J. Biomechanics Vol. 24, No. 2, pp. 117-126, 1991.
Printed in Great Britain

0021-9290/91 $3.00+.00
Pergamon Press plc

A TECHNIQUE FOR QUANTIFYING THE BENDING MOMENT ACTING ON THE LUMBAR SPINE *IN VIVO**

M. A. ADAMS† and P. DOLAN
University of Bristol, U.K.

Abstract—The bending properties of cadaveric lumbar spines were measured and used to convert *in vivo* measurements of lumbar flexion into bending moments ('stresses').

Forty-two lumbar motion segments were subjected to complex physiological loading and graphs were obtained of bending moment vs flexion angle. Variability was reduced by expressing both variables as a percentage of their values at the elastic limit. Data were averaged for each lumbar level, and a composite bending curve was compiled for the lumbar spine, L1–S1. A linear relationship was established between lumbar flexion measured *in vitro* and *in vivo*. This enabled values of 'per cent lumbar flexion' measured *in vivo* to be converted into 'per cent maximum bending moment' with a maximum likely error of about $\pm 8\,\%$, which is equivalent to about ± 5 Nm at L5–S1 for an average person.

The technique was applied to 28 subjects, using dynamic measurements of lumbar flexion obtained with the '3-Space Isotrack' system. The bending moment at L5–S1 was 12 Nm on average when picking a pen up off the floor. Highly significant increases in bending moment were observed when heavier and bulkier objects were lifted.

INTRODUCTION

Attacks of low back pain and sciatica are often started by forward bending and lifting activities, and occupations that involve a lot of bending and lifting are associated with an increased incidence of low back problems in general (Chaffin and Park, 1973; Frymoyer *et al.*, 1983; Svensson and Andersson, 1983; Troup, 1965; Troup *et al.*, 1981) and acute disc prolapse in particular (Kelsey *et al.*, 1984). Biomechanical analyses show that these activities greatly increase the compressive force on the lumbar spine (Bartelink, 1957; McGill and Norman, 1987) and this has been confirmed by *in vivo* measurements of lumbar intradiscal pressure (Nachemson, 1981). Currently, most investigations of spinal loading concentrate on the compressive force (Ortengren *et al.*, 1981; Schultz *et al.*, 1982) and industrial safety standards in the U.S.A. are based on minimising compression (NIOSH, 1980).

Compression, however, is not the whole story. Experiments on cadaveric spines suggest that compression is particularly damaging to the intervertebral discs when it is combined with bending in the sagittal plane. A combination of bending and compression can cause the intervertebral discs to prolapse posteriorly, either by a sudden extrusion of nucleus pulposus into the spinal canal (Adams and Hutton, 1982, 1988) or by the gradual formation, in response to fatigue loading, of a complete postero-lateral radial fissure which then 'leaks' nuclear material into the intervertebral foramen (Adams and Hutton, 1985, 1988). If the magnitude of the bending moment is decreased even slightly then no radial fissures can be produced, even if the compressive force remains high (Adams and Hutton, 1985). The failure mechanisms involved have been demonstrated by finite element analysis (Shirazi-Adl, 1989). High bending moments have also been shown to damage the ligaments of the neural arch and apophyseal joints (Adams *et al.*, 1980). In the absence of any bending moment, compressive failure of a cadaveric motion segment invariably occurs in the vertebral body, without posterior displacement of disc material (Brinckmann *et al.*, 1989; King Liu *et al.*, 1983; Lin *et al.*, 1978; Hardy *et al.*, 1958; Perey, 1957). This happens even if the posterior annulus is artificially weakened prior to the compressive force being applied (Brinckmann, 1986).

The cadaveric evidence, therefore, suggests that *in vivo* assessments of spinal loading should take account of the bending moment as well as the compressive force if the results are to be of direct clinical relevance. The purpose of the present study was to provide a technique to make this possible.

The approach adopted was to estimate bending moment from the angular deformation (flexion) that it produces in the osteo-ligamentous lumbar spine, as shown schematically in Fig. 1.

THE FORWARD BENDING PROPERTIES OF CADAVERIC LUMBAR SPINES

Cadaveric material

Lumbar spines were obtained at autopsy within 24 h of death from subjects who had not suffered spinal injury or prolonged bed rest prior to death. For convenience, they were stored in sealed plastic bags at $-17\,^{\circ}$C for up to one month prior to testing. Each

Received in final form 21 *August* 1990.

* ESB Award Paper 1990, European Society of Biomechanics.

† Address correspondence to: Dr M. A. Adams, The Comparative Orthopaedic Research Unit, Departments of Anatomy and Orthopaedic Surgery, University of Bristol, Bristol BS8 5LS, U.K.

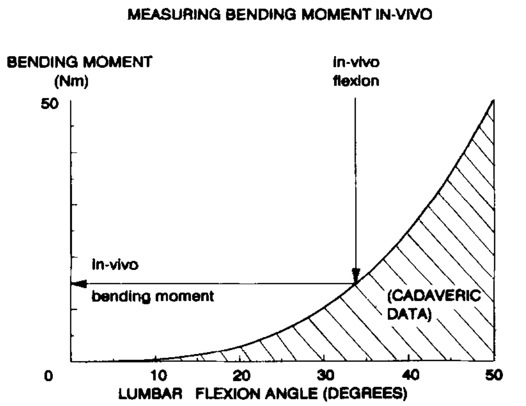

Fig. 1. In principle, bending moment can be quantified *in vivo* by measuring lumbar flexion and comparing it with the bending stiffness properties of the osteo-ligamentous lumbar spine.

spine was defrosted overnight at +3 °C and then dissected into motion segments consisting of two vertebrae and the intervening soft tissue. After dissection, one motion segment was tested immediately while the other was kept in a sealed plastic bag at +3 °C and tested the following day. The sex, ages and body masses of the 42 motion segments used are given in Table 1.

Each motion segment was set in two cups of dental plaster prior to mechanical testing. Full details of dissection and setting procedures have been described previously (Adams and Hutton, 1985).

Mechanical testing of motion segments in bending and compression

Motion segments were loaded on a 20 kN Dartec hydraulic materials testing machine controlled by a microcomputer (Fig. 2). The actuator of the machine was programmed to move upwards at a uniform displacement rate so that a roller attached to the upper cup made contact with a horizontal plate, causing the motion segment to flex forward. Each loading/unloading cycle lasted 2.0 s and during this time the vertical displacement of the actuator (x) and the vertical force on the load cell (F) were recorded by the microcomputer 500 times per second.

The apparatus shown in Fig. 2 not only applies a bending moment to the motion segment but a compressive force as well, increasing from zero up to a peak value ranging between about 500 and 2000 N, depending upon the size of the specimen. A compressive pre-load is necessary because the bending properties of motion segments are influenced by the compressive force acting on them (Janevic *et al.*, 1989; Panjabi *et al.*, 1977). In life, the compressive force rises from about 500 N in erect standing to 1700 N when bending to lift a 10 kg weight (Nachemson, 1981) and so our apparatus simulates this movement to a reasonable approximation. The ratio between compressive force and bending moment was determined by the

Table 1. Details of the 42 motion segments used in the experiments. The values of bending moment and flexion angle at the elastic limit show considerable variation

Specimen details			Results at the elastic limit	
Sex	Age	Lumbar level	Flexion angle (degrees)	Bending moment (Nm)
F	23	1–2	6.0	16.5
M	24	1–2	6.1	75.0
F	31	1–2	7.0	34.2
M	18	1–2	5.3	64.7
F	47	1–2	6.0	27.7
M	47	1–2	9.0	39.0
M	55	1–2	7.0	37.3
M	62	1–2	8.2	44.2
F	68	1–2	7.7	20.0
M	20	2–3	7.9	100.4
M	26	2–3	7.0	34.3
F	37	2–3	9.0	37.7
M	42	2–3	7.0	51.5
M	47	2–3	7.0	48.1
M	47	2–3	9.0	53.4
M	52	2–3	8.5	70.7
M	58	2–3	6.1	85.2
M	58	2–3	8.5	70.7
M	62	2–3	10.7	45.2
F	71	2–3	6.6	25.3
M	23	3–4	7.3	79.7
F	31	3–4	8.6	37.5
M	50	3–4	8.0	72.7
F	56	3–4	6.1	50.8
M	62	3–4	10.1	48.1
F	68	3–4	11.0	37.0
M	18	4–5	11.8	71.1
F	37	4–5	11.0	32.2
M	42	4–5	11.0	39.7
M	47	4–5	10.9	49.6
M	47	4–5	12.0	67.8
M	52	4–5	11.3	48.2
M	58	4–5	10.0	99.4
F	64	4–5	7.1	28.4
M	70	4–5	10.8	36.4
F	71	4–5	8.8	28.0
M	25	5–1	11.0	109.5
M	24	5–1	12.0	74.2
F	31	5–1	10.5	43.0
M	40	5–1	11.4	55.0
F	56	5–1	5.7	49.7
F	68	5–1	8.3	33.7
Mean	46.5			51.7

load offset distance (d in Fig. 2). A value of $d = 30$ mm was found to be suitable for most specimens. As the specimen flexed forward, d increased to about 50 or 60 mm.

Each motion segment was loaded up to 2000 N with $d = 0$ mm in order to determine its compressive properties. Then, with d set to about 30 mm, the specimen was twice flexed to a moderate flexion angle and the force–deformation curves recorded. If the residual deformation between the loading cycles was negligible, then the flexion angle was increased by 1° and two more flexion curves obtained. This procedure was

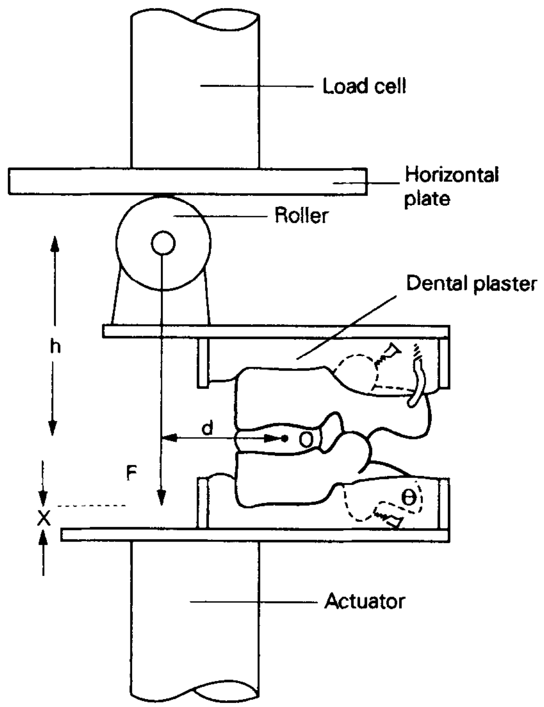

Fig. 2. The apparatus used to load motion segments in combined bending and compression.

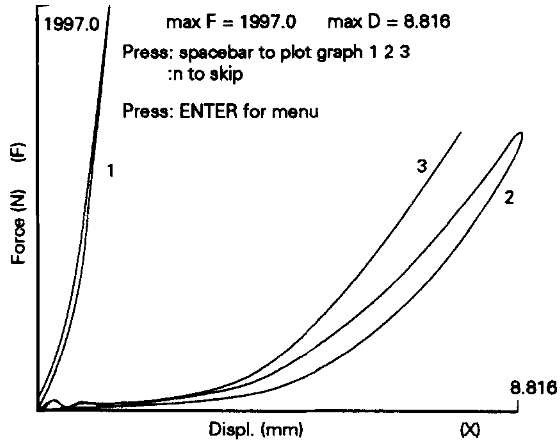

Fig. 3. The raw data from the cadaveric experiments. The first step in data analysis was to subtract the compression curve (1) from the bending curve (2) for the first half of the loading cycle.

repeated until a pair of curves indicated substantial residual deformation. The first of these curves always showed a slight change in gradient at high load, confirming that the elastic limit had just been exceeded. The second curve recorded at a flexion angle 1° less than this was recorded for analysis.

An additional test was performed on two motion segments. They were repeatedly flexed to the same angle, while the initial load offset distance d was varied between 5 and 60 mm. This was done to assess the influence of different compressive pre-loads on the specimens' bending properties.

Calculating bending moment and flexion angle data

The raw data were processed in order to obtain graphs of bending moment against flexion angle. Firstly, the compression curve was subtracted from the bending curve (at constant displacement) to eliminate that component of vertical displacement due to compression of the specimen and apparatus (Fig. 3). This procedure introduces a slight error, because the compressive deformation of a motion segment is affected by the bending moment acting upon it. However, even a 20 % change in compressive deformation would have only a slight effect on curve 3 in Fig. 3 because the displacement due to bending is much greater than that due to compression.

Secondly, the flexion angle was calculated using the following formula:

$$\text{flexion} = \Theta - \arctan(d/h)$$

where h is the vertical distance between the centre of the disc and the axis of the roller, d is the initial roller offset distance, x is the vertical displacement of the actuator, and Θ is defined by:

$$\cos(\Theta) = (h - x)/(d^2 + h^2)^{1/2}.$$

These equations assume that the centre of rotation lies in the geometric centre of the disc. Thirdly, the bending moment acting about the centre of the disc was calculated from the formula:

$$\text{bending moment} = F \times \sin(\Theta) \times (d^2 + h^2)^{1/2}$$

where F is the vertical force acting on the horizontal plate.

The validity of these formulae was tested by measuring the flexion angle independently, using the '3-Space Isotrack' system operating at 60 Hz (see below). The calculated and measured peak flexion angles never differed by more than 1°. It was necessary to calculate flexion angles because the Isotrack has a resolution of about 0.35°.

Results

A typical set of force–deformation curves for one motion segment flexed just short of the elastic limit is shown in Fig. 3. The average compressive pre-load at the elastic limit was 925 N (standard deviation 261 N, range 476–1795 N). Force and deformation were digitally transformed into bending moment and flexion angle as described above, and the values of these variables at the elastic limit are shown for each motion segment in Table 1.

Reducing the variability of the cadaveric data

There was considerable variation in the strength and flexibility of different motion segments. The following steps were taken in order to derive a consistent relationship between bending moment and flexion angle.

(1) Bending moment was expressed as a percentage of its value at the elastic limit. This reduced variability due to differences in specimen size and strength.

(2) Flexion angle was expressed as a percentage of its value at the elastic limit. This reduced variability due to differences in mobility (and age).

With bending moment and flexion angle normalised in this way, it became possible to make quantitative comparisons between motion segments.

The bending moment (expressed as a percentage of the maximum bending moment) was evaluated at 10 % intervals of full flexion, from 0 % up to 100 %. Preliminary statistical analysis then showed that these normalised bending moment values were significantly dependent on lumbar level, but not on age, sex, body mass or range of flexion. Therefore, results were averaged for each lumbar level, and these mean values are shown in Fig. 4. Table 2 gives both the mean values and the standard deviations. The variability for a particular lumbar level (L3–4) is depicted in Fig. 5. The broken lines refer to the mean value plus, or minus, one standard deviation, so that 67 % of curves for individual motion segments might be expected to lie between them.

Bending properties of the whole lumbar spine

The data were further processed in order to construct a composite curve of bending moment against flexion angle for the whole lumbar spine. This time, the '% flexion' value was evaluated at 10 % intervals of bending moment, and mean values obtained for each lumbar level. Then for each interval of bending moment, the mean 'per cent flexion' value for each lumbar level was multiplied by a weighting factor which expressed the relative mobility of that lumbar level, and then summed with the values for the other four levels.

The weighting factors were obtained from previous experiments on 87 motion segments aged between 15 and 49 yr (mean 33.1 yr) which showed the average

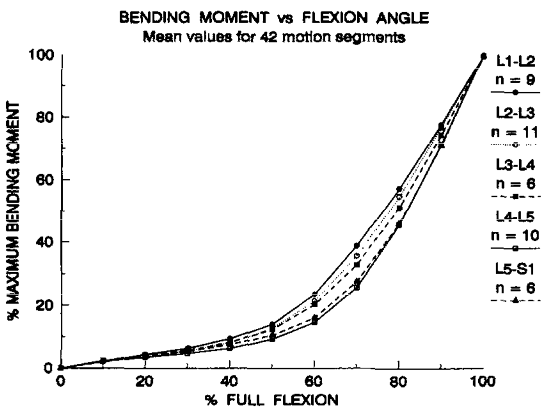

Fig. 4. Variability of the cadaveric data was greatly reduced by expressing bending moments and flexion angles as a percentage of their values at the elastic limit. There are still significant differences between lumbar levels.

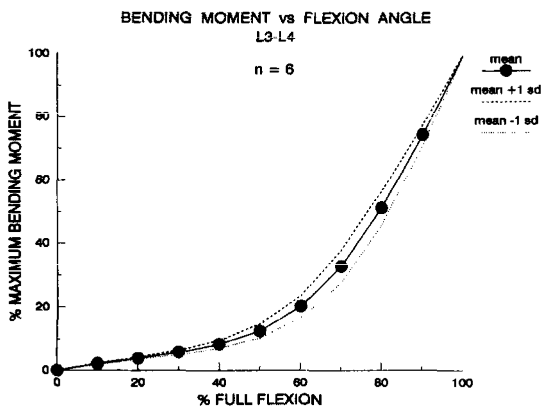

Fig. 5. Variability of the bending properties of the six L3–L4 motion segments tested, after bending moment and flexion angles have been normalised.

Table 2. Bending moment resisted at various intervals of the full range of flexion. Bending moment is expressed as a percentage of the bending moment resisted at the elastic limit (100 % flexion)

						Percentage of the maximum bending moment								
						Per cent of full flexion								
Lumbar level	*n*		Age	Range of flexion	Max BM (Nm)	10	20	30	40	50	60	70	80	90
1–2	9	Mean	44.1	6.9	39.8	2.2	4.4	6.4	9.4	14.0	23.5	38.8	57.3	77.8
		STD	16.1	1.2	19.4	0.8	1.6	2.5	3.2	4.2	5.5	5.3	4.0	2.2
2–3	11	Mean	47.1	7.9	56.6	2.1	3.7	5.5	8.2	12.8	21.4	35.6	54.8	76.9
		STD	15.7	1.4	22.8	0.9	1.0	1.4	2.1	3.7	6.8	8.0	5.9	3.4
3–4	6	Mean	48.3	8.5	54.3	2.2	3.8	5.7	8.2	12.4	20.2	32.8	51.2	74.4
		STD	17.8	1.8	18.0	0.3	0.4	0.8	1.2	2.2	3.4	5.1	5.3	3.2
4–5	10	Mean	50.6	10.5	50.1	2.3	3.6	4.7	6.3	9.3	14.6	25.7	45.6	71.4
		STD	16.2	1.5	23.0	2.0	1.9	1.9	2.2	3.6	6.3	8.8	7.9	4.4
5–1	6	Mean	40.3	9.8	60.9	2.1	3.8	5.4	7.5	10.6	16.0	27.6	46.0	71.1
		STD	18.3	2.4	27.4	0.8	1.0	1.8	3.0	5.0	7.6	11.1	12.0	6.9

range of flexion at each lumbar level to be 8° at L1–2, 9° at L2–3, 9° at L3–4, 12° at L4–5, and 12° at L5–S1 (Adams and Hutton, 1986), giving a total range of flexion of 50° for the whole lumbar spine. Therefore the weighting factor for L1–2 was 8/50, and for L2–3 was 9/50, and so on. By repeating this summation procedure at each 10 % interval of bending moment, a composite curve for the whole lumbar spine was obtained, as shown in Fig. 6. A polynomial was curve-fitted to the experimental data using the method of least squares and an excellent fit ($R > 0.999$) obtained with the following equation:

$$M = 1.03\ T^3 \times 10^{-4}$$

$$\text{or, for easy computation, } M = 0.1 \times (0.1\ T)^3 \quad (1)$$

where M, the bending moment and T, the flexion angle are both expressed as a percentage of their values at the elastic limit.

The whole procedure was repeated using the mean values plus, or minus, one standard deviation, and the results are shown by the open circles in Fig. 6. The solid line in Fig. 6 expresses the average bending characteristics of cadaveric lumbar spines, and 67 % of spines might be expected to fall within the boundaries marked by the open circles.

The effect of compressive pre-load

Figure 7 shows the results of the additional test which examined the effect of changing the compressive pre-load on the bending properties of two motion segments. Increasing the pre-load from 500 to 1400 N increased the resistance to bending by about 30 %. In life, the compressive force increases by an amount similar to this when a person bends forwards (Nachemson, 1981) so the variable pre-load applied in the main experiment is confirmed as an essential feature of the experimental design.

DYNAMIC MEASUREMENTS OF LUMBAR FLEXION *IN VIVO*

Lumbar flexion was measured with the '3-Space Isotrack' system which has been described previously (An *et al.*, 1988; Pearcy and Hindle, 1989). Briefly, a source of pulsed electromagnetic waves was attached to the skin surface overlying the sacrum, and a sensor of these waves was attached to the skin overlying the spinous process of L1. The output from the sensor gave the angle subtended between source and sensor, which in this case was the lumbar curvature in the sagittal plane. The lumbar curvature was measured while the subject was standing erect and while touching his toes, and the difference in curvature was taken as the range of lumbar flexion. We have previously shown that flexion angles measured by a device similar to the Isotrack bear a close linear relationship ($R = 0.91$) to angular movements of the vertebrae (Adams *et al.*, 1986). The lumbar curvature in the two 'end points', erect standing and full flexion, was found to be reproducible to within $\pm 2°$ when measured with the Isotrack (Dolan and Adams, 1990). Systematic errors in flexion angles can arise with the Isotrack (Pearcy and Hindle, 1989) but we were able to eliminate these by mounting the source and sensor on the back with the wires aligned horizontally so that they did not move appreciably during flexion movements (Dolan and Adams, 1990). Using this method, we tested the Isotrack system on a subject whose range of flexion had previously been established as 55° by X-ray measurements; the range of flexion measured with the Isotrack ($56.4 \pm 2.6°$, $n = 9$) was not significantly different.

During dynamic movements, the lumbar curvature was measured at a frequency of 28 Hz and the data stored on an Opus PC3 microcomputer. The curvature in erect standing was subtracted from these values in order to obtain lumbar flexion, and this was then

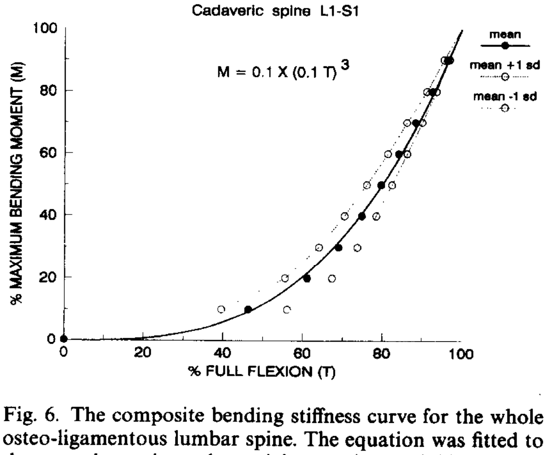

Fig. 6. The composite bending stiffness curve for the whole osteo-ligamentous lumbar spine. The equation was fitted to the mean data using polynomial regression, and this relationship is shown by the solid line.

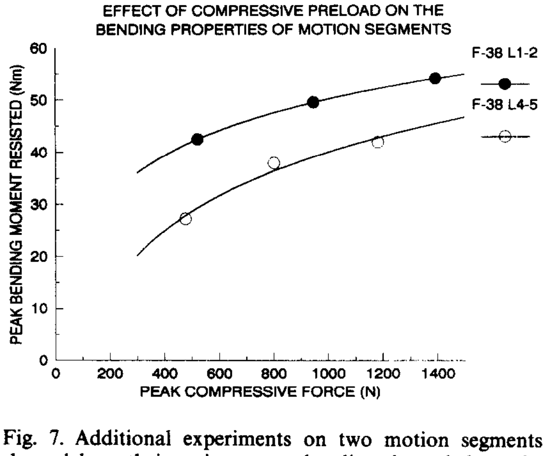

Fig. 7. Additional experiments on two motion segments showed how their resistance to bending depended on the compressive pre-load. The L1–2 specimen was flexed to 10.6° three times, and the L4–L5 specimen was flexed to 11.6°. Neither specimen was taken beyond the elastic limit.

expressed as a percentage of that subject's full range of flexion.

COMPARING FLEXION *IN VITRO* AND *IN VIVO*

Figure 6 is not sufficient to estimate bending moments from *in vivo* flexion angles because the scale of flexion for a cadaveric specimen is different from the scale for a living person. For a cadaveric specimen, '0 % flexion' refers to the unloaded state, and '100 % flexion' is the elastic limit. In a living person, it is convenient to consider '0 % flexion' as standing erect, and '100 % flexion' as the extreme toe-touching posture. Extreme backward bending was not used to represent zero flexion because we found it to be more variable when measured with the Isotrack system.

It can be assumed that the *in vivo* and *in vitro* scales are linearly related, so

$$T = k1 + k2 \times V \tag{2}$$

where T is the *in vitro* per cent flexion and V is the *in vivo* per cent flexion. Two points of correspondence between the scales are required to solve for $k1$ and $k2$.

Let v be the *in vivo* range of flexion, in degrees; let t be the *in vitro* range of flexion, in degrees; let $v(0)$ be the *in vivo* flexion angle, in degrees, that corresponds to the unloaded spine *in vitro*. Then, with reference to Fig. 8, it can be seen that:

$$\text{when } T=0, \quad V = 100 \times v(0)/v$$

$$\text{when } V=100, \quad T = 100 \times [v - v(0)]/t.$$

Substituting in equation (2) yields:

$$k1 = -100 \times v(0)/t$$

$$k2 = v/t$$

so that equation (2) becomes:

$$T = -100 \times v(0)/t + (v/t) \times V. \tag{3}$$

Suitable values of v, t, and $v(0)$ can be obtained from the literature. The mean *in vivo* range of flexion (v) of 49 healthy volunteers of mean age 30 yr was 57.9°.

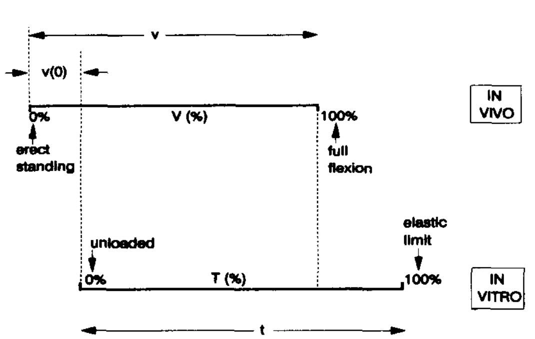

Fig. 8. Lumbar flexion angles measured *in vivo* must be converted to equivalent angles on the *in vitro* scale before they can be used to calculate bending moment. See text for details.

(This angle is the mean value from the present study and a previous one by Adams *et al.* (1987), both of which measured flexion from changes in lumbar curvature. This value is used in preference to other available data because the subjects were encouraged to flex as far as they possibly could.) The mean *in vitro* range of flexion (t), as calculated from experiments on 87 cadaveric lumbar motion segments of mean age 33.1 yr, was 50.1° (Adams and Hutton, 1986). (These data were used because both the mean age and the age distribution closely matched that of the *in vivo* data. The cadaveric specimens used in the present study were older and therefore less mobile.) This angle (t) should be increased by 12.5° to 62.6° to account for the increased mobility which occurs after a period of creep loading has reduced disc height by an amount equal to the diurnal height loss seen in life (Adams *et al.*, 1987, 1990). The cadaveric data are then comparable to *in vivo* data acquired in the afternoon or evening. Finally, comparisons between the curvature of the lumbar spine in the erect standing posture and that of excised cadaveric lumbar spines suggest that $v(0)$ is 14.1° on average (Adams and Hutton, 1986). Substituting for v, t and $v(0)$ in equation (3) gives the following relationship between the *in vivo* and *in vitro* flexion scales:

$$T = -22.5 + 0.93\ V. \tag{4}$$

DYNAMIC MEASUREMENTS OF BENDING MOMENT *IN VIVO*

Method

Twenty-eight subjects, none of whom had any history of low back pain, were asked to adopt certain postures and to perform a series of forward bending and lifting exercises while their lumbar flexion was measured as described above. Equations (4) and then (1) were used to transform the peak normalised flexion angle recorded in each posture or activity into the bending moment acting on the lumbar spine.

Results

Table 3 gives the peak bending moment recorded in three of the exercises. The values vary considerably between individuals. Some of this variation is attributable to a highly significant negative correlation between peak bending moment and the mobility of the lumbar spine and hips (Dolan and Adams, 1990). The rest may be due to random variation in the bending–flexion curves of individual spines as indicated by the wide error limits in Fig. 6.

The results are much less variable, however, when the three activities are compared for a given individual. Peak bending moments in the three activities consistently follow the pattern: $c > b > a$ and these differences are highly significant ($P < 0.0001$, using matched pair t-tests).

Table 4 shows the bending moment recorded in each posture and movement investigated. These were

Table 3. Peak values of bending moment acting on the lumbar spines of 28 subjects while they lifted each of three objects from the floor: (a) a pen; (b) a small 10 kg weight; and (c) a large box weighing 10 kg. Bending moment is expressed as a per cent of the bending moment at the elastic limit. The final columns indicate consistent and highly significant intra-individual differences between the three lifts

		Per cent maximum bending moment				
Sex	Age	Pen (a)	10 kg Weight (b)	10 kg Box (c)	(b–a)	(c–b)
F	33	11.1	21.4	29.7	10.3	8.3
M	36	14.1	22.3	25.6	8.2	3.3
F	32	26.2	27.7	29.5	1.5	1.8
F	35	28.1	19.5	31.1	−8.6	11.6
M	30	24.5	36.4	42.7	11.9	6.4
M	36	11.7	22.4	27.8	10.8	5.4
F	33	24.1	20.5	31.9	−3.6	11.3
M	28	32.6	39.4	45.0	6.8	5.6
F	21	31.3	33.0	36.2	1.7	3.1
F	22	21.7	29.7	31.7	8.0	2.0
F	37	19.8	21.3	26.4	1.6	5.0
F	36	21.1	20.2	24.7	−1.0	4.5
M	25	6.2	11.1	23.3	5.0	12.2
M	28	11.4	18.6	21.7	7.3	3.1
M	36	15.9	24.1	27.6	8.2	3.5
F	33	18.3	24.3	33.0	6.0	8.7
M	36	15.3	19.6	25.6	4.4	6.0
F	19	33.4	44.7	42.2	11.3	−2.5
F	25	18.4	17.8	16.3	−0.5	−1.5
M	45	23.5	28.8	34.9	5.3	6.0
M	36	16.8	25.8	32.1	9.1	6.3
M	22	19.2	28.8	32.7	9.5	4.0
F	44	10.3	10.0	15.9	−0.3	5.8
F	27	31.5	40.1	38.3	8.7	−1.8
M	33	23.8	28.2	26.1	4.3	−2.0
M	35	28.0	35.5	38.1	7.6	2.5
M	58	22.7	26.2	31.2	3.5	5.0
F	21	6.9	26.7	22.0	19.7	−4.7
Mean	32.2	20.3	25.9	30.1	5.6	4.3
STD	8.4	7.7	8.2	7.3	5.6	4.3

calculated from the 'per cent maximum bending moment' values using the average value of 60.9 Nm from Table 1 to represent the bending moment at the elastic limit for L5–S1 motion segments.

Table 4. Values of peak bending moment acting on the lumbar spine at the level L5–S1. These were calculated using the mean value of 'per cent bending moment' from our 28 subjects, and an average value of 60.9 Nm for the strength in bending of L5–S1 motion segments (see Table 2)

Posture/movement	Peak bending moment (Nm)
Squatting	6 ± 4
Sitting on floor, knees flexed	11 ± 4
Picking a pen up off the floor	12 ± 4
Lifting a compact 10 kg weight	16 ± 5
Lifting a large box weighing 10 kg	18 ± 4
Standing, attempting to touch toes	20 ± 3

DISCUSSION

The steps required to quantify bending moment acting on the lumbar spine *in vivo* are summarised in Fig. 9. Some of these require further justification.

Validity of the technique

The technique assumes that, when someone bends forwards, each lumbar motion segment experiences a bending moment that is proportional to its strength in bending, and that each is flexed by a similar proportion at any given point in the movement. In other words, the back muscles 'share out' the bending according to the strength in bending of each lumbar level, so that localised stress concentrations are avoided. There is some justification for this assumption: mobility studies of healthy people in the erect standing and toe-touching postures indicate that intervertebral flexion increases between L1–2 and L4–5, and decreases with age (Adams and Hutton, 1982; Jonck and van Niekerk, 1961; Pearcy *et al.*, 1984). A similar pattern of mobility, and of variation in mobility, was found in cadaveric motion segments in the present experiment and in a previous, larger study

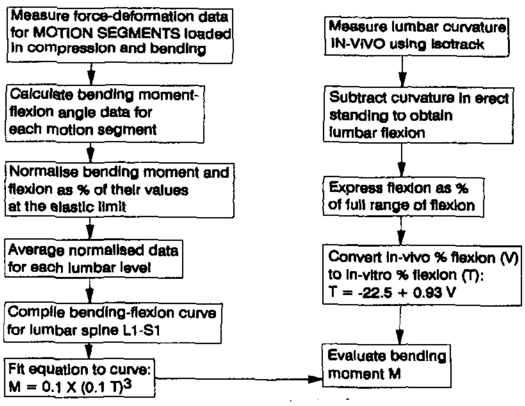

Fig. 9. The flow chart summarises the technique for quantifying bending moment *in vivo*.

(Adams and Hutton, 1986). This suggests that flexion angles *in vivo* are closely related to the bending properties of the underlying osteo-ligamentous spine.

It is possible, however, that a person with an injured back will attempt to protect the site of injury rather than share out the bending, and the technique may then be of less value. Also, activities which only partially flex the lumbar spine may cause an *in vivo* moment distribution which is different from that assumed *in vitro*. In order to check this possibility, we performed additional tests on two individuals and compared the 'per cent flexion' in the upper (L1–L3) and lower (L3–S1) lumbar spine with that of the whole lumbar spine (L1–S1). The values were very similar in flexed standing postures and when sitting on the floor, but erect sitting caused rather more flexion in the lower region of the lumbar spine than the upper region. This suggests that the technique is valid only for standing postures, and for those sitting postures which involve a substantial amount of lumbar flexion.

Another major assumption of the technique is that erect standing causes the lumbar spine of each individual to be extended by an angle equal to about 22.5 % of the range of flexion [equation (4)]. This is equivalent to saying that someone with a particularly lordotic/flat stance has a lumbar spine with a high/low natural curvature, and that there is a consistent tendency to increase this curvature when standing upright.

Several details of the cadaveric experiments require further justification. The material is dead tissue and tested at normal laboratory temperatures rather than 37 °C. However, what little evidence there is on post-mortem, freezing and temperature effects (Hasberry and Pearcy, 1986; Panjabi *et al.*, 1985; Smeathers and Joanes, 1988) suggests that they are unlikely to affect the spine's mechanical properties nearly as much as the natural variability shown in Fig. 6.

The formulae used to calculate bending moment and flexion angles for each motion segment assume that the centre of sagittal rotation remains in the centre of the disc throughout the range of movement. In fact, the centre of rotation probably moves slightly but this does not affect the validity of the technique: it just means that the estimated values of bending moment acting *in vivo* refer to the centre of the discs and not to the exact centre of rotation. According to a recent bilateral X-ray study *in vivo*, the distance between these two points is likely to be small (Pearcy and Bogduk, 1988).

The cadaveric bending stiffness data were obtained using compressive pre-loads appropriate to light manual tasks only. The graphs in Fig. 7, and also preliminary data from Janevic *et al.* (1989), suggest that the bending stiffness of motion segments continues to increase as the pre-load rises to higher levels. However the increase in stiffness accompanying an increase in pre-load from 2200 to 4400 N was not significant (Janevic *et al.*, 1989) so the present data may still be applicable to heavy lifting.

Accuracy of the technique

The largest source of error is the variability in the bending properties of different spines, even after the normalisation procedures. Figure 6 shows that 60 % of flexion implies a bending moment somewhere between 13 and 26 % of that at the elastic limit; for 70 % flexion, the figures are 24 and 40 % bending moment, and for 80 % flexion, 44 and 58 %. Thus it appears that the 'per cent bending moment' values have an uncertainty of about ± 7 % of the maximum bending moment. The experimental values in Table 3 have a standard deviation of about ± 8 %. This includes both the random error and the variability in subject performance, and so we suggest that it represents an upper bound for the uncertainty in 'per cent bending moment'. Using the mean value of 60.9 Nm for the mean strength in bending of L5–S1 motion segments (Table 2), this uncertainty can be roughly expressed as ± 5 Nm.

Although the random errors are large, the technique does not appear to be sensitive to sources of systematic error. Even if the *in vivo* flexion angles were consistently measured too low or too high (perhaps because of limitations in the Isotrack technique) the error would be reduced when the flexion angle was expressed as a percentage of the full range of flexion.

General discussion

The cadaveric data are in good agreement with previous investigations of the bending properties of lumbar motion segments. The average flexion angles at the elastic limit (Table 1) are of the same magnitude, and show the same variation with age and lumbar level, as those reported in a study on 122 motion segments (Adams and Hutton, 1986). The average bending moment required to reach the elastic limit in the present study (51.7 Nm) compares with an average of 49 Nm (Adams *et al.*, 1980) and 33 Nm (Adams *et al.*, 1987) in two previous studies from our laboratory. The value of 33 Nm was obtained using a criterion for the elastic limit which was perhaps over-cautious because the motion segments were to be taken to this limit more than once. The limit was determined more accurately in the present study because the rapid 2.0 s loading cycles reduced visco-elastic deformation to a minimum and made it easy to spot residual deformation due to specimen damage. [Beyond the elastic limit, damage occurs first in the supra- and interspinous ligaments (Adams *et al.*, 1980; McGill, 1988).] Most previous investigations of the bending properties of motion segments have kept well short of the elastic limit (Lin *et al.*, 1978; Markolf, 1972; Schultz *et al.*, 1979). The only comparable data are those of Miller *et al.* (1986). They reported that motion segments showed 'no overt signs of failure' at 59 Nm of bending, and flexed between 11.7 and 13.8° at 70 Nm. There is no discrepancy with the present results, since the elastic limit is not a particularly overt sign, and may well have been exceeded in Miller's experiment.

The bending stiffness curves shown in Figs 4, 5 and 6 show a considerable region of movement in which the bending moment on the spine is slight. It is commonly assumed that this is the normal 'physiological' range and that the bending moment on the spine can be ignored (Brinckmann *et al.*, 1989; Brinckmann, 1986; McGill and Norman, 1987). This assumption appears to be valid for erect sitting postures, but not for bending and lifting activities, during which all of the 28 subjects applied considerable bending moments to their spines. On average, lifting a 10 kg box generated a bending moment of 30 % of the maximum, and using our average value of 60.9 Nm from Table 2, this corresponds to about 18 Nm at the level of L5–S1. The peak value recorded was 45 % (27 Nm). Note that in the static, fully flexed posture (100 % *in vivo* flexion), equations (1) and (4) indicate that the bending moment is 35 % (21 Nm) so this static limit can be exceeded during dynamic movements. Rapid movements and heavy lifting may well increase the bending moment to the point where fatigue damage could accumulate in the ligaments and discs.

Peak bending moments are also likely to be higher in the early morning when the discs and ligaments resist much more of the bending moment acting on the spine (Adams *et al.*, 1987). Our technique must be adapted before it can be applied in these circumstances. Both the instantaneous flexion angles and the range of flexion must be measured in the morning. The value of 't' in equation (3) should be changed to 50.1° so that equation (4) becomes:

$$T = -22.5 + 1.16\, V.$$

The same cadaveric data can be used because the increased flexibility of the motion segments is adequately taken care of when flexion angles are divided by the full range of flexion. (This was checked by comparing normalised flexion curves for 13 motion segments, before and after a period of creep loading that simulated the diurnal disc height loss of about 1.5 mm. No significant differences were noted.) *In vivo* estimations of bending moment made in this earlier study (Adams *et al.*, 1987) are lower than those presented here. This is mainly due to the lower strengths of the cadaveric specimens upon which the estimates were based.

CONCLUSIONS

(1) The technique described above can measure the bending moment acting on the lumbar spine *in vivo* to an accuracy of about ±8 % of the bending moment at the spine's elastic limit. This is equivalent to about ±5 Nm at L5–S1 for an average person.

(2) The error is greatly reduced when different activities are compared using the same subjects, since intra-subject comparisons are not affected by the variation in the mechanical properties of different spines.

(3) The bending moment on the lumbar spine rises to about 18 Nm at L5–S1 during everyday bending and lifting tasks, but may be much higher in the early morning or when lifting heavy weights.

(4) Some of the assumptions incorporated in the technique make it unsuitable for analysing activities such as erect sitting which cause an uneven distribution of bending moment on the lumbar spine. Also, the technique may be unsuitable for use with patients who bend unevenly in an attempt to protect a painful motion segment.

Acknowledgement—This work was supported by a grant from the Arthritis and Rheumatism Council.

REFERENCES

Adams, M. A., Dolan, P. and Hutton, W. C. (1987) Diurnal variations in the stresses on the lumbar spine. *Spine* **12**, 130–137.

Adams, M. A., Dolan, P., Hutton, W. C. and Porter, R. W. (1990) Diurnal variations in spinal mechanics and their clinical significance. *J. Bone Jt Surg.* **72B**, 266–270.

Adams, M. A., Dolan, P., Marx, C. and Hutton, W. C. (1986) An electronic inclinometer technique for measuring lumbar curvature. *Clin. Biomech.* **1**, 130–4.

Adams, M. A. and Hutton, W. C. (1982) Prolapsed intervertebral disc. A hyperflexion injury. *Spine* **7**, 184–191.

Adams, M. A. and Hutton, W. C. (1985) Gradual disc prolapse. *Spine* **10**, 524–531.

Adams, M. A. and Hutton, W. C. (1986) Has the lumbar spine a margin of safety in forward bending? *Clin. Biomech.* **1**, 3–6.

Adams, M. A. and Hutton, W. C. (1988) Mechanics of the intervertebral disc. In *The Biology of the Intervertebral Disc* (Edited by Ghosh, P.). CRC Press, Boca Raton, FL.

Adams, M. A., Hutton, W. C. and Stott, J. R. R. (1980) The resistance to flexion of the lumbar intervertebral joint. *Spine* **5**, 245–253.

An, K.-N., Jacobsen, M. C., Berglund, L. J. and Chao, E. Y. S. (1988) Application of a magnetic tracking device to kinesiologic studies. *J. Biomechanics* **21**, 613–620.

Bartelink, D. L. (1957) The role of abdominal pressure in relieving the pressure on the lumbar intervertebral discs. *J. Bone Jt Surg.* **39B**, 718–725.

Brinckmann, P. (1986) Injury of the annulus fibrosus and disc protrusions. An *in vitro* investigation on human lumbar discs. *Spine* **11**, 149–53.

Brinckmann, P., Biggemann, M. and Hilweg, D. (1989) Prediction of the compressive strength of human lumbar vertebrae. *Clin. Biomech.* **4**, Suppl. 2.

Chaffin, D. B. and Park, K. S. (1973) A longitudinal study of low-back pain as associated with occupational weight lifting factors. *Am. Ind. Hygiene Ass. J.* **34**, 513–525.

Dolan, P. and Adams, M. A. (1990) The influence of lumbar and hip mobility on the bending moment acting on the lumbar spine. (Submitted for publication.)

Frymoyer, J. W., Pope, M. H., Clements, J. H., Wilder, D. G., MacPhearson, B. and Ashikaga, T. (1983) Risk factors in low back pain. *J. Bone Jt Surg.* **65A**, 213–218.

Hardy, W. G., Lissner, H. R., Webster, J. E. and Gurdjian, E. S. (1958) Repeated loading tests of the lumbar spine: a preliminary report. *Surg. Forum* **9**, 690–695.

Hasberry, S. and Pearcy, M. J. (1986) Temperature dependence of the tensile properties of interspinous ligaments of sheep. *J. biomed. Engng* **8**, 62–66.

Janevic, J., Ashton-Miller, J. A. and Schultz, A. B. (1989) Large compressive preloads decrease lumbar motion segment flexibility. Int. Soc. Study Lumbar Spine, Kyoto, Japan, 15–19 May 1989.

Jonck, L. M., Niekerk, J. M. van (1961) A roentgenological study of the motion of lumbar spine of the Bantu. *S. Afr. J. Lab. Clin. Med.* **7**, 67–71.

Kelsey, J. L., Githens, P. B., White, A. A., Holford, T. R. *et al.* (1984) An epidemiologic study of lifting and twisting on the job and risk for acute prolapsed lumbar intervertebral disc. *J. Orthop. Res.* **2**, 61–66.

King Liu, Y., Njus, G., Buckwalter, J. and Wakano, K. (1983) Fatigue response of lumbar intervertebral joints under axial cyclic loading. *Spine* **8**, 857–865.

Lin, H. S., Liu, Y. K. and Adams, K. H. (1978) Mechanical response of the lumbar intervertebral joint under physiological (complex) loading. *J. Bone Jt Surg.* **60A**, 41–55.

Markolf, K. L. (1972) Deformation of the thoracolumbar intervertebral joints in response to external loads. *J. Bone Jt Surg.* **54A**, 511–33.

McGill, S. M. (1988) Estimation of force and extensor moment contributions of the disc and ligaments at L4–L5. *Spine* **13**, 1395–1402.

McGill, S. M. and Norman, R. W. (1987) Effects of an anatomically detailed erector spinae model on L4/L5 disc compression and shear. *J. Biomechanics* **20**, 591–600.

Miller, J. A. A., Schultz, A. B., Warwick, D. N. and Spencer, D. L. (1986) Mechanical properties of lumbar spine motion segments under large loads. *J. Biomechanics* **19**, 79–84.

Nachemson, A. (1981) Disc pressure measurements. *Spine* **6**, 93–97.

National Institute of Occupational Safety and Health. (1980) A work practices guide for manual lifting. U.S. Dept of HEW, NIOSH, Cincinnati.

Ortengren, R., Andersson, G. B. J. and Nachemson, A. (1981) Studies of relationships between lumbar disc pressure, myoelectric back muscle activity, and intra-abdominal (intra-gastric) pressure. *Spine* **6**, 98–103.

Panjabi, M. M., Krag, M., Summers, D. and Videman, T. (1985) Biomechanical time-tolerance of fresh cadaveric human spine specimens. *J. Orthop. Res.* **3**, 292–300.

Panjabi, M. M., Krag, M. H., White, A. A. and Southwick, W. (1977) Effects of preload on load displacement curves of the lumbar spine. *Orthop. Clins North Am.* **8**, 181–192.

Pearcy, M. J. and Bogduk, N. (1988) Instantaneous axes of rotation of the lumbar intervertebral joints. *Spine* **13**, 1033–1041.

Pearcy, M. J. and Hindle, R. J. (1989) New method for the non-invasive three-dimensional measurement of human back movement. *Clin. Biomech.* **4**, 73–79.

Pearcy, M., Portek, I. and Shepherd, J. (1984) Three-dimensional X-ray analysis of normal movement in the lumbar spine. *Spine* **9**, 294–297.

Perey, O. (1957) Fracture of the vertebral endplate. A biomechanical investigation. *Acta orthop. scand.* Suppl. 25.

Schultz, A., Andersson, G., Ortengren, R., Haderspeck, K. and Nachemson, A. (1982) Loads on the lumbar spine. *J. Bone Jt Surg.* **64A**, 713–20.

Schultz, A. B., Warwick, D. N., Berkson, M. H. and Nachemson, A. L. (1979) Mechanical properties of human lumbar spine segments. Part 1. Response in flexion, extension, lateral bending and torsion. *J. biomech. Engng* **101**, 46–52.

Shirazi-Adl, A. (1989) Strain in fibers of a lumbar disc. Analysis of the role of lifting in producing disc prolapse. *Spine* **14**, 96–103.

Smeathers, J. E. and Joanes, D. N. (1988) Dynamic compressive properties of human lumbar intervertebral joints: a comparison between fresh and thawed specimens. *J. Biomechanics* **21**, 425–433.

Svensson, H. O. and Andersson, G. B. J. (1983) Low back pain in 40-to-47-year-old men: work history and work environment factors. *Spine* **8**, 272–276.

Troup, J. D. G. (1965) Relation of lumbar spine disorders to heavy manual work and lifting. *Lancet* **1**, 857–861.

Troup, J. D. G., Martin, J. W. and Lloyd, D. C. E. F. (1981) Back pain in industry. *Spine* **6**, 61–69.