differences between 0% and 75% flexion. With a compressive force of 500 N, stress peaks often occurred in the posterior annulus at 0% of flexion, while at 75% flexion the profiles were generally flat but with an elevated stress in all regions. At 2000 N and 0% flexion, stress peaks greater than 10% of the IDP were observed in the posterior annulus in 12 of 16 discs for which profiles were obtained at this angle. At 75% flexion the posterior stress peaks were removed or greatly reduced in 9 of 12 discs (Figures 6 and 7) and high peaks were observed in the anterior annulus in 3 of 12 discs.

Averaged results are summarized in Figure 8. With a compressive force of 500 N, 75% flexion significantly increased the pressure in the nucleus, and the peak stress in the anterior annulus. At 2000 N, however, 75% flexion caused no significant increase in nuclear

Table 2. The effect of flexion and extension on the pressure in the nucleus pulposus (IDP). Changes in pressure are expressed as a % increase (+) or decrease (−) of the pressure recorded in the neutral position (0°) for the same compressive force. Values are the mean for observations on nine motion segments (occasionally, n = 5–8). Numbers in brackets are the standard error of the mean

	Per cent increase (+) or decrease (−) in IDP					
Compressive force (N)	*4° Extension*	*2° Extension*	*0°*	*50% Flexion*	*75% Flexion*	*100% Flexion*
Intact motion segment						
500	−40 (5)	−10 (6)	0	+12 (6)	+45 (8)	+110 (17)
1000	−30 (7)	−6 (5)	0	+4 (1)	+23 (3)	+79 (10)
3000	−15 (4)	−4 (2)	0	+1 (1)	+6 (1)	+30 (6)
Disc–vertebral body unit						
500	+8 (5)	+3 (2)	0	+3 (3)	+12 (4)	+38 (9)
1000	+3 (3)	+4 (2)	0	0 (2)	+5 (3)	+12 (8)
3000	−0 (2)	+2 (1)	0	−1 (1)	+1 (2)	+4 (3)

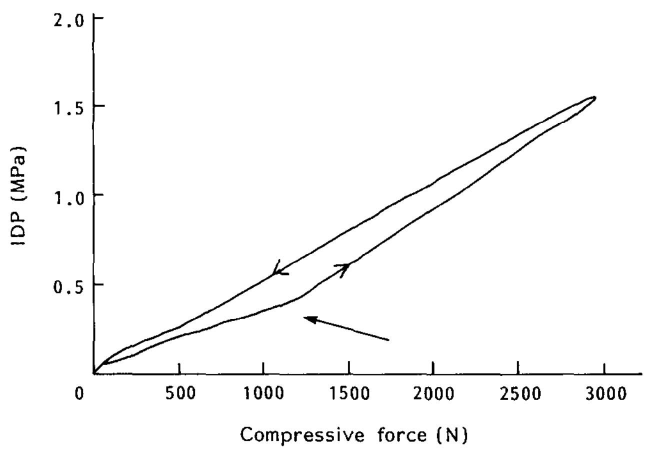

Figure 5. When a motion segment is compressed while held in 4° of extension, damage occurs at relatively low loads (large arrow) and causes an abrupt change in the IDP/force relationship. The small arrow heads distinguish between the loading and unloading curves (M, 49; L_{2-3}).

pressure, and reduced slightly the peak stress in the posterior annulus.

Additional stress profiles were recorded at angles between 2° and 5° of extension in eight motion segments. In five of these eight, peaks of compressive stress greater than 25% of the IDP were found in the posterior annulus (Figure 9). In all five cases, the peak disappeared or was greatly reduced when the specimen was loaded in flexion. Peaks of equivalent size were seen only once in the anterior annulus, and that was in a specimen loaded in full (100%) flexion.

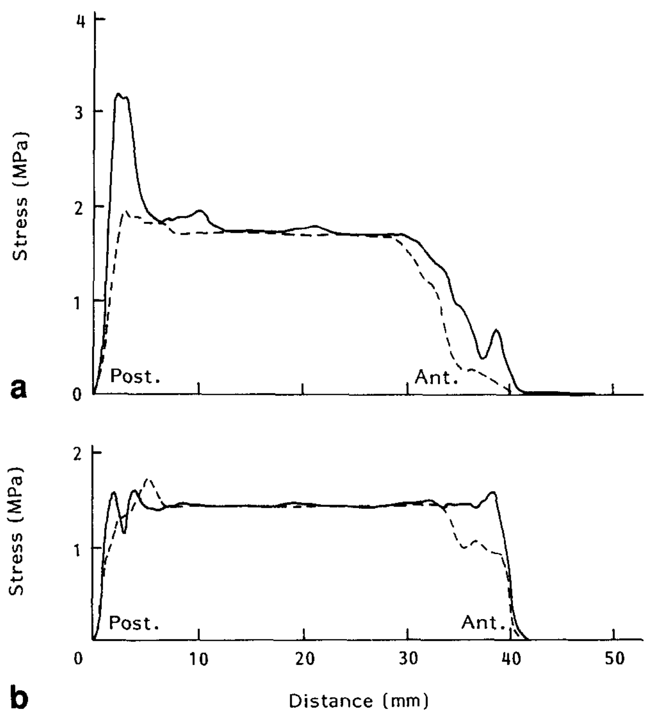

Figure 6. Stress profiles for a motion segment held in **a,** 0° of flexion and **b,** 4° of flexion. In the central region of the disc the vertical and horizontal components of stress are the same, indicating that the nucleus behaves as a fluid. Flexion removes the stress peak in the posterior annulus and, in this case, reduces nuclear pressure also. 2000 N; —— vertical; ------ horizontal. (Specimen: F, 67; L_{4-5}.)

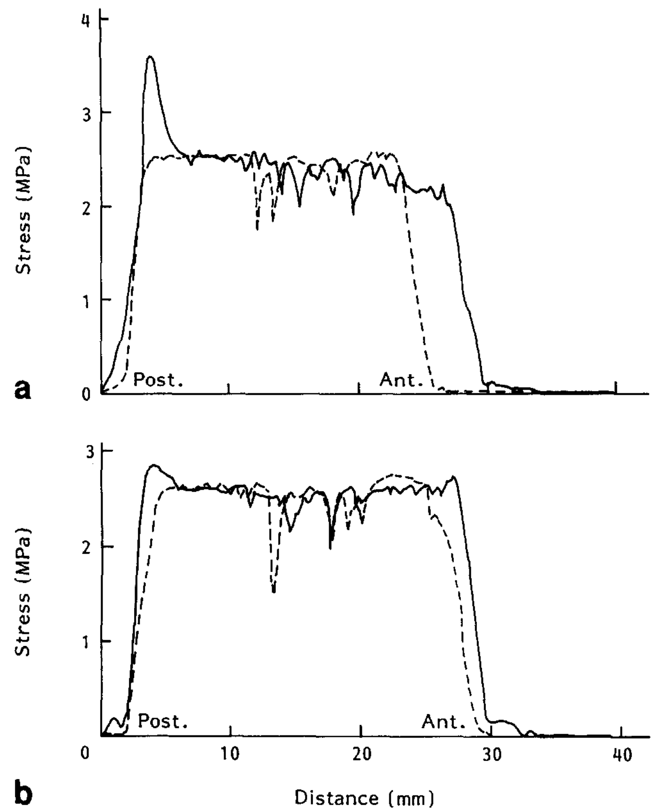

Figure 7. Stress profiles for a motion segment held in **a,** 0° of flexion and **b,** 7° of flexion. Note that practically no compressive stress is measured in the outermost few millimetres of the annulus, where the water content is low and the collagen content high. 2000 N; —— vertical; ------ horizontal. (Specimen: F, 42; L_{2-3}.)

Experiment 3: Posture and compressive strength

The results are presented in Table 3. Two values of 'compressive strength' were recorded: the force at which the stiffness (gradient) first decreased by more than 10%, and the ultimate compressive strength. The first point represents the threshold of damage to the specimen. Flexion angle had no significant effect on strength, regardless of which criterion was used, and this was also true if no correction was made for changes in strength at different lumbar levels. On average, lordotic specimens were 2.9% stronger (95% confidence interval: 8.9% stronger to 3.0% weaker) so it is unlikely that the two postures affect compressive strength by more than a few per cent, if at all.

The photographs of the vertebral end-plates showed that in 5 of 9 cases, flexion resulted in an end-plate fracture close to the anterior margin of the vertebral body. In all other motion segments, including those tested at 0% flexion, the fracture was approximately central.

Discussion

The validity of cadaveric experiments needs to be considered, because recent animal experiments have suggested that death alters the spine's time-dependent mechanical properties[25]. We suggest that these results are unreliable in view of the poor repeatability of the measurements[26] but we do accept that the water

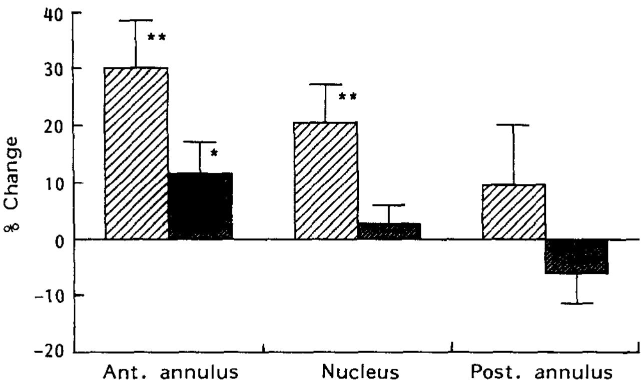

Figure 8. Postural changes in mean nuclear pressure and peak annulus compressive stress: values measured in 75% flexion are compared with those in 0% flexion. Bars indicate SEM. Significant changes denoted: $*P<0.05$, $**P<0.01$. $n = 12$; ▨, 500 N; ▩, 2000 N.

content of intervertebral discs may change post-mortem as a result of the prolonged absence of muscle forces[24]. However, the water content of intervertebral discs changes during the course of each day[22,27] and doubtless explains much of the diurnal variation in human stature[28–30]. It is not necessary, therefore, for cadaver experiments to reproduce any precise disc hydration; it is sufficient to ensure that the disc hydration falls somewhere within the physiological range. This was the purpose of the preliminary creep tests on each motion segment. The applied creep load (1800 N) corresponds to light manual labour[31] and was used to obtain, in a fairly short time, a height loss that corresponds to about half of the inferred diurnal loss of 1.5–2.0 mm from each lumbar disc[22]. We have previously shown that changes in spinal mechanics following creep loading depend on height loss rather than the applied load[22]. We suggest therefore that disc hydration in these experiments corresponds roughly to that found *in vivo* during the afternoon.

Other factors that may affect the mechanical properties of cadaveric spines include freezing, tissue degradation, and ambient temperature, but their effects have been shown to be slight[32–34]. The 'matched pair' design of the present experiment was intended to minimize the effect of any post-mortem artefacts on the results presented.

Several of our techniques have been validated in previous work. The use of repeated bending stiffness curves to establish the range of flexion of each motion segment is based upon experiments showing that the interspinous and supraspinous ligaments are the only structures to sustain damage just beyond this range[13]. Rapid loading cycles ensure that the very first signs of inelastic deformation can be detected without causing substantial damage to the specimen and so subsequent tests on that specimen are not compromised. A motion segment's resistance to bending depends on the compressive preload[18,35] and this is why the apparatus in Figure 1 was designed to apply physiologically reasonable combinations of bending and compression, rather than just bending. The pressure transducer used in these experiments does not significantly perturb the tissues under investigation, and gives stress values which are reproducible to better than 5%[12]. In the compressive strength tests the applied compressive force C acting perpendicular to the mid-plane of the disc would be accompanied by a forward shear force S equal to $C \times \tan(A/2)$ where A is the flexion angle. In life a substantial shear force accompanies high compressive loading in flexed postures[36] and it was our intention to simulate *in vivo* loading as closely as possible.

The results obtained using these techniques agree in many respects with previously published work. The average bending moment at the limit of flexion (61.2 Nm: see Table 1) compares with our own previous values of 49.4 Nm[13] and 51.7 Nm[18]. Larger bending moments are required to cause gross damage[37,38]. The average flexion angle (12.7°) is higher than our previous value of 8.7°[18] but this can be attributed to the period of creep loading[22] and to a greater preponderance of mobile L_{4-5} specimens in the present study. The compressive strengths of motion segments tested in pure compression ('0% flexion') are in the same range as those reported previously[5,6] although exact comparisons are prevented by the wide variability due to specimen, sex, age, and body mass. The

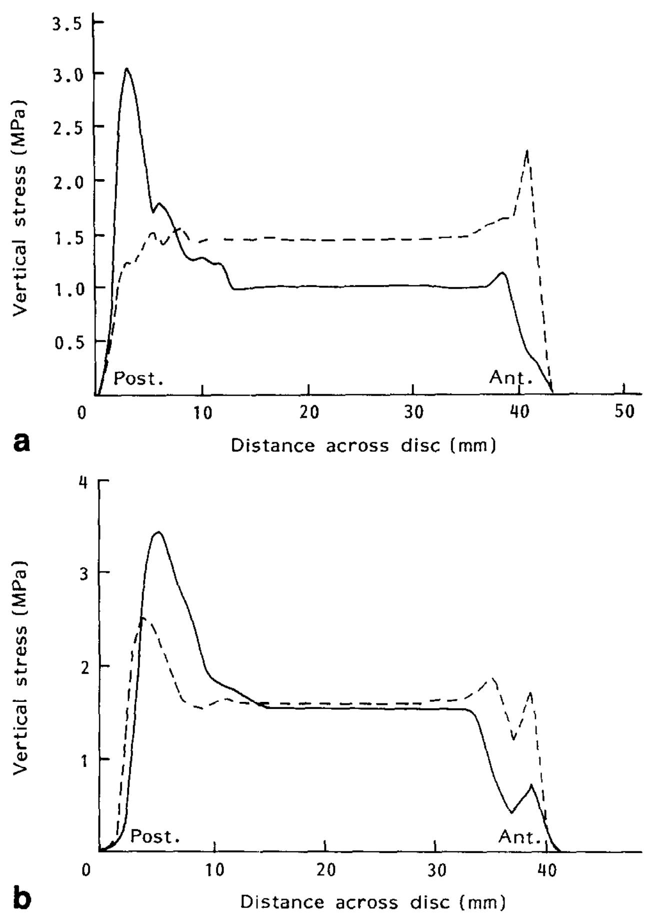

Figure 9. Lordotic posture, simulated by 2° of extension, generates high peaks of vertical compressive stress in the posterior annulus. Flexion **a**, removes or **b**, reduces this peak, but may increase stresses in the nucleus and anterior annulus. —— 2° ext.; ------ 10° flex. **a**, F, 31, L_{4-5}, 2000 N, after creep; **b**, M, 34, L_{4-5}, after creep.

Table 3. Results from the compressive strength tests on the matched pairs of motion segments. The differences between the strength of the two motion segments from each spine are shown. Differences were corrected to account for changes in strength at different lumbar levels. Specimens 20–23 had no neural arch when tested. Their strengths are shown for comparison only

Specimen no.	*Flexion angle (°)*	*Lumbar level*	*Compressive strength (N)*	*Difference (N)*	*Difference after correction (N)*
1	0	4–5	2704	−249	+103
2	4	2–3	2450		
3	9	3–4	7947	+736	−297
4	0	1–2	7211		
5	9	4–5	7974	+49	−988
6	0	2–3	7925		
7	9	2–3	5772	−1677	−709
8	0	4–5	7449		
9	0	2–3	4245	+187	−389
10	7	4–5	4432		
11	9	2–3	6805	−552	+404
12	0	4–5	7357		
13	0	4–5	10064	−746	+562
14	7	2–3	9318		
Mean (SD)					−188 (572)
20	0	4–5	9307	−586	+624
21	10	2–3	8721		
22	0	2–3	10469	+918	−562
23	10	4–5	11387		

compressive strengths of our flexed motion segments show no marked differences from previous results[39,40].

These last two studies also considered the effect of flexion on a motion segment's compressive strength. Our own results[40] suggested that flexed motion segments might actually be stronger, but the experiment did not compare 'flexed' and 'lordotic' specimens from the same spine, and the high average strength reported for young male spines (10219 N) may simply reflect strong or physically active individuals. Granhed et al.[39] showed that the relationship between vertebral compressive strength and bone mineral density was not significantly affected by whether the specimens were flexed or not, implying that flexion did not significantly weaken the motion segments. In several instances, they tested 'flexed' and 'lordotic' motion segments from the same spine and the flexed specimens tended to be weaker. This may be because the flexion angles used (usually 10° or 15°) were high for specimens that had not previously been creep-loaded, and so would have generated high ligament forces and high intradiscal pressures (Table 2).

The presence of such internal forces acting within the motion segment can be inferred from our intradiscal pressure measurements, which show that ligament tension remains low for much of the range of flexion, but then increases rapidly as the limit of flexion is approached (Figure 2). Conversely, loading of the apophyseal joints increases markedly in lordotic posture (Figure 3). Similar effects have been described before in cadaveric experiments[41–43] but it was necessary for our purposes to extend these findings to cover the full physiological range of flexion and extension, and to consider higher compressive forces.

By doing so we were able to demonstrate the lumbar spine's vulnerability to compressive damage in lordotic postures. The exact nature of the damage sustained by the motion segments at −2° and −4° is difficult to determine, but in one case subsequent loading cycles applied to the isolated disc at −6° showed no further damage. This indicates that earlier damage at −4° was sustained by the neural arch, either in the apophyseal joints, or the interspinous ligament. However, the occasional finding of high stress peaks in the posterior annulus fibrosus (Figure 9) suggests that disc damage cannot be ruled out. Graphs of compressive force against IDP (Figure 5) suggest that damage mostly occurs at about 1000 N. The force acting on the neural arch is then about 300 N, since in this range of force the IDP is reduced by about 30% compared to 0% flexion (Table 2, column 2). Higher compressive forces appear to cause little extra damage. Perhaps the apophyseal joint capsules are ruptured at about 1000 N, allowing the inferior articular processes to deflect posteriorly under higher loads[10,44] and thereby avoid further damage? Certainly this would explain why apophyseal joint loading (as inferred from changes in IDP: see Table 2) does not rise as rapidly as the applied compressive force.

Experiment 1 clearly defined a range of movement, between 0% and 75% flexion, in which the neural arch was unloaded and the tension in the posterior ligaments was slight (Table 2). Within this range, compressive forces are resisted almost entirely by the intervertebral disc. Therefore, in Experiments 2 and 3, the matched-pair comparison of 0% and 75% flexion relied on intradiscal stress profiles and measurements of ultimate compressive strength. The averaged results derived from the profiles (Figure 8) show that at 500 N, flexion increases intradiscal stresses in all regions of the disc.

However, the increased pressure (IDP) in the nucleus pulposus largely disappears when the compressive force rises to 3000 N (Figure 4) and this probably explains why the compressive strength of the motion segments was the same in both postures. Flexion may stiffen the anterior annulus so that it resists an increasing proportion of the applied compressive force and is able to 'stress-shield' the nucleus[7]. Also, a high compressive force will press the vertebrae more closely together and reduce the tension in the posterior ligaments.

Overall, the results concerning nuclear pressure (IDP) and compressive strength suggest no clear advantage for either 0% or 75% flexion. However, the evidence from the stress profiles must be considered quite apart from their effect on compressive strength. This is because the annulus fibrosus has a very limited capacity to repair itself, and stress concentrations which are insufficient to cause failure in a single loading cycle may nevertheless lead, in the course of time, to progressive disruption of the lamellar structure which is essential for normal disc function. The high stress peaks shown in Figures 6, 7, and 9 indicate high radial stress gradients acting in opposite directions to pull apart a thin region of annulus. They may thus contribute to the separation and inwards buckling of the inner lamellae found in early stages of disc degeneration[45] and in animal models of disc failure[46]. Flexion tends to remove these stress peaks from the posterior annulus without generating comparable peaks in the anterior annulus (Figures 6 and 7) and so it might be thought that the optimum position for the lumbar spine would be at the flexed end of the range defined above. However, there are other stresses acting within the annulus which are not detected by our transducer, and these are the tensile stresses acting in the highly collagenous outermost lamellae. The outer lamellae act like a tensile 'skin' surrounding the highly hydrated regions of the disc[12]. They allow considerable stretching in the vertical direction[23] but can resist tensile stresses of up to 9 MN/m^2 when stretched vigorously[47]. A disc's resistance to bending remains low up to about 50% flexion[13] so it is likely that the tensile stresses in the outer annulus are also low at 50% flexion. Therefore a balance could be achieved between the *compressive* stresses in the inner posterior annulus and the *tensile* stresses in the outer posterior annulus if the motion segment were flexed by about 50%. It would be reasonable to conclude, therefore, that although the compressive strength of the lumbar spine is largely unaffected by angulation within the range 0–75% flexion, the optimum stress distribution in the annulus occurs at about 50% flexion.

Flexion angles for motion segments can be converted into a scale appropriate for living people, in which 0% and 100% flexion refer to the upright standing and toe-touching postures respectively[18]. Hence, 0% and 75% flexion for motion segments corresponds to about 20% and 100% of lumbar flexion *in vivo*, while an extension angle of 2° in a motion segment roughly corresponds to erect standing in a living person[48]. The results of the present experiments therefore suggest that the lumbar spine should be flexed by about 80% of its *in vivo* range in order to achieve an optimum compressive strength, and an even distribution of stress in the annulus fibrosus.

This conclusion may be applicable to lumbar posture during heavy lifting, but factors outside the scope of this cadaveric study must also be considered. Some degree of lordosis may improve the ability of the erector spinae muscles to counter the forward shear force acting on the spine[21] whereas flexed postures which stretch the non-contractile tissues in and around the erector spinae muscles may enhance their ability to generate powerful extensor moments[49].

Conclusions

In full flexion, intradiscal pressure (IDP) is high because of tension in the posterior intervertebral ligaments. In extension (lordosis) the neural arch becomes weight-bearing and can be damaged by compressive forces as low as 500 N. Therefore the lumbar spine is best able to resist high compressive forces when positioned between 0% flexion and 75% flexion.

Within this range, IDP at high load levels is little affected by flexion/extension angle. The compressive strength of a motion segment is the same at 0% and 75% flexion.

Intradiscal measurements of compressive stress usually reveal stress peaks in the inner posterior annulus at 0% flexion, and an even stress distribution at 75% flexion. Tensile stresses in the outer posterior annulus increase considerably above 50% flexion. Therefore, the optimum stress distribution in the disc is achieved at about 50% flexion.

Comparisons between the ranges of sagittal plane movement *in vivo* and *in vitro* suggest that, during heavy lifting activities, the lumbar spine should be flexed by about 80% to achieve an optimum compressive strength and an even distribution of compressive and tensile stresses in the annulus fibrosus.

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