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Influence of lumbar and hip mobility on the bending stresses acting on the lumbar spine

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Summary

Bending and lifting activities are associated with injury to the lumbar discs and ligaments, and cadaveric experiments suggest that this damage is most attributable to a high bending moment (bending stress) acting on the osteoligamentous spine. We examined the hypothesis that people with poor sagittal mobility in the lumbar spine and hips apply higher bending stresses to their spines during everyday lifting activities.

Forty-nine subjects performed a series of simple forward bending and lifting exercises while their lumbar flexion was measured continuously using a skin-surface technique (3-SPACE ISOTRAK). Peak flexion angles were compared with the bending properties of cadaveric osteoligamentous spines in order to calculate the peak bending moment (bending stress) acting on the lumbar spine during each exercise.

All subjects flattened or reversed their lumbar lordosis when lifting, and most came close to or exceeded their static *in vivo* limit of lumbar flexion in many of the activities. The bending moment acting on the lumbosacral junction rose to about 30 Nm, which is about 50% of that required to cause injury in a single lift. Bending moments were significantly lower in subjects who had good sagittal mobility in the lumbar spine. Good hip mobility was similarly associated with a reduction in bending moment, but this reached significance only in subjects who reported a history of low back pain.

Relevance

This study shows that the lumbar spine is commonly subjected to substantial bending stresses during normal everyday activities. Bending stresses are higher in people with poor mobility in the lumbar spine and hips, suggesting that 'stiff' people are at greater risk of injuring their backs during bending and lifting activities.

Key words: Bending moment, lumbar spine, *in vivo*, mobility

Introduction

The lumbar spine offers little resistance to bending over much of its range of movement, but the bending stress (or, more correctly, bending 'moment') rises rapidly as the elastic limits of flexion and extension are approached (Figure 1). The region of low bending moment comprises a fairly constant proportion of the full range of movement[1] and so is much larger in a supple spine. This suggests that an extremely supple person can probably touch his toes without generating high bending stresses, whereas a very stiff person will be able to flex only a short distance before the bending moment begins to rise to high levels.

The possibility has serious implications because bending is potentially harmful to the intervertebral discs and ligaments of the lumbar spine. Cadaveric experiments have shown that bending can sprain the ligaments of the neural arch[3] and that a combination of bending and compression can cause the intervertebral discs to prolapse[4–6]. Compression and torsion, without any component of bending, damage the vertebrae first before the soft tissues[7–12]. Mathematical models support these experimental findings[13]. A recent epidemiological survey has demonstrated the clinical relevance of this work by showing a close association between mechanical (over)loading of the lumbar spine and the incidence of acute disc problems[14].

The purpose of the present investigation was to quantify the bending moment acting on the

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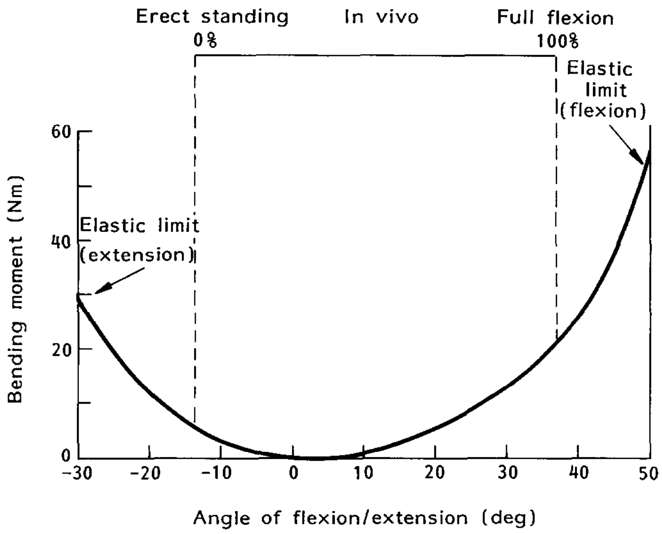

Figure 1. Diagrammatic representation of how the bending moment acting on the osteoligamentous lumbar spine varies across the full range of flexion and extension. Negative angles indicate lumbar extension relative to the unloaded spine (0°). The bending moment increases rapidly near the limits of motion, and non-recoverable deformation (injury) occurs just after. (Based on data from experiments on cadaveric lumbar motion segments[1,2].) For comparison, a typical *in-vivo* range of lumbar flexion is shown at the top of the figure.

osteoligamentous lumbar spine *in vivo*, and to test the hypothesis that it rises to higher levels in people with poor sagittal mobility in the lumbar spine and hips. Use was made of a recently developed technique for quantifying bending moment from measurements of lumbar flexion and lumbar mobility[1].

Materials and methods

Subjects

The study group comprised 49 people, most of whom were employed in light manual or office work. All subjects were asymptomatic at the time of testing although sixteen of them had previously suffered from low back pain or sciatica that required medical attention. In view of the possibility that previous back problems might have some bearing on the results, these subjects were considered as a separate group; however, it was not the purpose of this investigation to draw comparisons between specific patient groups. Details of age and sex are given in Table 3. The mean age of subjects with and without a history of low back pain was 38.7 (SD, 13.7) and 32.1 (SD, 8.0) respectively.

Informed consent was obtained from all subjects, but the purpose of the investigation was not revealed to them.

Measurement of lumbar flexion

The lumbar curvature, as defined here, is calculated by measuring the angle made between the tangents to the skin surface at L_1 and S_1 in the sagittal plane (θ in

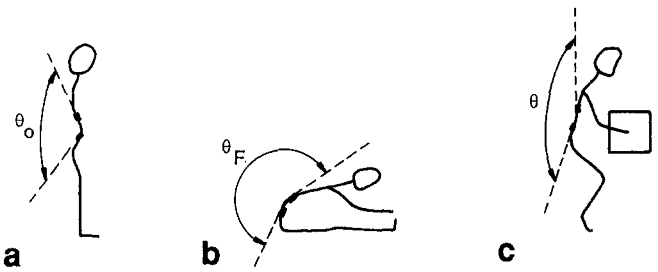

Figure 2. Lumbar flexion *in vivo* measured from changes in the curvature of the surface of the back. **a** Erect standing; **b** fully flexed; **c** general. Flexion is expressed as a percentage of the full range of lumbar flexion. Lumbar curvature = $\theta - 180°$; lumbar flexion = $\theta - \theta_0$ deg. $\equiv (\theta-\theta_0)/(\theta_F-\theta_0)*100\%$ of full flexion

Figure 2) and subtracting 180°. Lumbar flexion was measured from changes in this curvature, as shown in Figure 2. The curvature in erect standing was taken to represent 'zero flexion' because it was found to be more reproducible than the curvature in full extension.

Lumbar curvature was measured using the 3-SPACE ISOTRAK system: 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 surface overlying the spinous process of L_1. The signal from the sensor was fed to a systems electronic unit which calculated the angle between the source and sensor in the sagittal plane, at a frequency of 28 Hz. The results were stored in a microcomputer. The curvature in erect standing was subtracted and the resulting values of lumbar flexion stored for subsequent analysis. A method was devised to attach the ISOTRAK to the skin in a way that eliminated the large systematic errors reported by previous users of the ISOTRAK[15]. This involved mounting the source and sensor on the back with the wires aligned horizontally so that they did not move appreciably during flexion movements (Figure 3).

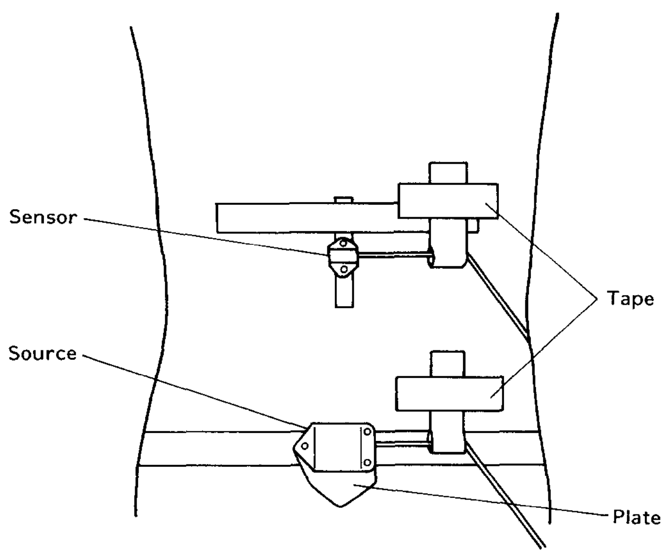

Figure 3. Posterior view of the ISOTRAK system attached to the skin overlying the spinous processes of L_1 and S_1.

We have previously found good correlation ($r = 0.91$) between flexion angles obtained from X-rays and measurements of lumbar flexion obtained with similar skin-mounted inclinometers[16]. Errors were large only in full extension, and when the subject was obese. Skin-surface techniques for measuring lumbar flexion are not very accurate when large hand-held inclinometers are used[17,18] or when the skin surface and X-ray measurements are made on different occasions in different postures as in the study by Stokes et al.[19].

Calculation of bending moment

Bending moment was quantified using a recently developed technique which involves comparing *in-vivo* measurements of lumbar flexion with the bending properties of cadaveric osteoligamentous spines[1]. It was necessary to express bending moments and flexion angles as a percentage of their values at the elastic limit in order to obtain a relationship between the two variables that was independent of the body mass, mobility and age of the specimens. It was also necessary to establish a linear relationship between '%flexion' measured *in vivo* and '%flexion' measured on a cadaveric specimen, because the scales are different: erect standing involves some backwards bending of the osteoligamentous spine, and full flexion *in vivo* does not bend the spine right up to its elastic limit (see Figure 1).

The technique measures bending moment *in vivo* with an accuracy of about ±8% of its value at the elastic limit. For an individual of average body mass, this is equivalent to about ±5 Nm at the lumbosacral junction, since the strength in bending of L_5–S_1 motion segments is 61 Nm on average[1].

Experimental procedures

The L_1 and S_1 spinous processes were located by palpation, and the ISOTRAK sensor and source were attached.

Initially each subject's static range of lumbar flexion was determined by measuring the lumbar curvature in the static erect standing and extreme toe-touching positions. In the erect standing position, subjects were instructed to stand straight but not to attention and to keep their line of sight level. In the fully flexed posture, subjects sat on the floor with their legs apart, knees straight, and hands behind the neck. They were verbally encouraged to make a strenuous effort to touch the floor with their forehead. (Maximum effort was found to be more reproducible than submaximal efforts.)

Hip mobility was also assessed for each subject. The sensor was attached to a vertical surface positioned close to the source, which remained on S_1. The angle between source and sensor was then measured with the subject standing in the straight-legged toe-touching position. This angle, the 'sacral inclination in full flexion' was recorded as an overall measure of hip mobility and hamstring tightness.

Table 1. The seven bending and lifting activities performed by the subjects during the experiment

1	Sitting down on the floor and then standing up.
2	Putting on a sock while seated.
3	Picking up a pen from the floor.
4	Lifting a small box weighing 3 kg.
5	Reaching forward and to one side to lift a small box weighing 3 kg.
6	Lifting a compact 10-kg weight.
7	Lifting a large box weighing 10 kg.

Once the subjects' mobility had been established, they were asked to perform the series of exercises listed in Table 1, while the ISOTRAK recorded lumbar curvatures at 28 Hz. The heaviest load lifted was 10 kg. Subjects were given no specific instructions regarding lifting technique but were asked to perform the tasks 'in any way that seemed natural'.

Lumbar and hip mobility were measured again after the exercises in order to assess their repeatability. Two subjects were also tested once a week for 8 and 9 consecutive weeks respectively, to assess the overall reproducibility of the methods.

Statistical analysis

Subjects with a history of low back pain ('history lbp') were considered separately in order to take account of any differences in mobility or lifting technique that may have resulted from their previous condition. Relationships between continuous variables were assessed using single or stepwise linear regression, while differences between the 'history lbp' and 'no history' subjects were assessed using group t-tests.

Results

The results of the reproducibility study on two people are shown in Table 2. The lumbar curvature in erect standing and full flexion was reproducible, with a standard deviation of ±2.5°. These values include all the variation due to possible inaccuracies in palpation, and changes in subject performance, as well as error in the ISOTRAK measurements. In one of the subjects (male aged 37) the range of lumbar flexion had previously been established as 55° by X-ray measurements. The values obtained with the ISOTRAK

Table 2. Reproducibility of lumbar curvature measurements made in two subjects who were tested on separate occasions. Values given are the mean (SD)

Subject	*Lumbar curvature*		*Range of lumbar flexion*
	Erect standing	*Full flexion*	
M 37 ($n = 9$)	−34.7 (2.0)	21.7 (1.2)	56.4 (2.6)
F 33 ($n = 8$)	−32.9 (2.5)	25.9 (1.4)	58.8 (3.0)

(56.4, SD, 2.6°, n = 9) were not significantly different.

Measurements of the (static) range of lumbar flexion made on each of the 49 subjects, before and after the exercises, showed no significant difference in a matched pair comparison (mean difference, –0.1°, SD, 3.3°).

Neither lumbar nor hip mobility varied significantly with age in our sample, probably because most of the subjects were aged between 25 and 36 years.

In the 'history lbp' group, lumbar mobility was reduced by about 7° on average ($P<0.05$) and hip mobility by 20° ($P<0.001$) compared to the 'no history' group. In the 'no history' group, hip mobility was slightly greater in women than in men ($P<0.01$).

During the exercises all subjects flattened or reversed their lumbar lordosis, even when lifting with the knees flexed (Figure 4). A typical dynamic recording from the ISOTRAK (Figure 5) shows the peaks in lumbar flexion corresponding to five exercises. It is apparent that this subject consistently flexed his lumbar spine close to, or even beyond, the limit of his static range of flexion. In general, 'stiff' subjects with low lumbar mobility flexed closer to (or further beyond) this limit than did more supple people (Figure 6). This was equally true of subjects in the 'no history' and 'history lbp' groups.

The differences between subjects appear slight when their peak flexion is expressed as degrees short of, or beyond, their static range. However, as the limit of flexion is approached, small changes in flexion angle imply large changes in bending moment acting on the spine (Figure 1). Consequently, the stiff subjects applied much higher peak bending moments to their lumbar spine than did the supple subjects (Figure 7). Maximum bending moment was increased by about 100% in the least flexible subjects, compared to the most flexible.

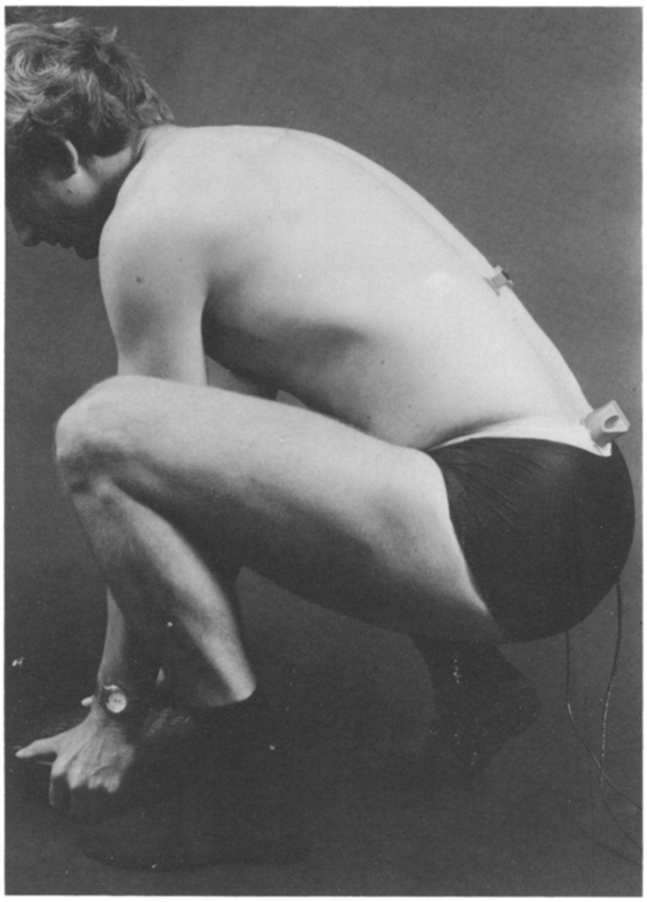

Figure 4. Lateral view of the ISOTRAK system attached to a subject's back, showing how the lumbar curvature is reversed during lifting.

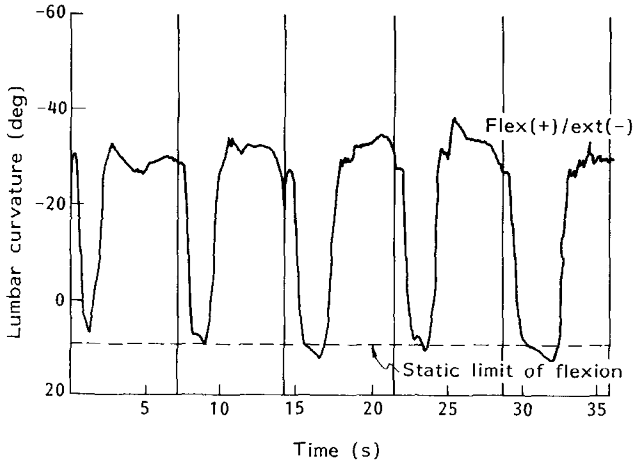

Figure 5. Raw data from the ISOTRAK. The trace shows flexion/extension during each of five different lifts (activities 3 – 7 respectively in Table 1). The limit of the subject's static range of lumbar flexion is indicated by the horizontal dashed line. The vertical lines indicate where one lift finishes and the next begins. The time scale is not continuous as subjects were allowed a short recovery time between lifts.

The peak bending moment calculated for each subject in each activity is shown in Table 3. These detailed results show differences between the two groups of subjects. Among those without a history of low back pain there was a consistent tendency for the bending moment to be higher in stiff people, even during fairly innocuous tasks such as putting on socks or picking up a pen from the floor. Consequently, the peak bending moment was significantly related to lumbar mobility even when values were averaged over the seven activities ($P<0.001$). Also, in this group, peak bending moment increased with the size and weight of the object lifted ($P<0.001$). Among those with a history of low back pain, neither of these trends was apparent, and the average value of peak bending moment during the five lifts (activities 3–7 in Table 1) was significantly lower than in the 'no history' group ($P<0.05$).

Hip mobility, as expressed by the sacral inclination in full flexion, had a less pronounced effect on the bending moment and this only reached significance in the 'history lbp' group (Figure 8).

Stepwise multiple regression showed that age, as well as the range of lumbar flexion, was a significant predictor of peak bending moment, but its influence was comparatively small. These two variables accounted for 84% of the variation in peak bending moment for the 'no history' group, and 70% of the variation for the 'history lbp' group.

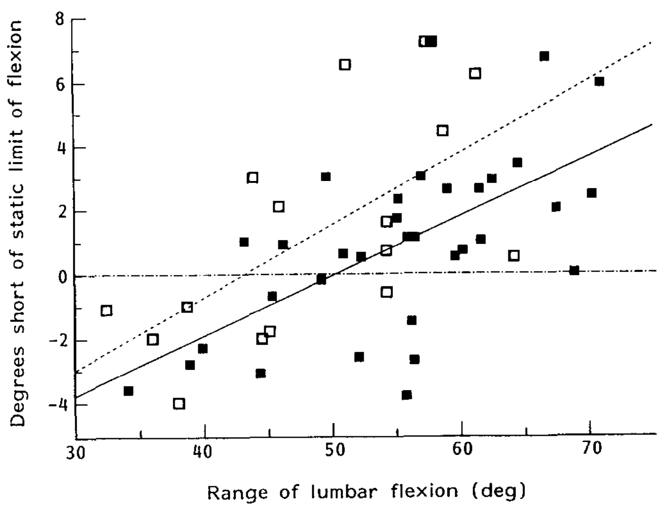

Figure 6. Subjects with a large range of lumbar flexion stayed several degrees short of their static limit during the exercises. The least supple subjects often exceeded this limit (negative values). Flexion values refer to the greatest value attained by each subject in any of the seven activities. The gradient of the linear regression line was significantly greater than zero in both groups of subjects ($P<0.001$). ——■—— No history, $R = 0.612$; --□-- history lbp, $R = 0.631$.

Discussion

The mobility of the present group of subjects with no history of low back pain was similar to that reported in the literature for similarity aged groups of healthy individuals, assessed using X-ray measurements[4,20] and other skin surface techniques[21-23].

In the 'history lbp' group, mobility was reduced more in the hip than the spine, and this may explain why hip mobility was an important determinant of the bending moment acting on the spine in this group but not in the 'no history' group. The other main difference between the two groups was that those with a history of low back pain applied significantly lower bending moments to their spines when the average values for the five lifting activities were compared with those of the 'no history' group. This may be because they tended to lift more carefully, and with a straighter back, whenever the demands of the task allowed them to.

Every one of our subjects flexed the lumbar spine to within a few degrees of the static limit during routine bending and lifting activities. A similar result can be inferred from the experiments of Davis et al.[24] but it does not appear to have been widely recognized, and most studies of spinal loading consider compressive forces only and simply ignore the effects of bending. In this study, bending moments frequently rose to 40–50% of that required to cause damage to a cadaveric spine in a single loading cycle (Table 3), suggesting that bending might well be responsible for fatigue injury to discs and ligaments in life. Little is known, however, about the long-term fatigue life of these tissues. The risk of injury would probably be greater in the early morning when the discs are swollen with fluid and have an increased bending stiffness[25]. Also, rapid movements and lifting heavy weights might increase the risks further.

The main result of this study clearly supports our hypothesis: bending stresses on the lumbar spine are indeed increased in subjects with poor spinal mobility, and by up to 100%. It must be admitted that the quantitative relationship between mobility and bending moment shown in Figure 7 pushes the technique for measuring bending moment to its limit. It requires that the relationship between '% bending moment' and '% flexion' is the same for the most supple and least supple spines. Whilst this was true of the 42 cadaveric spinal specimens tested previously[1] there may be exceptions to the rule in the general population. However, the fact that the least mobile subjects approached their limits of flexion more closely than the others (Figure 6) indicates that they must have applied higher bending moments to their spines. The precise relationship between flexion and bending is not critical in establishing this trend, although it does have a considerable effect on the magnitude of the inferred increase in bending moment.

It could be argued that stiff people have shorter back muscles which do not permit them to approach the spine's elastic limit as nearly as supple people do. However, there is evidence that the *in-vivo* limit of lumbar flexion is determined largely by the ligaments and discs rather than the muscles. Intervertebral flexion movements measured in X-ray studies of healthy people show a regular decrease in mobility with increasing age and at higher lumbar levels[4,26]. The same pattern of mobility, and the same amount of variability, are shown by cadaveric lumbar motion segments[27]. There is no reason to suppose that the low

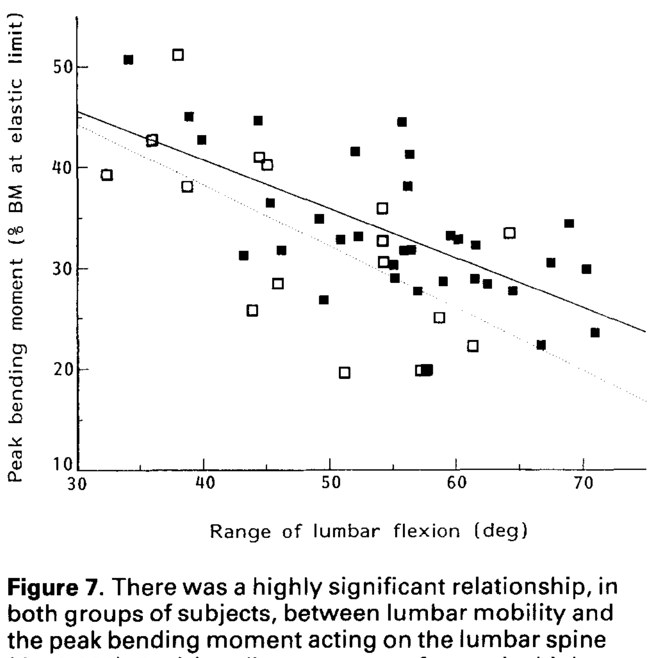

Figure 7. There was a highly significant relationship, in both groups of subjects, between lumbar mobility and the peak bending moment acting on the lumbar spine ($P<0.001$). Peak bending moment refers to the highest value attained by each subject in any of the seven activities listed in Table 1, and is expressed as a percentage of the value at the elastic limit of flexion (100%). ——■—— No history, $R = 0.644$; --□-- history lbp, $R = 0.650$.

Table 3. Peak bending moment acting on the lumbar spine in each of the seven activities listed in Table 1. Subjects in the 'no history' group are shown above the dashed line; the 'history lbp' group are shown below it. Bending moment is expressed as a percentage of the value at the elastic limit. RoF, range of lumbar flexion; FF, full flexion

Subject	*Lumbar RoF (deg)*	*Sacral angle (FF) (deg)*	*Peak bending moment in each of the seven activities*						
			1	*2*	*3*	*4*	*5*	*6*	*7*
F33	56.5	97.5	22	32	11	16	23	21	30
M36	55.2	92.9	16	29	14	20	28	22	26
F32	46.2	96.1	24	17	26	28	29	32	29
F35	45.3	76.9	28	31	28	35	35	36	31
M30	39.9	71.9	26	25	24	31	40	36	43
M36	59.0	91.3	12	28	12	25	29	22	28
F33	59.6	97.0	33	26	24	32	27	29	32
M28	38.9	58.6	32	43	33	34	35	39	45
F21	52.1	88.6	32	41	31	34	32	38	36
F22	55.9	116.8	28	30	22	30	31	30	32
F37	55.1	115.2	23	29	20	30	27	29	29
F36	49.6	106.4	20	25	21	27	20	22	25
M25	57.0	89.0	23	28	6	14	18	11	23
M28	70.3	74.0	29	12	11	24	30	29	22
M40	67.5	82.5	22	24	27	30	26	30	22
M36	64.5	105.4	19	24	16	25	26	24	28
F33	52.3	97.7	26	25	18	31	32	28	33
M36	61.5	107.4	18	23	15	24	29	22	26
F19	44.4	65.6	34	31	33	38	42	45	44
F25	66.7	103.2	9	6	18	22	19	18	16
M45	49.2	66.8	31	32	23	30	30	29	35
M36	61.6	108.1	20	22	17	27	28	27	32
M22	60.2	86.6	21	23	19	28	29	29	33
F34	34.1	109.4	34	17	19	30	51	41	49
F44	57.7	116.5	17	10	10	12	20	10	16
F27	56.4	131.8	34	41	31	35	34	40	38
M33	62.5	85.6	11	12	24	27	27	28	28
M35	56.2	88.8	27	11	28	32	33	35	38
M58	43.2	75.8	28	22	23	31	25	29	31
F27	55.8	77.0	24	27	37	42	43	44	43
F21	50.9	109.8	20	27	7	33	26	27	22
M27	68.9	95.0	24	26	32	31	34	28	17
M30	71.0	63.5	23	21	8	20	17	17	24
F29	51.2	98.0	17	13	10	17	14	20	18
M33	44.5		28	32	26	32	22	24	41
M45	38.7	77.6	37	38	10	13	9	10	26
M43	38.0	67.5	51	32	13	15	11	12	16
M33	58.7	63.1	23	25	12	25	22	24	25
M28	61.3	81.5	22	19	20	17	10	7	20
F25	32.4	94.7	36	39	8	12	14	13	18
F57	45.1	72.3	38	40	18	23	29	35	29
F25	54.2	52.0	30	25	31	34	36	33	30
F59	45.9	86.7	22	25	17	19	28	25	22
F38	36.0		38	30	32	36	40	34	43
M60	43.9	60.7	26	23	20	20	15	16	19
M61	54.2	80.9	22	9	21	27	27	33	27
M23	64.2	48.0	32	24	32	31	32	29	33
F30	57.6	68.5	16	17	12	13	15	14	20
M30	54.3	65.3	26	24	24	30	29	26	29
Mean	53.2	86.5	25.6	25.2	20.3	26.4	27.2	26.6	29.0
SD	9.7	18.8	7.8	8.6	8.1	7.4	8.8	9.1	8.4

levels of spinal mobility in old people are caused by anything other than changes in the osteoligamentous spine, and the same is probably true of young people as well.

There is conflicting evidence in the literature concerning links between mobility on the one hand, and low back injury and back pain on the other. Injury and pain are themselves poorly differentiated in many epidemiological studies, especially when the condition is self-reported, and the former is sometimes inferred from the presence of the latter. For this reason, we will refer to studies concerning both injury and pain.

Poor hip mobility is associated with a greater severity of low back pain[28], although the relationship may not

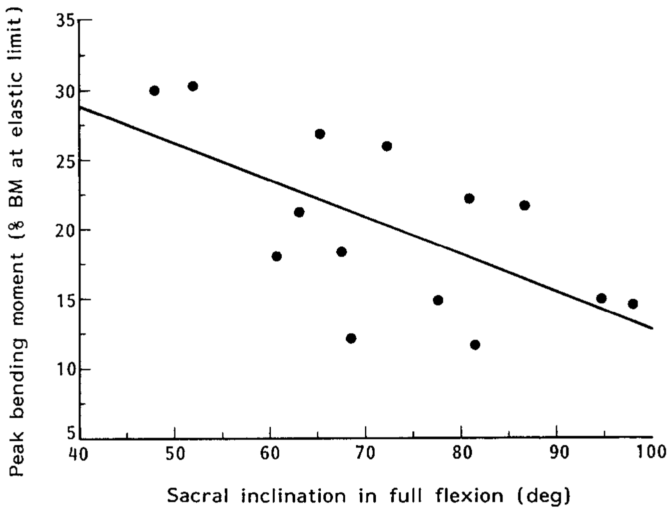

Figure 8. Sacral inclination in full flexion is a measure of the amount of straight-legged hip flexion permitted by the hamstrings in the toe-touching position. In the 'history lbp' group, it is inversely related to the average peak bending moment measured during the seven activities listed in Table 1 ($P<0.02$; data was lost on two subjects, giving $n=14$). Bending moment is expressed as a percentage of the value at the elastic limit of flexion (100%). —●— History lbp, $R=0.629$.

be causal. Links with spinal mobility are more difficult to establish. Comparisons between spinal mobility and the prevalence of current or previous back pain are ambiguous because a painful back may be more mobile on account of some ligamentous instability[29–31] or less mobile because of pain aggravation, real or anticipated[32–34]. Furthermore, pain reduction during a course of manipulative therapy can be accompanied by either an increase or a decrease in lumbar mobility[35]. Studies of people at the extreme ends of the mobility scale may be misleading because extreme spinal mobility may be just one manifestation of some general (and painful) 'hypermobility syndrome'[36] or it may be associated with activities such as gymnastics, which often injure the back[37].

A prospective survey by Biering-Sørensen[38] suggested that men with good spinal mobility (as measured by the 'modified Schober' skin-stretching test) were more likely to suffer low back pain for the first time in the following year. The opposite trend was found for women, although this did not reach significance. Battié et al.[39] found that mobility had no predictive value for either sex. These conflicting results may be attributable to the inadequacies of the Schober test in general[40] and to the poor correlation between Schober measurements and the true angular movements of vertebrae in particular[18]. An early study by Macrae and Wright[41] has been used to justify the use of the modified Schober technique, but an unspecified number of their subjects were suffering from ankylosing spondylitis and showed very little movement at all in the lumbar spine. Only three subjects were able to flex more than 30°, and these did not show any apparent relationship between Schober measurements and angular movements of the vertebrae. Consequently, prospective surveys using Schober measurements are not competent to predict whether or not poor spinal mobility increases the risk of back injury. The results of the present study provide indirect evidence to suggest that it does.

Conclusion

Poor mobility in the lumbar spine and hips increases the bending moment acting on the lumbar spine during forward bending and lifting activities. This may lead to an increased risk of injury to the intervertebral discs and ligaments.

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References

1 Adams MA, Dolan P. A technique for quantifying bending moment acting on the lumbar spine in-vivo. (European Society of Biomechanics Award Paper 1990). *J Biomech* 1991; 24(2): 117–26
2 Adams MA, Dolan P, Hutton WC. The lumbar spine in backward bending. *Spine* 1988; 13: 1019–26
3 Adams MA, Hutton WC, Stott JRR. The resistance to flexion of the lumbar intervertebral joint. *Spine* 1980; 5: 245–53
4 Adams MA, Hutton WC. Prolapsed intervertebral disc. A hyperflexion injury. *Spine* 1982; 7: 184–91
5 Adams MA, Hutton WC. The effect of fatigue on the lumbar intervertebral disc. *J Bone Joint Surg* 1983; 65-B: 199–203
6 Adams MA, Hutton WC. Gradual disc prolapse. *Spine* 1985; 10: 524–31
7 Adams MA, Hutton WC. The relevance of torsion to the mechanical derangement of the lumbar spine. *Spine* 1981; 6: 241–8
8 Ahmed AM, Duncan NA, Burke DL. The effect of facet geometry on the axial torque–rotation response of lumbar motion segments. *Spine* 1990; 15: 391–401
9 Brinckmann P, Biggemann, M, Hilweg D. Prediction of the compressive strength of human lumbar vertebrae. *Clin Biomech* 1989; 4: [Suppl. 2]
10 Brinckmann P. Injury of the annulus fibrosus and disc protrusions. An in-vitro investigation on human lumbar discs. *Spine* 1986; 11: 149–53
11 Liu YK, Goel VK, Dejong A et al. Torsional fatigue of the lumbar intervertebral joints. *Spine* 1985; 10: 894–901
12 Perey O. Fracture of the vertebral endplate. A biomechanical investigation. *Acta Orthop Scand* 1957; (Suppl. 25)
13 Shirazi-Adl A. Strain in fibers of a lumbar disc. Analysis of the role of lifting in producing disc prolapse. *Spine* 1989; 14: 96–103
14 Kelsey JL, Githens PB, White AA et al. An epidemiologic study of lifting and twisting on the job and risk for acute prolapsed intervertebral disc. *J Orthop Res* 1984; 2: 61–6
15 Pearcy MJ, Hindle RJ. New method for the non-invasive three-dimensional measurement of human back movement. *Clin Biomech* 1989; 2: 73–9
16 Adams MA, Dolan P, Marx C, Hutton WC. An electronic inclinometer technique for measuring lumbar curvature. *Clin Biomech* 1986; 1: 130–4

17 Gill K, Krag MH, Johnson GB, Haugh LD, Pope MH. Repeatability of four clinical methods for assessment of lumbar spinal motion. *Spine* 1988; 13: 50–3
18 Portek I, Pearcy MJ, Reader GP, Mowat AG. Correlation between radiographic and clinical measurement of lumbar spine movement. *Br J Rheumatol* 1983; 22: 197–205
19 Stokes IAF, Bevins TM, Lunn RA. Back surface curvature and measurement of lumbar spine motion. *Spine* 1987; 12: 355–61
20 Pearcy M, Portek I, Shepherd J. Three-dimensional X-ray analysis of normal movement in the lumbar spine. *Spine* 1984; 9: 294–7
21 Adams MA, Dolan P, Hutton WC. Diurnal variations in the stresses on the lumbar spine. *Spine* 1987; 12: 130–7
22 Loebl WY. Measurement of spinal posture and range of spinal movement. *Ann Phys Med* 1967; 9(3): 103–10
23 Troup JDG, Hood CA, Chapman AE. Measurements of the sagittal mobility of the lumbar spine and hips. *Ann Phys Med* 1968; 9: 308–21
24 Davis PR, Troup JDG, Burnard JH. Movements of the thoracic and lumbar spine when lifting: a chrono-cyclophotographic study. *J Anat* 1965; 99(1): 13–26
25 Adams MA, Dolan P, Hutton WC, Porter RW. Diurnal variations in spinal mechanics and their clinical significance. *J Bone Joint Surg* 1990; 72-B: 266–70
26 Jonck LM, Van Niekerk JM. A roentgenological study of the motion of the lumbar spine of the Bantu. *SA J Lab Clin Med* 1961; 67–71
27 Adams MA, Hutton WC. Has the lumbar spine a margin of safety in forward bending? *Clin Biomech* 1986; 1: 3–6
28 Mellin G. Correlations of hip mobility with degree of back pain and lumbar spinal mobility in chronic low-back pain patients. *Spine* 1988; 13: 668–70
29 Howes RG, Isdale IC. The loose back – an unrecognised syndrome. *Rheumatol Phys Med* 1971; 11: 72–7
30 Lankhorst GJ, Van de Stadt RJ, Van der Korst JK. The natural history of idiopathic low back pain: a 3 year follow-up study of spinal motion, pain and functional capacity. *Scand J Rehabil Med* 1985; 17(1): 1–4
31 Stokes IAF, Wilder DG, Frymoyer JW, Pope MH. Assessment of patients with low back pain by biplanar radiographic measurement of intervertebral motion. *Spine* 1981; 6: 233–40
32 Burton AK, Tillotson KM, Troup JDG. Variation in lumbar sagittal mobility with low back trouble. *Spine* 1989; 14: 584–90
33 Mayer TG, Tencer AF, Kristoferson S, Mooney V. Use of noninvasive techniques for quantification of spinal range-of-motion in normal subjects and chronic low-back dysfunction patients. *Spine* 1984; 8: 588–95
34 Pearcy M, Portek I, Shepherd J. The effect of low back pain on lumbar spinal movements measured by three-dimensional X-ray analysis. *Spine* 1985; 10: 150–3
35 Burton AK, Tillotson KM, Edwards VA, Sykes DA. Lumbar sagittal mobility and low back symptoms in patients treated with manipulation. *J Spinal Disord* 1990; 3: 262–8
36 Grahame R. The hypermobility syndrome. *Ann Rheum Dis* 1990; 49: 190–200
37 Ohlen G, Wredmark T, Spangfort E. Spinal sagittal configuration and mobility related to low-back pain in the female gymnast. *Spine* 1989; 14: 847–50
38 Biering-Sørensen F. Physical measurements as risk indicators for low back trouble over a one-year period. *Spine* 1984; 9: 106–18
39 Battié MC, Bigos SJ, Fisher LD et al. The role of spinal flexibility in back pain complaints within industry: a prospective study. *Spine* 1990; 15: 768–73
40 Miller SA, Mayer T, Cox R, Gatchel RJ. Reliability problems associated with the modified Schober technique for true lumbar flexion measurement. *Spine* 1992; 17: 345–8
41 Macrae IF, Wright V. Measurement of back movement. *Ann Rheum Dis* 1969; 28: 584–9