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PASSIVE TISSUES HELP THE BACK MUSCLES TO GENERATE EXTENSOR MOMENTS DURING LIFTING

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Abstract—We examined the possibility that passive tissues can help the erector spinae to generate large extensor moments during lifting. One hundred and forty-nine healthy men and women participated in the study. Subjects pulled upwards with steadily increasing force on a floor-mounted load cell, while EMG activity was recorded from electrodes overlying the erector spinae at L3 and T10. Extensor moment was calculated from the load cell data, and was plotted against the full-wave rectified and averaged EMG signal. The relationship was linear with an intercept on the extensor moment axis (I) which indicated the flexion moment resisted by 'passive' (electrically silent) tissues. The dependence of I on lumbar flexion angle was studied by repeating the isometric pulls between 6 and 12 times, with the subject positioned in varying amounts of flexion, as measured by the '3-Space Isotrak'. Subjects then lifted weights of up to 20 kg from the floor, using 'stoop', 'squat' and 'freestyle' techniques, while lumbar flexion and EMG activity were recorded at 28 Hz. The isometric pulls showed that, on average, I increased from 25 Nm in lordotic postures to 120 Nm (for men) and 77 Nm (for women), in full flexion. During the lifts, peak extensor moment was generated with the lumbar spine flexed by 78–97% of the range between erect standing and full flexion. Comparisons with the static calibrations showed that between 16 and 31% of the peak extensor moment generated during lifting was unrelated to EMG activity in the erector spinae. Comparisons with cadaveric data suggested that less than a quarter of this 'passive' extensor moment was due to the intervertebral discs and ligaments.

INTRODUCTION

In stooped postures, the electrical activity of the erector spinae muscles falls to zero leaving the forward bending moment of the upper body to be resisted entirely by passive (electrically silent) structures. This 'flexion–relaxation' phenomenon depends on lumbar flexion rather than the overall angle of the trunk (Andersson *et al.*, 1976; Floyd and Silver, 1955; Kippers and Parker, 1984) and it can persist even when substantial weights are held in the hands (Floyd and Silver, 1955; Kippers and Parker, 1984; Schultz *et al.*, 1985). The origin of the 'passive' extensor moment is unknown, but it may involve the intervertebral disc and ligaments, the iliolumbar ligaments, the lumbodorsal fascia, collagenous tissue within the muscles themselves, and a raised intra-abdominal pressure. The 'passive' moment may also involve remote muscles pulling actively on these structures.

Whatever its origin, a passive extensor moment has the potential to assist the muscles of the lumbar spine during manual handling. When heavy weights are lifted from the ground, the lumbar spine must generate a high extensor moment in order to raise the upper body and weight into the upright position. Calculations based on muscle cross-sectional area suggest that the required forces are beyond the capability of the lumbar extensor muscles alone (McNeill *et al.*, 1980). The thoracic erector spinae can generate an extensor moment about the lumbar spine by means of long tendons lying just underneath the lumbo-dorsal fascia (Bogduk *et al.*, 1992; McGill and Norman, 1987) but even so, there is barely enough active extensor strength in the muscles (McGill *et al.*, 1988).

It appears likely that the muscles do receive assistance from passive tissues, and the extent and origin of this assistance is of practical importance. If a substantial extensor moment is supplied by a structure lying just underneath the skin (for example, the lumbo-dorsal fascia) then that structure has the advantage of a long lever arm about the 'pivot' in the centre of the disc (Pearcy and Bogduk, 1988) and the spine will be subjected to a smaller compressive force than if the same extensor moment had come from the muscles. Similarly, a substantial extensor moment from the intervertebral ligaments would have a high 'cost' in terms of lumbar compression because they lie closer to the pivot than the muscles. Therefore, calculations of spinal compression during manual handling are reliant on accurate information concerning the origins of the extensor moment.

Another reason for wanting to apportion the extensor moment between active and passive tissues is the likelihood that they will respond differently to repetitive or chronic loading: metabolically active muscle will fatigue (Roy *et al.*, 1989) and passive tissues will 'creep' (McGill and Brown, 1992). The highest risk factor so far reported for acute disc prolapse is *frequent* bending and lifting (Kelsey *et al.*, 1984) but the mechanisms leading to this association cannot be satis-

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factorily explored until the role of active and passive tissues in lifting tasks is better understood.

Gracovetsky and Farfan have suggested several mechanisms which might lead to extensor moment generation from the lumbo-dorsal fascia (Gracovetsky and Farfan, 1986; Gracovetsky *et al.*, 1981, 1985) but their theories and calculations lack experimental verification (MacIntosh *et al.*, 1987; McGill and Norman, 1988). McGill and colleagues (1988) have suggested that passive tissue involvement in heavy lifting is not essential and that, in practice, its contribution is small (Potvin *et al.*, 1991). However, the predictions of their mathematical model depend greatly on an assumed function relating flexion angle and passive tissue moment, and this has lead to some inconsistent results (Potvin *et al.*, 1991).

The objectives of the present study were to measure the passive extensor moment during everyday bending and lifting activities, and to indicate its likely origin.

MATERIALS AND METHODS

Experimental design

Extensor moment generation was calibrated against EMG activity in the erector spinae muscles during a series of isometric pulls, each performed with a different amount of lumbar flexion. These calibrations showed how the passive component of extensor moment depended on lumbar flexion. Then, lumbar flexion was measured continuously during dynamic lifting activities, using the '3-Space Isotrak' device, and the peak flexion angles used to calculate the passive extensor moment during each lift. A large-scale study was undertaken to overcome the random errors inherent in skin-surface EMG measurements.

Subjects participating in the study

One hundred and forty nine healthy men and women volunteered for this study. None had any history of severe low back pain. Most were nurses from local hospitals, and had previously received some training in how to lift. The other volunteers were white collar workers at the University and local hospitals. Informed consent was obtained, but the objectives of the study were not revealed and lifting technique was not discussed. Table 1 gives details of the subjects' age, bodymass, height, mobility, and strength in lifting (see below).

A simple analysis of moments

A 'moment arm analysis' of the forces and moments generated during lifting is shown in Fig. 1. The

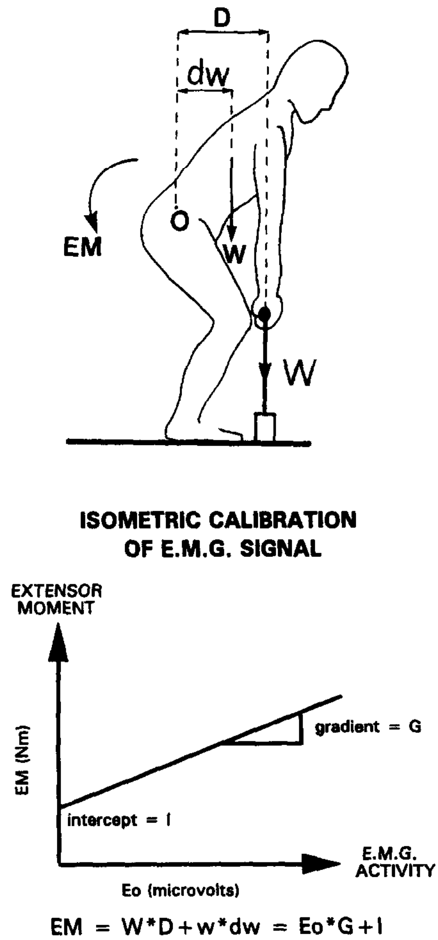

Fig. 1. Upper. A simple sagittal-plane model of forces and moments acting on the lumbar spine when the subject pulls upwards with force *W*. Lower. There is a linear relationship between extensor moment (EM) and EMG activity from the erector spinae muscles (*E*o).

Table 1. Details of the 149 subjects. ROF = range of flexion, and EM_{max} = maximum extensor moment generated during the isometric pull

	Women (n=126)			Men (n=23)		
	Mean	STD	Range	Mean	STD	Range
Age (yr)	27.6	5.8	18–45	30.6	5.3	22–39
Bodymass (kg)	61.9	8.1	46.8–84.2	77.7	9.2	62.4–99.5
Height (cm)	165	6	150–183	179	7	167–193
Lumbar ROF (deg)	56.8	8.9	33.7–74.8	53.3	7.7	38.5–69.4
EM_{max} (Nm)	245	51	145–383	430	116	272–640

assumptions underlying this model have been discussed previously (Dolan and Adams, 1993a). In static equilibrium:

$$\mathrm{EM} = WD + w\,\mathrm{d}w, \tag{1}$$

where EM is the extensor moment, W, the vertical force exerted on load cell and w, the weight of the upper body and arms. D and dw are defined in Fig. 1.

During an isometric contraction, EM is linearly related to the EMG activity (E_0) of the back muscles (Dolan and Adams, 1993a) so that

$$\mathrm{EM} = GE_0 + I, \tag{2}$$

where G is the gradient of the graph and I is the intercept (see Fig. 1).

G and I can be obtained from calibrations as described below, and so EM can be calculated from measured values of E_0. By definition, the active component of EM is zero when $E_0 = 0$, and so the intercept I indicates the passive extensor moment.

Isometric calibrations of EMG activity and extensor moment

The full range of lumbar flexion was established for each subject by measuring lumbar curvature in the erect standing and extreme toe-touching positions using the 3-Space Isotrak device. This consists of a source and sensor of pulsed electromagnetic waves which can be attached to the skin surface overlying the spinous processes of L1 and S1 in order to measure angular movements of the lumbar spine (Adams and Dolan, 1991).

Subjects then stood in a wooden frame with the front of their upper thighs resting against a cushioned crossbar, and with a strap fastened around their hips. Whilst leaning forwards in the frame, the subjects pulled upwards on a handlebar attached by a variable-length chain to a floor-mounted load cell, as shown in Fig. 1. The handlebar was constrained by the frame to move only in the vertical direction. As described previously (Dolan and Adams, 1993a), subjects pulled up on the handlebar with increasing force to reach a 'comfortable' maximum in 3.3 s, whilst the output from the load cell was A–D converted at 60 Hz and input to a microcomputer. The electrical activity of the erector spinae muscles was measured by surface electrodes attached over the belly of the muscle on the left-hand side at T10 and L3. The location of the electrodes was chosen to optimise the linearity between extensor moment and EMG activity, and also to take account of tensile forces generated across the lumbar spine by thoracic regions of the erector spinae (Bogduk *et al.*, 1992). The EMG signal was full-wave rectified, averaged with a time interval of 0.05 s, filtered with a band-pass of 5–300 Hz, A–D converted at 60 Hz and recorded on the microcomputer. Subsequently, the EMG data were subjected to five-point smoothing and corrected for the effects of electromechanical delay.

During each pull, the curvature of the lumbar spine in the sagittal plane was measured at a frequency of 60 Hz using the 3-space Isotrak. The lever arms D and dw shown in Fig. 1 were estimated (Dolan and Adams, 1993a) and input to the computer so that extensor moment EM could be evaluated from the load cell data using equation (1). Extensor moment was then plotted against EMG activity (E_0) as shown in Fig. 2. (For ease of interpretation, every fifth data point was plotted so that the graph comprised of 40 points, although all data were taken into account in subsequent analysis). Each subject performed between 6 and 12 calibrations with the lumbar spine ranging between the slightly lordotic and fully flexed positions. About 3 min was allowed for recovery between successive calibrations.

After a further recovery period, the subject stood with the lumbar spine flexed by 70% and pulled up with maximum force for 3 s. Between two and four repetitions were necessary to establish a consistent maximum extensor moment which could be used as an index of that subject's strength (Table 1). It is interesting to note that higher extensor moments could be generated in more flexed postures, both in the static calibrations and during dynamic lifts.

Additional experiments were performed on eight subjects in order to clarify the extent of the EMG

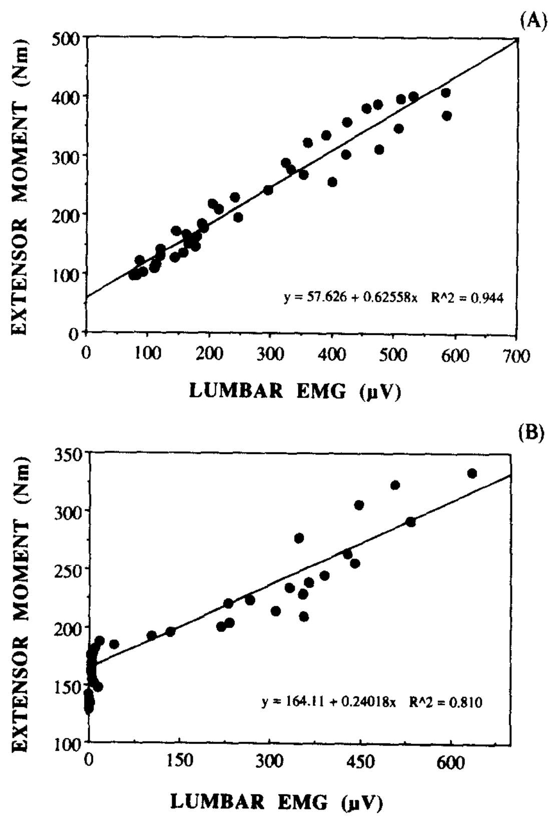

Fig. 2. Isometric EMG-extensor moment calibration when the subject's lumbar spine was (A) slightly flexed, and (B) fully flexed. In (A) the intercept, and hence the passive extensor moment, was small, but in (B) it was large.

silence in full flexion. Extra electrode pairs were attached over the erector spinae at L3, on the right-hand side, and over the latissumus dorsi (7 cm lateral to T7, at 45° to the long axis of the back). Calibrations were performed as described above.

Spinal movements and loading during lifting tasks

Each subject performed five lifts. In the first three, a pen (0 kg), a 10 kg weight-lifter's disc and a 20 kg weight-lifter's disc were lifted from the floor in whatever manner the subject preferred (freestyle). The fourth and fifth lifts were standardised 'squat' and 'stoop' lifts of the 10 kg disc, performed over 4 s to the beat of a metronome. For the squat lift, the subject was instructed to bend the knees, and in the stoop lift to keep them straight. Subjects were allowed to practice the lifts before recordings were made so that they would become familiar with the effort required, and with the feel of the instruments attached to their back. At least one minute was allowed for recovery between lifts, and no subjects showed any signs of fatigue. Eight subjects performed an additional lift of the 10 kg disc after being instructed to 'keep as much lordosis as possible'.

During the lifts, lumbar curvature (LC) and the full-wave rectified and averaged EMG activity were sampled at 28 Hz. Lumbar flexion was calculated for each interval of 1/28 of a second and expressed as a percentage of the subject's full range of movement using the formula:

$$\text{Percentage flexion} = 100 \times [\text{LC} - \text{LC(standing)}]/[\text{LC(full flexion)} - \text{LC(standing)}]. \quad (3)$$

The overall extensor moment (EM) was calculated from the EMG activity, using data from the isometric calibrations on the same subject (Dolan and Adams, 1993a). Essentially, linear or polynomial curve fitting was used to obtain expressions for G and I (Fig. 1) in terms of lumbar curvature, and then equation (2) was used to calculate extensor moment. EMG data were corrected for the electromechanical delay, and for the effects of muscle contraction velocity on the EMG/extensor moment relationship (Dolan and Adams, 1993a).

Comparisons between lifts were made using a repeated measures ANOVA. Significant differences are quoted at the 1% level unless stated otherwise.

RESULTS

Isometric calibrations

The EMG–extensor moment relationship was essentially linear, as shown in Fig. 2. In flexed postures, the linear region was preceded by a short steep region typified by the trace in Fig. 2(B). The start of the linear region was found by visual inspection of the computer monitor, and the gradient and intercept of the linear region was calculated using a least-squares algorithm. The correlation coefficient (R) between E_0 and extensor moment was usually in the range 0.87–0.98, as described previously by ourselves (Dolan and Adams, 1993a) and others (Seroussi and Pope, 1987).

The intercept tended to be higher at L3 than at T10, especially in flexed postures. A passive extensor moment implies electrical silence in *all* regions of the erector spinae, and so for each subject and each calibration, the lower of the two intercepts was used as the 'passive' extensor moment I in subsequent analyses. Additional EMG measurements on eight subjects showed that intercepts were similar bilaterally, and tended to be greater for the latissimus dorsi, so I represents true 'flexion relaxation' of the back muscles.

I increased markedly with lumbar flexion, and in order to assess general trends, nine ranges of lumbar flexion were chosen, as listed in the first column of Table 2. Values of I and 'percentage flexion' were averaged using data from all subjects for all pulls that fell within each range of flexion. Data for men and women were analysed separately and the results are given in Table 2. Figure 3 shows how the 'passive' extensor moment increased with lumbar flexion. From the lordotic standing position up to 80% flexion, the increase is slight, but above 80% it rises considerably to about 120 Nm for men, and 77 Nm for women.

The women showed a weak but highly significant relationship between I and body mass ($r^2 = 0.053$,

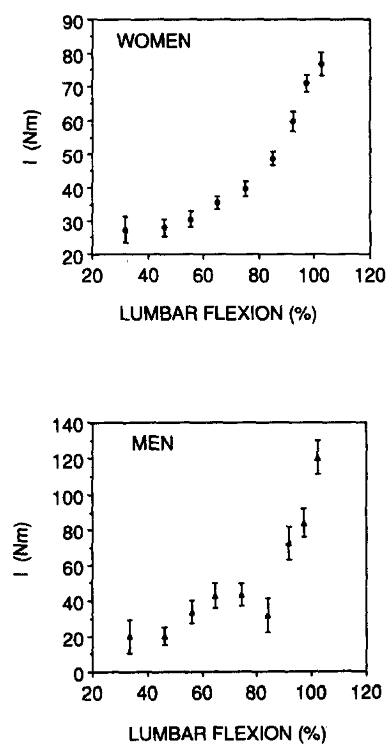

Fig. 3. The 'passive' extensor moment I increased with lumbar flexion for both men and women. Bars indicate the SEM.

Table 2. In the isometric pulls the intercept (I) of the extensor moment–EMG relationship depended on lumbar flexion. Values shown are the mean (SEM) for all pulls within the specified flexion range

	Women		Men	
Flexion (%)	Average flexion (%)	I (Nm)	Average flexion (%)	I (Nm)
<40	31.7	27.4 (3.9)	33.4	19.8 (9.7)
40–50	46.0	28.1 (2.6)	46.2	20.0 (4.9)
50–60	55.4	30.5 (2.4)	56.4	33.8 (6.0)
60–70	64.7	35.5 (1.9)	64.6	42.7 (6.7)
70–80	75.0	39.7 (2.2)	74.5	43.3 (6.4)
80–90	85.2	48.6 (2.1)	84.2	31.8 (9.7)
90–95	92.1	59.6 (2.8)	92.0	72.3 (9.6)
95–100	97.0	71.0 (2.5)	97.3	84.0 (8.3)
>100	102.4	76.7 (3.4)	102.4	120.5 (9.5)

$P<0.0001$; using pooled data for all flexion angles) but this was not apparent for the men ($r^2=0.0003$, $P=0.823$). Stepwise multiple regression showed that none of the other anthropometric variables accounted for any further variation in I which was therefore best described by the following equations:

$$\text{Women:}\quad I=4.91\times 10^{-5}\times F^3+0.7573\times B-24.8 \quad (R^2=0.321,\ p<0.0001), \tag{4}$$

$$\text{Men:}\quad I=7.97\times 10^{-5}\times F^3+12.9 \quad (R^2=0.389,\ p<0.0001), \tag{5}$$

where F is the percentage flexion as defined in equation (3), and B the bodymass (kg).

Spinal movements and loading during lifting tasks

For each subject, and for each lift, the peak extensor moment was calculated, together with the 'percentage flexion' at that moment in time. The flexion values were averaged for each lift, separately for women and men, and equations (4) and (5) were used to calculate the passive contribution to the overall extensor moment (using the average value of bodymass for women given in Table 1). The five lifts are compared in Table 3. In the 10 kg lifts, stoop lifting required significantly more flexion than squat or freestyle lifting and consequently generated a higher 'passive' extensor moment. The 10 kg freestyle lift, however, generated a significantly higher extensor moment than either the 'stoop' or squat lifts, and this may have been because it was generally performed faster. With increasing weight lifted, there was a significant increase in both the total extensor moment and in the 'passive' component I. Total extensor moment rose more markedly than I so that I accounted for a lower percentage of the total extensor moment in the heavier lifts.

The eight subjects who attempted to lift with as much lordosis as possible actually flexed their lumbar spine by 57.5% (STD 13.9%) at the moment of peak extensor moment.

Table 3. Peak values of extensor moment (EM) for the five lifts (±STD). Lumbar flexion and the passive extensor moment resisted (I) were evaluated at the point of peak extensor moment. Peak flexion at any time during the lift is also shown

	Stoop	Squat	Freestyle lifts		
	10 kg	10 kg	0 kg	10 kg	20 kg
Women					
Peak EM (Nm)	210±61	226±61	164±60	244±65	311±83
Flexion (%)	96±10	84±15	81±14	85±15	90±13
I (Nm)	65.8	51.6	47.8	52.6	58.2
I (percentage of peak EM)	31.3	22.7	29.2	21.6	18.8
Peak Flexion (%)	100±8	87±14	83±12	88±12	93±11
Men					
Peak EM (Nm)	276±111	278±85	209±20	328±109	384±140
Flexion (%)	97±9	81±16	78±12	79±16	85±12
I (Nm)	85.0	55.6	50.1	52.8	61.3
I (percentage of peak EM)	30.8	20.0	24.0	16.1	16.0
Peak Flexion (%)	100±8	82±17	79±12	82±15	86±12

DISCUSSION

The size of the extensor moment resisted by passive tissues depended very much on lumbar flexion, so the validity of the Isotrak measurements must be considered. As discussed previously, these measurements can be accurate if the device is mounted on the skin correctly (Dolan and Adams, 1993a) and the average range of flexion of our subjects (Table 1) is in excellent agreement with mobility data obtained from radiographs (Adams and Hutton, 1982; Dvorak et al., 1991; Pearcy et al., 1984). The reproducibility of Isotrak measurements varies from about $\pm 2°$ in erect standing to $\pm 1°$ in full flexion (Adams and Dolan, 1991). It should be appreciated that the inferred passive extensor moments generated during dynamic movements are insensitive to any systematic errors in flexion angles because lumbar flexion serves only to compare isometric calibrations with dynamic lifts, and the Isotrak was used to measure flexion on both occasions.

The origins of the passive extensor moment I are difficult to determine but some inferences may be drawn from the nature of its dependence on lumbar flexion (Fig. 3). In lordotic postures I does not fall much below 25 Nm even though there can be little tension in ligaments and fascia lying posterior to the centre of rotation. This 25 Nm may possibly be due to a raised intra-abdominal pressure. In flexed postures, I increases markedly as the limit of movement is approached, suggesting that passive structures posterior to the spine are being brought into tension, after being slack initially.

It is of interest to compare I in fully flexed postures with the known tensile strength of these posterior structures, and since the available strength data mostly refers to cadaveric material from both sexes, we will consider the average value of I for men and women in the highest flexion range, which is 99 Nm (Table 2). If we suppose that a raised intra-abdominal pressure contributes about 25 Nm in flexed postures also, then there is a further 74 Nm of passive extensor moment to be accounted for. The intervertebral discs and ligaments can resist about 60 Nm when flexed to their elastic limit (Adams and Dolan, 1991), but the protective action of the back muscles would normally limit their contribution to about 20 Nm in static full flexion (Adams and Dolan, 1991). This leaves 54 Nm of I to be accounted for. Mechanical tests on the posterior layer of the lumbo-dorsal fascia (Tesh et al., 1987) and supraspinous ligament (Myklebust et al., 1988) suggest that each of them could sustain a longitudinal force of about 330 N in the lower lumbar spine. Since these lie about 9 and 8 cm posterior to the pivot (Tracy et al., 1989), they could contribute a total of 56 Nm of extensor moment ($330\text{ N} \times 0.09\text{ m} + 330\text{ N} \times 0.08\text{ m}$). This is just sufficient to supply the rest of the passive extensor moment but implies a very small margin of safety. However, there is another potential source of I in addition to those already considered, and that is the stretched, electrically silent muscles themselves. The relative contributions of active and passive tissues to muscle force production is not known, but cadaveric rectus abdominis muscle has a tensile strength of about 14 N cm^{-2} after rigor mortis subsides, and this, presumably, can be attributed to non-contractile tissue (Katake, 1961). If the same figure is applicable to the erector spinae, then the lumbar and thoracic regions would contribute 66 Nm between them (calculated from McGill et al., 1988). This gives a total passive extensor moment of 167 Nm ($25+20+56+66$) which comfortably exceeds our average value of 99 Nm. Several subjects recorded passive extensor moments approaching 167 Nm, but these were particularly strong men who should not be compared with average cadaveric data.

The contribution of passive tissues may not be entirely divorced from muscle activity. The isometric calibrations in flexed postures showed that extensor moment could increase substantially, without any EMG activity from the erector spinae (Fig. 2(B)) and without any significant change in lumbar curvature. This increasing extensor moment must have been generated by muscles remote from our electrodes, and then transmitted across the lumbar spine by passive tissues. Skin-surface electrodes have a relatively wide pick-up area and depth, and would detect strong activation of muscles such as multifidus, as well as the main erector spinae mass (Floyd and Silver, 1955). If the spine is sufficiently flexed, then 'flexion relaxation' extends to the latissimus dorsi, psoas major, and mutlifidus (Floyd and Silver, 1955), so the muscles responsible for this covert action may be the gluteals. The action of the abdominal muscles pulling laterally on the lumbo-dorsal fascia could make a minor contribution to I (MacIntosh et al., 1987) although their main function may be to prevent lateral contraction of the lumbo-dorsal fascia (Gracovetsky and Farfan, 1986) thereby enhancing the action of the gluteals. The relationship between I and lumbar flexion is approximately bi-linear (Fig. 3, upper) suggesting that two separate mechanisms may contribute to the passive extensor moment in different postures.

In the analysis of the dynamic lifting tasks (Table 3) the dynamic flexion angle served as the point of comparison with the static calibrations. This procedure is justified by our previous finding that lumbar curvature (i.e. flexion angle) is the major determinant of the EMG–extensor moment relationship, and that factors such as the knee angle, trunk angle and the height of the hands above the floor have little further influence (Dolan and Adams, 1993a). However, the passive extensor moment may be increased in rapid lifting movements, because collagenous tissue is viscoelastic and resists rapid deformations more strongly than slow ones. There is little quantitative data available to correct for this, but results from our own laboratory show that a motion segment's resistance to bending increases by 8% if the duration of the loading cycle is decreased from 10 to 3 s, and by a further 2% if the loading cycle lasts only 1 s (Adams and Dolan, 1994). Conversely, visco-elastic effects would cause the 'passive' extensor moments to be reduced during sustained or repetitive loading. Our own data suggest that the bending moment on the spine would fall by 17% after 100 full flexion movements performed over 5 min, and by 42% after 5 min of sustained full flexion (Adams and Dolan, 1994).

The results for stoop and squat lifting could be criticised on the grounds that the subjects were constrained to lift in a particular manner. However, this does not apply to the 'freestyle' lifts, and these yielded similar results. Since the lifts were not performed in random order, it is possible that some learning effect, or fatigue, might influence the comparison between different lifts. Any such influence is unlikely to be large, however, because the subjects were allowed to practice the lifts beforehand, and an adequate recovery period was allowed between lifts. On average, our subjects flexed their lumbar spine by 81% in the 0 kg lift, rising to 89% in the 20 kg lift, even though most of them had previously received instruction on lifting technique, and were careful to bend their knees. Many individuals exceeded 100%, and this is not a contradiction in terms: 100% refers to the *static* toe-touching posture, and static limits can be exceeded during dynamic movements because the gravitational forward bending moment is augmented by the forward angular momentum of the trunk. Similar flexion angles in lifting have been measured previously using a variety of techniques (Adams and Dolan, 1991; Davis *et al.*, 1965; Dolan and Adams, 1993b; Potvin *et al.*, 1991). None of our eight subjects found it possible to lift a weight from the ground without substantially flexing their lumbar spine, and the frequently offered advice to 'keep the lumbar lordosis when lifting' appears to be based upon unreliable visual estimates of spinal posture. Perhaps a slight concavity in the region T9-L1 has been confused with a true lumbar lordosis?

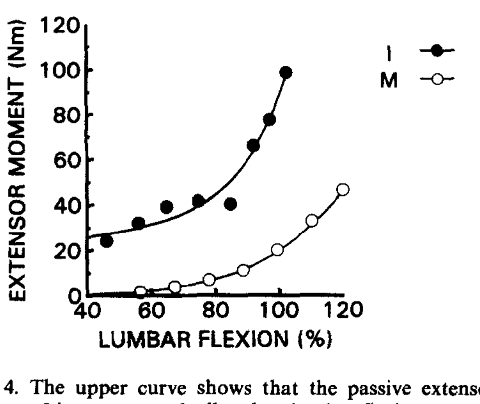

Fig. 4. The upper curve shows that the passive extensor moment I increases markedly when lumbar flexion exceeds 80% of the *in vivo* range of movement (data from Table 2, averaged for men and women). In the range 80–95% flexion, less than 25% of I is attributable to the intervertebral discs and ligaments. This component of I is denoted M (data from Adams and Dolan, 1991).

We have shown how nurses and white collar workers lift moderately heavy weights off the floor, but this may not necessarily be the best way for them to lift. The high incidence of back injuries among nurses (Videman *et al.*, 1984) prompts the question, should they be doing it differently? Our subjects found it impossible to reduce lumbar flexion below about 57%, and it may not be wise for them to do so because lordotic postures generate high stress concentrations in the apophyseal joints (Adams and Hutton, 1980; Dunlop *et al.*, 1984) and posterior annulus fibrosus (Adams *et al.*, 1993, 1994; McNally and Adams, 1992). On the other hand, flexing beyond the normal static limit (100%) would increase the risk of ligament injury (Adams *et al.*, 1980) and leave the posterior annulus vulnerable to prolapse if high compressive forces were present (Adams and Hutton, 1982, 1985; Gordon *et al.*, 1991). There remains a considerable range of movement in which the lumbar spine could be positioned, and the question now is: which part of this range, if any, is to be preferred?

There are two schools of thought. McGill and co-workers advocate lordosis because it allows the back muscles to resist much of the forward shear force acting on the spine, and because it reduces the tensile forces in the intervertebral ligaments which lie close to the centre of sagittal movement and therefore impose a high compressive 'penalty' on the discs (Potvin *et al.*, 1991). The need to reduce shear forces has not been established, since the lower lumbar apophyseal joints are able to resist 1.2–2.5 kN before fracture (Cyron *et al.*, 1976) but the second benefit is readily apparent. Conversely, Gracovetsky and colleagues (Gracovetsky and Farfan, 1986; Gracovetsky *et al.*, 1981, 1985) appear to advocate full flexion because it enables the lumbo-dorsal fascia to resist more of the flexion moment. This structure lies about 9 cm posterior to the centre of rotation (Tracy *et al.*, 1989) and so imposes a small compressive penalty on the discs. Also, tension in the middle layer of the fascia increases the stability of the lumbar spine in the frontal plane (Tesh *et al.*, 1987), and the energy-storing potential of the fascia can reduce the metabolic cost of lifting weights up off the floor (Gracovetsky and Farfan, 1986).

We suggest that this debate has become polarised because different sources of 'passive' extensor moment have not been considered separately. Figure 4 suggests that it is possible to benefit from a high 'passive' extensor moment, and yet avoid high tensile forces in those structures which lie closest to the centre of rotation. The data concerning the intervertebral discs and ligaments is from a previous study on cadaveric lumbar 'motion segments' (Adams and Dolan, 1991) and it shows that their resistance to bending (M) remains relatively low in the physiological (*in vivo*) range of movement. Most of the subjects in the present study flexed their lumbar spines by 80–95% during the lifts, and in this range of movement, M contributes less than 25% of the total passive extensor moment I.

CONCLUSIONS

(1) When lifting weights from the ground, most people flex their lumbar spine by about 80–95% at the time when the extensor moment is greatest.

(2) During stoop and squat lifts of 10 kg, about 31 and 21%, respectively, of the total extensor moment is unrelated to EMG activity in the erector spinae.

(3) Less than 25% of this 'passive' extensor moment comes from the intervertebral discs and ligaments. The rest probably comes from the lumbodorsal fascia, the supraspinous ligament, non-contractile tissue within the erector spinae muscles, and a raised intra-abdominal pressure.

(4) Muscles remote from the lumbar spine are able to increase the passive extensor moment without increasing lumbar flexion.

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