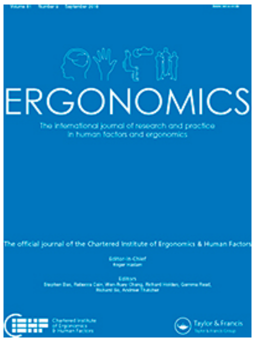

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Dynamic biomechanical modelling of symmetric and asymmetric lifting tasks in restricted postures

SEAN GALLAGHER and CHRISTOPHER A. HAMRICK

US Department of the Interior, Bureau of Mines,
Pittsburgh Research Center, PO Box 18070, Pittsburgh, PA 15236, USA

ARNOLD C. LOVE

Ford Motor Company, Plastic and Trim Products Division,
16800 Executive Plaza Drive, Dearborn, MI 48126, USA

and WILLIAM S. MARRAS

Department of Industrial and Systems Engineering,
The Ohio State University, 1971 Neil Avenue, Columbus, OH 43210, USA

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This article describes investigations of dynamic biomechanical stresses associated with lifting in stooping and kneeling postures. Twelve subjects volunteered to participate in two lifting experiments each having two levels of posture (stooped or kneeling), two levels of lifting height (350 or 700 mm), and three levels of weight (15, 20, or 25 kg). One study examined sagitally symmetric lifting, the other examined an asymmetric task. In each study, subjects lifted and lowered a box every 10 s for a period of 2 min in each treatment combination. Electromyography (EMG) of eight trunk muscles was collected during a specified lift. The EMG data, normalized to maximum extension and flexion exertions in each posture, was used to predict compression and shear forces at the L3 level of the lumbar spine. A comparison of symmetric and asymmetric lifting indicated that the average lumbar compression was greater in sagittal plane tasks; however, both anterior–posterior and lateral shear forces acting on the lumbar spine were increased with asymmetric lifts. Analysis of muscle recruitment indicated that the demands of lifting asymmetrically are shifted to ancillary muscles possessing smaller cross-sectional areas, which may be at greater risk of injury during manual materials handling (MMH) tasks. Model estimates indicated increased compression when kneeling, but increased shear forces when stooping. Increasing box weight and lifting height both significantly increased compressive and shear loading on the lumbar spine. A multivariate analysis of variance (MANOVA) indicated complex muscle recruitment schemes—each treatment combination elicited a unique pattern of muscle recruitment. The results of this investigation will help to evaluate safe loads for lifting in these restricted postures.

1. Introduction

The height of an underground coal mine is generally determined by the thickness of the coal seam. In many cases, the coal bed may not exceed 1·2 m in height. A mine with such a seam thickness may be termed a 'low-seam' coal mine. Workers in these mines are often required to lift heavy materials in severely restricted postures. Two common postures used for lifting in low-seam mines are stooping or kneeling on both knees

(Bobick 1987). Both postures may result in significant compressive and shear loading on the lumbar spine due to external forces (caused by accelerations acting on the mass of both the trunk and the object lifted) and internal forces (for example, those generated by the trunk muscles). Demands on the supporting structures of the column are likely to differ considerably in these postures compared to unrestricted lifting situations. As a result, recommendations for lifting in unrestricted postures (e.g., NIOSH 1981) may have limited applicability to the underground mining environment (Gallagher *et al.* 1990).

Several studies of the stresses of underground lifting have focused on the metabolic and psychophysical responses to underground work. Moss (1934) and Bedford and Warner (1955) both reported increased metabolic costs of working in restricted postures compared to normal upright positions. Durnin and Passmore (1967) reported daily energy expenditures from 2970 to 4560 kcal/day while working in low-coal mines. Humphreys and Lind (1962) and Ayoub *et al.* (1981) also investigated the metabolic cost of performing tasks such as shovelling in restricted postures. These studies supported previous findings that restricted postures result in increased physiological costs when performing underground work activities compared to unconstrained postures. Psychophysical lifting capacity, physiological cost, and electromyography (EMG) of trunk muscles of underground miners in stooping and kneeling postures have been reported in recent studies by the US Bureau of Mines (Gallagher *et al.* 1988, Gallagher 1991, Gallagher and Unger 1990). Results of these studies indicated a significantly lower lifting capacity in the kneeling posture than when stooped, and also demonstrated that EMG activity of the trunk musculature (especially the erectores spinae) was increased in the kneeling posture.

The literature regarding biomechanical stresses of lifting in restricted postures is less extensive. Several studies have studied the intra-abdominal pressure associated with lifting in restricted postures (Davis and Troup 1966, Davis and Ridd 1981, Ridd 1985, Sims and Graveling 1988). These studies generally found increased intra-abdominal pressure when handling materials in stooped postures. Gallagher and Unger (1988) reported biomechanical analysis of kneeling and stooped manual materials handling (MMH) using the model developed by Schultz and Andersson (1981). However, the validation for this model (Schultz *et al.* 1982) only investigated postures with less than 30° of trunk flexion. Thus, it is not entirely surprising that results reported by Gallagher and Unger (1988) indicated this model possesses neither face validity nor predictive validity when analysing the stresses on the lumbar spine in stooping postures.

Recently, an EMG-assisted biomechanical model has been developed which uses EMG to estimate the forces exerted by the trunk muscles and uses this data to calculate lumbar compression and shear, and torques about the spine in three dimensions (Marras and Sommerich 1991a). A validation of the model has been reported by the same authors (Marras and Sommerich 1991b). Among the benefits of this model is the fact that analysis of muscle co-activation is possible, which may significantly affect loading experienced by the lumbar spine. Furthermore, the model allows a dynamic analysis of three-dimensional loading on the lumbar spine which is sensitive to changes in muscle recruitment patterns that are evident as different lifting postures are adopted. An early version of this model was used recently to predict internal forces acting on the spine in standing and kneeling postures during an analysis of an underground scaling bar task (Lavender and Marras 1987), and appeared to provide useful data with which to analyse the stresses in restricted postures. Thus, the purpose of the studies reported in this paper was to use an EMG-assisted biomechanical model to estimate relative changes in compressive and shear loading on the lumbar spine and describe muscle recruitment patterns when performing symmetric and asymmetric lifting tasks in stooped and kneeling postures.

2. Method

2.1. *Subjects*

Twelve healthy men (M = 35·7 years of age ± 6·8 SD) participated in two studies examining the biomechanics of lifting in restricted postures. All subjects were volunteers and received a thorough medical screening (i.e., physical examination and graded exercise tolerance test) prior to taking part in the testing. Subjects operated under terms of informed consent.

2.2. *Experimental design*

Three independent variables were manipulated in each experiment—posture (*P*), having two levels (stooped or kneeling), height (*H*) to which the box was lifted (350 or 700 mm), and weight (*W*) of the lifting box (15, 20, or 25 kg). Split plot designs were employed with the order of treatments being randomized to each subject. EMGs of eight trunk muscles (l. and r. erectores spinae, latissimus dorsi, external oblique, and rectus abdominis) were collected during lifting tasks, and were later digitized and entered into a dynamic biomechanical model. Model estimates of muscle force from eight trunk muscles, compression, anterior–posterior (A–P) shear, and right–lateral (R–L) shear at the L3 level of the lumbar spine were calculated, as well as torques about the *X*, *Y*, and *Z* axes, using the co-ordinate system described by Schultz and Andersson (1981). According to this system, the *X* axis lies in the intersection of the coronal and transverse horizontal planes and is positive to the right. The *Y* axis lies in the intersection of the sagittal and transverse horizontal planes and is positive anteriorly. The *Z* axis lies in the intersection of the coronal and sagittal planes and is positive inferiorly.

2.3. *Apparatus*

Bipolar surface electrodes were used to obtain integrated EMGs of the eight trunk muscles of interest. Belt wearable preamplifiers were used to amplify the EMG signals, which were sent through shielded cables to an amplifier/integrator where the signals were conditioned using an 80 Hz high pass filter and a 1000 Hz low pass filter. The integrator constant was 500 ms. Integrated EMG of all eight trunk muscles for a lifting task were plotted on a computer screen to ensure acquisition of good data prior to saving it on the computer. The fully conditioned signals were routed to a MicroVax II equipped with an A/D board for data storage. (Reference to specific products does not imply endorsement by the Bureau of Mines.) A modified CYBEX II isokinetic dynamometer was used to perform maximum flexion and extension exertions in both stooping and kneeling postures for normalization of EMG data.

The lifting tasks were performed under a 1·2 m roof that restricted the subject's posture. An aluminum lifting box (508 × 330 × 178 mm) with two covered compartments was used to perform the lifting tasks. Adjustable wooden shelves were used to vary the height to which the box was lifted. The frequency of lifting was controlled using a computer generated voice prompt.

2.4. *Experimental procedure*

Subjects performed symmetric and asymmetric lifting studies on separate days. Each day, upon entering the lab, the subject was prepared for the experimental session. The eight trunk muscles were identified and the skin above the muscle was prepared by shaving (if necessary) and cleaning the site with alcohol. Two electrodes filled with electrolyte gel were placed 30 mm apart (centre to centre) over each selected muscle. A single electrode, which served as a ground for the entire system, was affixed at a remote site. After all electrodes had been placed, the subject performed maximum voluntary isometric flexion and extension exertions in both stooping and kneeling postures using a modified isokinetic dynamometer. This allowed EMG data collected during the lifting tests to be normalized to the maximums elicited using the dynamometer. EMG for stooping lifting tasks were normalized to stooping maximum exertions, and EMG for kneeling lifting tasks were normalized to kneeling maximums. Procedures specific to each study are presented in detail below.

2.5. *Biomechanical model*

The EMG-assisted model was used to show the collective effects of trunk muscle activity on spine loading. It was assumed that the lifting tasks studied involved static trunk position or smooth 'isokinetic' motion and dynamic muscle action. The integrated EMG data were analysed by a process of digitizing each separate signal. Four points were identified on each signal: initial activation of the muscle, point where the terminal resting level was achieved, and two points in the middle of the signal such that the shape of the EMG signal was approximated by a rhomboid (Marras and Sommerich 1991a). Figure 1(*a*) shows typical un-normalized EMG data for a lift in the stooped posture. All eight signals for each lift were digitized and data were normalized with respect to time where time = 0·0 indicated time of the first muscle activation, and time = 1·0 represented the time when the final muscle reached it's terminal resting level (Reilly and Marras 1989). Figure 1(*b*) presents estimated muscle forces in the normalized time scale for the data shown in figure 1(*a*).

The biomechanical model estimated compression, shear, torque, and muscle force data at 32 points during the lift. A new calculation was made each time a muscle signal changed slope. Thus, the model allowed estimates of the time-varying loading on the spine. As a result, peak compression and shear loadings on the spine could be analysed. For the current study, the assumption was made that the maximum force that could be exerted by a muscle was $50\,N/cm^2$ [$5\,N/m^2 \times 10^5$] (Reid and Costigan 1987). Estimates of the cross-sectional areas of the trunk muscles were derived from regression models based on trunk anthropometric measurements, following Schultz *et al.* (1982). Figure 1(*c*) demonstrates the compression and shear estimates of the model for the data from figure 1(*a*). Figure 1(*d*) illustrates the estimated torques about *X*, *Y*, and *Z* for the same data.

It is incumbent on the authors to discuss several points regarding the use of this model in the analysis of restricted postures. The validity of the current model has not been established for trunk postures with greater than 45° of flexion. Thus, the model output for stooping conditions must be interpreted cautiously. A particular problem of this (and most) biomechanical models in analysis of this posture is that passive forces, believed to be important contributors to the restorative moment in full flexion, are not addressed. Thus, the authors propose that one should present the results of stooped lifts as representing *relative* changes in spine loading due to muscular forces, and not as absolute compression, shear or torque values. The validity of this approach appears to

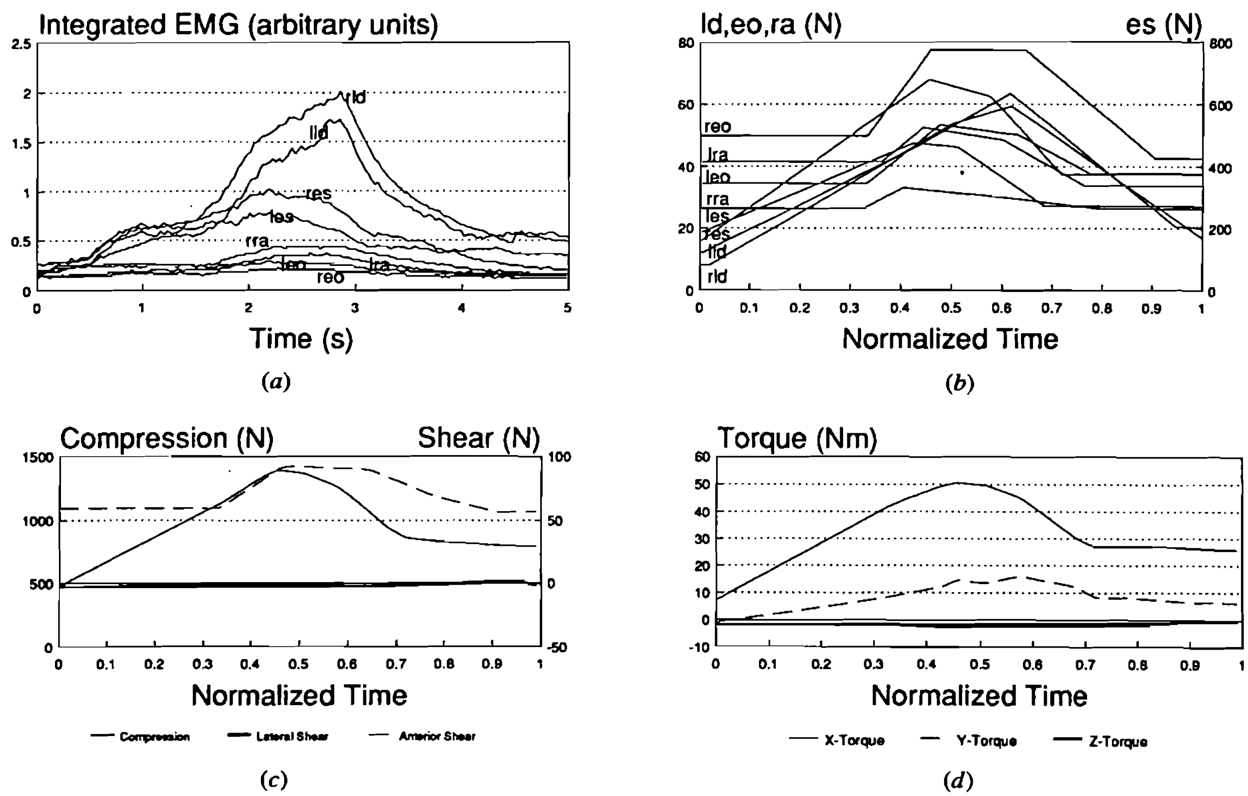


Figure 1. Examples of output from the biomechanical model: (*a*) integrated EMG from a lift in the stooped posture; (*b*) estimated muscle forces for the same lift; (*c*) model estimates of compression and shear for this data; (d) predicted torques about the *X*, *Y*, and *Z* axes.

be supported by recent biomechanical modelling studies that have suggested that forces due to passive components appear to remain relatively constant over a wide range of loads (5·8–32·4 kg) in stooped lifting (Potvin *et al.* 1991).

It should be noted that model estimates are due solely to internal (muscle) forces and do not include forces due to external loads (i.e., weight of the upper body and load). Therefore, the model estimates may be somewhat lower than expected at first glance.

2.6. *Data analysis*

Results of the biomechanical model output included both peak and average values for compressive loading, A–P shear, R–L shear, *X* torque, *Y* torque, *Z* torque, and muscle forces. For each study, dependent variables were subjected to a $2 \times 2 \times 3$ (posture $\times$ lifting height $\times$ box weight) analysis of variance. A multivariate analysis of variance was used in the analysis of forces generated by the eight trunk muscles in the various treatment combinations. On the basis of the univariate ANOVAs, regression models were developed which incorporated only those effects found to be significant. If an interaction was found to be significant, the lower order terms were also included in the model (D. L. Meyer 1989, personal communication). Residuals were examined to determine the additivity of the model. Expected values were then calculated by pooling non-significant effects with the error term.

3. Part 1. Symmetric lifting

3.1. *Experimental task*

The task for the symmetric lifting experiment consisted of lifting and lowering a box with handles from the floor to a shelf directly in front of the subject (i.e., a sagittally symmetric lift). A series of twelve lifting and lowering tasks was performed by each subject in a randomized order. The twelve conditions in this study were a factorial combination of two postures. (stooping and kneeling), two lifting heights (350 to 700 mm), and three box weights (15, 20, and 25 kg). In each treatment combination, the subject was required to lift and lower the box every 10 s for a period of 2 min. EMG data were collected during a specified lift in each treatment combination. Upon completion of each test the subject was allowed 5 min rest, after which the next experimental condition would be run. This process continued until all twelve experimental conditions had been completed.

3.2. *Results and discussion*

Means and standard deviations for model estimates of peak compression, shear, and torques about the three axes are given in table 1. While statistical analyses were performed on both peak and average values for each of the dependent variables, the results reported here represent the analysis of the peak values, except as noted. In general, results of the peak and average ANOVAs were quite similar.

3.2.1. *Effects due to posture*: Changes in posture significantly affected model estimates of loading on the lumbar spine. Peak compression was significantly higher when kneeling ($F_{1,11} = 13{\cdot}08, p < 0{\cdot}005$). However, the stooping posture was associated with increased peak lateral shear forces ($F_{1,11} = 5{\cdot}46, p < 0{\cdot}05$) and increased average A–P shear forces ($F_{1,11} = 5{\cdot}41, p < 0{\cdot}05$). Torque about *X* was higher in the kneeling position ($F_{1,11} = 21{\cdot}8$, $p < 0{\cdot}001$), while torque about *Y* was increased when stooping ($F_{1,11} = 5{\cdot}46$, $p < 0{\cdot}05$).

Table 1. Model estimates of peak compression, shear, and torque for symmetric lifting tasks. Values represent the mean for all 12 subjects. Compression and shear force are expressed in Newtons, while torque values are in Newton-metres. Numbers in parentheses represent the standard deviation.

	Compression	A–P shear	R–L shear	*X* torque	*Y* torque	*Z* torque
Kneeling						
350 mm	1827·3	144·5	17·4	74·7	11·8	4·3
15 kg	(356·4)	(96·2)	(16·6)	(19·4)	(6·9)	(3·5)
350 mm	1933·8	152·9	18·8	77·1	12·5	4·2
20 kg	(412·8)	(100·6)	(10·8)	(29·8)	(6·6)	(2·6)
350 mm	2194·3	183·6	19·3	87·5	13·5	4·5
25 kg	(479·6)	(92·4)	(11·9)	(28·4)	(6·9)	(2·4)
700 mm	1856·1	134·5	19·6	75·7	10·3	3·7
15 kg	(303·1)	(59·0)	(16·3)	(17·6)	(3·6)	(3·0)
700 mm	2220·2	173·4	21·2	88·1	15·3	5·1
20 kg	(359·4)	(76·9)	(14·5)	(19·5)	(9·0)	(3·9)
700 mm	2483·4	208·7	19·8	97·6	11·0	5·1
25 kg	(411·4)	(102·1)	(13·7)	(23·9)	(4·8)	(3·5)
Stooped						
350 mm	1396·0	151·1	19·8	48·2	11·9	4·5
15 kg	(423·8)	(91·5)	(8·7)	(23·4)	(6·0)	(3·5)
350 mm	1586·4	166·4	19·8	53·3	14·0	6·0
20 kg	(501·7)	(50·1)	(11·2)	(27·4)	(5·9)	(4·8)
350 mm	1738·1	180·0	20·5	58·1	15·2	6·1
25 kg	(460·1)	(71·9)	(12·4)	(24·3)	(9·2)	(3·1)
700 mm	1632·3	195·0	20·3	52·2	16·4	8·0
15 kg	(388·6)	(98·5)	(11·0)	(21·3)	(10·4)	(6·6)
700 mm	1807·7	213·4	21·5	61·4	17·4	6·9
20 kg	(440·2)	(104·1)	(9·9)	(25·0)	(10·1)	(4·0)
700 mm	1972·9	250·2	22·5	65·4	19·9	7·5
25 kg	(566·0)	(136·9)	(17·3)	(30·0)	(13·2)	(4·7)

The preceding results appear to illustrate a trade-off in forces experienced by the lumbar spine as one adopts different lifting postures. Lumbar compression due to active muscle contraction is moderated somewhat in the stooping posture; however, the cost of this decrease in compression is that shear forces are increased in both lateral and anterior–posterior directions. The increased shear forces experienced by the lumbar spine when stooping may significantly affect the wear and tear of the intervertebral disks of this region, which are in a particularly vulnerable position in trunk flexion (Adams and Hutton 1981, Pearcy and Hindle 1991). On the other hand, increased compression has been associated with higher rates of back injuries (Chaffin and Andersson 1991). Thus, it is noteworthy that kneeling results in greater compressive forces due to muscle loading. The reason for higher compression when lifting while on one's knees is that the erectores spinae contract more vigorously in this position. In comparison, the 'muscle sparing' action of the posterior ligaments leads to decreased activity of the long erectores when stooping.

3.2.2. *Effects due to vertical distance of lift*: Manipulation of the vertical distance of lift significantly affected model estimates of compression ($F_{1,11} = 8{\cdot}99$, $p < 0{\cdot}05$), A–P shear ($F_{1,11} = 9{\cdot}08$, $p < 0{\cdot}05$), and torque about the X axis ($F_{1,11} = 4{\cdot}88$, $p < 0{\cdot}05$). Each of these dependent variables were increased when lifting from floor to 700 mm compared to lifting from floor to 350 mm. The increase in compression associated with greater vertical load excursion was 12·1%. The increase in A–P shear related to higher lifts was 20·1%, and torque about X was increased 10·4%.

The difference in compression between the two lifting heights is primarily due to the increased activity of the erectores spinae during lifts to the higher level. However, increased activity was observed for all trunk muscles in the high lift conditions, and this co-activity also added to the compression experienced by the lumbar spine. The increased A–P shear observed with high lifts is solely the result of increased activity of the external obliques, according to the equations governing the model.

3.2.3. *Effects due to box weight*: As with lifting height, increases in box weight significantly elevated lumbar compression ($F_{2,22} = 33{\cdot}74$, $p < 0{\cdot}001$), A–P shear ($F_{2,22} = 20{\cdot}25$, $p < 0{\cdot}001$), and torque about X ($F_{2,22} = 15{\cdot}2$, $p < 0{\cdot}001$). On the average, increasing the box weight from 15 kg to 20 kg increased the compressive load by 12·5%, A–P shear by 13·0%, and X torque by 11·6%. Increasing the box weight from 20 kg to 25 kg resulted in an additional 11·1% increase in compression, a 16·5% increase in A–P shear, and a 10·3% rise in X torque. The rise in compression with increased box weight is the result of a general increase in muscular activity of the trunk muscles, particularly of the long erectores, and is related to the increased moment associated with lifting a heavier weight. The only muscles that did not rise consistently with greater box weight were the rectus abdominis. The rise in A–P shear with increasing weight reflects the elevated activity of the external obliques as more weight was handled. The escalation of X torque with increasing weight reveals a relatively greater increase in activity of the back muscles compared to the abdominals.

3.2.4. *Muscle forces*: Model estimates of muscle forces (average of the peak values for all 12 subjects) for each of the experimental conditions are given in table 2. Figure 2 provides normalized EMG activity for each of the symmetric lifting conditions. The MANOVAs for muscle forces indicated a complex relationship of muscle recruitment for the treatments studied in this experiment. The $P*H*W$ interaction for peak muscle forces was significant ($F_{16,28} = 87{\cdot}51$, $p < 0{\cdot}001$), indicating a unique recruitment scheme for each treatment combination. This finding illustrates the tremendous complexity of the trunk neuromuscular system and underscores the difficulty inherent in modelling the forces experienced by the lumbar spine during MMH tasks.

4. Part 2. Asymmetric lifting

4.1. *Task description*

The postures, lifting heights, box weights, and procedures for the asymmetric experiment were identical to the symmetric study, with the exception that the subject was required to lift the box from the floor directly in front of them to a shelf positioned directly to the subject's left. Thus, the box had to be lifted and rotated 90° for placement on the shelf in each treatment combination.

Table 2. Predicted muscle forces (N) for the eight trunk muscles for symmetric lifting tasks. Values represent the mean of peak values over all twelve subjects. Numbers in parentheses represent the standard deviation.

	LLD	RLD	LES	RES	LEO	REO	LRA	RRA
Kneeling								
350 mm	61·9	46·3	825·1	772·2	101·8	104·8	21·3	48·1
15 kg	(36·1)	(28·0)	(148·3)	(191·4)	(73·1)	(70·4)	(10·9)	(36·7)
350 mm	68·1	48·1	845·9	818·8	111·6	106·4	24·9	45·8
20 kg	(34·8)	(23·0)	(236·3)	(261·0)	(82·7)	(64·3)	(11·3)	(33·9)
350 mm	74·0	56·8	1000·6	890·4	132·3	129·9	31·3	51·8
25 kg	(27·5)	(23·0)	(260·0)	(251·2)	(66·0)	(70·6)	(19·6)	(38·3)
700 mm	74·3	50·7	836·1	763·4	89·9	101·6	25·0	46·7
15 kg	(32·1)	(21·3)	(151·8)	(124·8)	(33·2)	(56·1)	(9·8)	(40·0)
700 mm	69·4	57·4	986·0	915·9	113·6	136·0	25·2	63·1
20 kg	(27·4)	(27·4)	(178·1)	(188·0)	(46·4)	(68·1)	(10·3)	(59·8)
700 mm	73·8	59·2	1116·0	1015·4	141·3	155·0	26·8	64·7
25 kg	(32·8)	(36·9)	(214·9)	(195·8)	(65·0)	(83·6)	(11·4)	(43·1)
Stooped								
350 mm	65·0	63·1	531·2	580·1	99·5	116·6	38·3	61·6
15 kg	(25·4)	(33·7)	(199·0)	(226·3)	(64·0)	(73·2)	(20·1)	(43·5)
350 mm	75·6	71·2	600·6	644·9	102·5	133·8	39·0	66·0
20 kg	(27·5)	(24·4)	(224·8)	(284·9)	(35·6)	(52·0)	(12·6)	(39·1)
350 mm	82·9	78·4	674·4	692·7	115·2	141·6	37·9	76·6
25 kg	(34·7)	(36·6)	(222·4)	(251·5)	(59·7)	(55·3)	(11·5)	(47·2)
700 mm	82·3	74·9	631·6	628·5	117·7	160·0	40·7	66·9
15 kg	(32·5)	(35·2)	(190·9)	(180·4)	(78·3)	(83·7)	(14·2)	(30·7)
700 mm	82·5	82·5	674·7	736·0	137·7	165·6	39·7	63·6
20 kg	(29·7)	(38·8)	(220·7)	(225·1)	(88·4)	(68·7)	(13·3)	(33·9)
700 mm	73·3	78·9	779·2	766·4	159·6	196·0	42·6	77·8
25 kg	(21·9)	(37·0)	(297·3)	(271·7)	(112·4)	(96·9)	(16·2)	(46·2)

Note: LLD = left latissimus dorsi, RLD = right latissimus dorsi, LES = left erector spinae, RES = right erector spinae, LEO = left external oblique, REO = right external oblique, LRA = left rectus abdominis, RRA = right rectus abdominis.

4.2. *Results and discussion*

Means and standard deviations for estimated peak values of lumbar compression, shear forces and torques for asymmetric lifting are presented in table 3. Muscle force estimates are displayed in table 4. The following sections concentrate on results of ANOVAs for peak forces and torques.

4.2.1. *Effects due to posture*: Posture significantly affected peak compression ($F_{1,11} = 7{\cdot}57, p < 0{\cdot}05$), which averaged 22% higher in the kneeling posture than when stooping. However, the ANOVA also indicated a significant 'monotonic' (Campbell and Stanley 1963) $P*W$ interaction ($F_{2,22} = 6{\cdot}22$, $p < 0{\cdot}05$). Figure 3 illustrates the interaction between posture and box weight. Examination of this figure shows that peak compression increases at a higher rate in the kneeling posture as W increases. However, compression remained consistently higher when kneeling. In fact, in terms of active

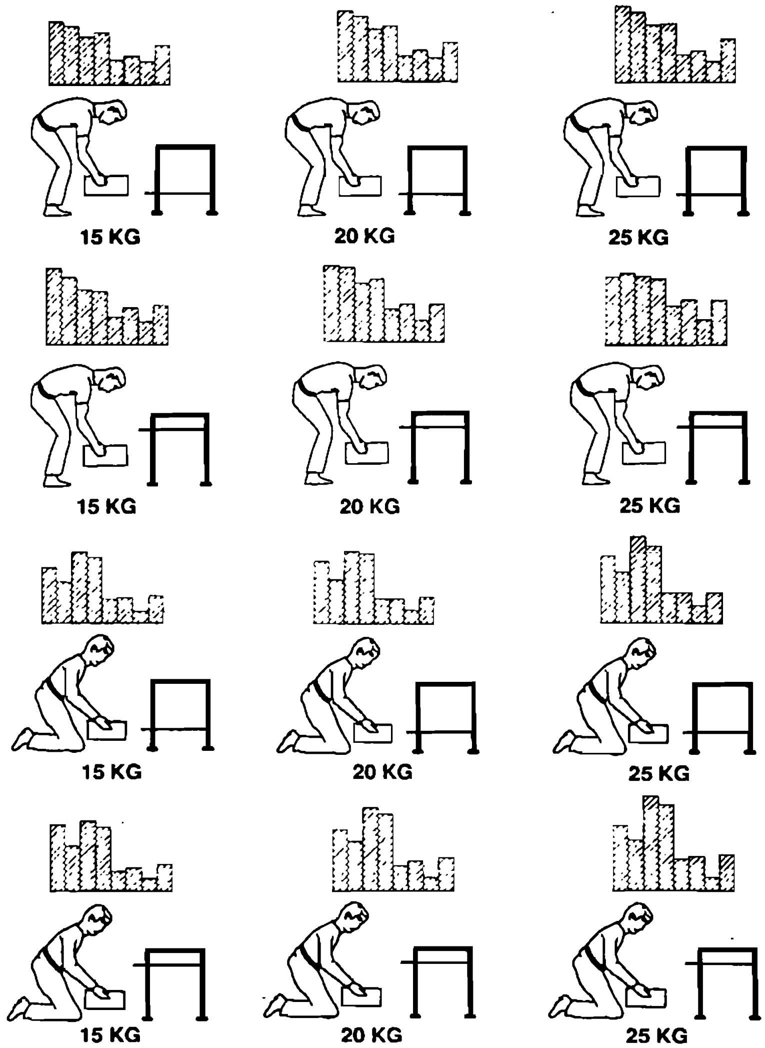


Figure 2. Normalized peak EMG averaged over the 12 subjects (for symmetric lifts). From left to right, the muscles represented in histogram icons represent activity of l. and r. latissimus dorsi, l. and r. erector spinae, l. and r. external oblique, and l. and r. rectus abdominis.

compressive loading, figure 3 illustrates that lifting 25 kg when stooped is approximately equivalent to lifting 15 kg when kneeling.

As with the symmetric experiment, greater compression was predicted in the kneeling posture compared to stooping. However, in asymmetric tasks compression was observed to increase at a higher rate when kneeling as box weight increased. This result was not observed in the symmetric study described above, and is the result of a relatively greater increase in trunk muscle activity (particularly of the erectores spinae) in the kneeling posture as heavier weights were lifted. Previous research has demonstrated that the increase in muscle activity relative to the external load may be non-linear and that critical thresholds exist at which muscles become more stressed (Hamrick 1990). This

Table 3. Model estimates of peak compression (N), shear (N), and torque (Nm) for all asymmetric conditions. Values represent the mean for all 12 subjects. Numbers in parentheses represent the standard deviation.

	Compression	A–P shear	R–L shear	*X* torque	*Y* torque	*Z* torque
Kneeling						
350 mm	1599·6	187·3	21·9	56·5	22·0	5·9
15 kg	(383·4)	(86·0)	(16·0)	(18·5)	(11·2)	(5·2)
350 mm	1788·8	207·5	20·0	62·9	24·7	6·2
20 kg	(345·7)	(88·0)	(16·7)	(20·0)	(11·2)	(4·3)
350 mm	2114·2	275·2	22·3	75·6	25·4	9·4
25 kg	(374·2)	(144·8)	(16·4)	(21·1)	(12·9)	(4·8)
700 mm	1942·1	244·5	24·0	69·4	22·3	8·0
15 kg	(378·6)	(114·2)	(17·2)	(18·0)	(10·1)	(5·5)
700 mm	2188·2	262·5	24·4	78·2	24·9	8·9
20 kg	(322·8)	(98·3)	(16·0)	(17·0)	(11·1)	(4·9)
700 mm	2560·2	360·1	23·6	89·4	22·1	11·1
25 kg	(343·5)	(150·9)	(18·5)	(17·7)	(9·8)	(6·3)
Stooped						
350 mm	1375·0	159·8	24·8	44·8	13·6	6·6
15 kg	(450·9)	(75·8)	(17·1)	(23·0)	(9·3)	(4·3)
350 mm	1521·6	185·9	27·1	49·2	13·5	9·4
20 kg	(502·0)	(75·0)	(14·4)	(25·2)	(8·8)	(5·3)
350 mm	1719·3	194·5	23·4	58·2	15·9	9·4
25 kg	(433·3)	(66·4)	(11·5)	(21·0)	(9·7)	(4·8)
700 mm	1593·5	208·5	27·7	49·7	15·5	12·8
15 kg	(507·6)	(63·7)	(16·6)	(22·9)	(10·4)	(9·0)
700 mm	1810·1	244·3	25·7	57·3	17·5	13·0
20 kg	(549·6)	(99·6)	(13·9)	(27·3)	(12·7)	(7·7)
700 mm	1974·3	279·6	31·4	64·3	20·2	17·1
25 kg	(600·5)	(97·0)	(21·2)	(31·6)	(10·8)	(9·0)

phenomenon may be what is being observed in asymmetric kneeling tasks, and may be responsible for the $P*W$ interaction observed in this study.

Forward bending (X) torque was significantly affected by posture ($F_{1,11} = 9{\cdot}02$, $p < 0{\cdot}05$), being higher in kneeling tasks. Furthermore, posture was the only main effect that had significant bearing on Z (axial) torque in asymmetric lifting ($F_{1,11} = 11{\cdot}43$, $p < 0{\cdot}01$), which was greater in the stooped posture. However, Z torque was also significantly affected by two monotonic interactions: $P*H$ ($F_{1,11} = 21{\cdot}90$, $p < 0{\cdot}001$) and $P*W$ ($F_{2,22} = 4{\cdot}24$, $p < 0{\cdot}05$). These interactions are depicted in figure 4. As can be seen in this figure, Z torque increases at a higher rate in the stooped posture with increases in lifting height. The $P*W$ interaction is characterized by a greater difference in Z torque at the 20 kg condition that at the two more extreme box weight conditions.

4.2.2. *Effects due to lifting height*: Lifting height significantly affected estimates of peak compression ($F_{1,11} = 34{\cdot}33$, $p < 0{\cdot}001$), A–P shear ($F_{1,11} = 39{\cdot}58$, $p < 0{\cdot}001$), and torque about X ($F_{1,11} = 17{\cdot}20$, $p < 0{\cdot}01$). Peak compression increased slightly over 19%

Table 4. Predicted peak muscle forces (N) for the eight trunk muscles for asymmetric experimental conditions. Values represent the mean over twelve subjects. Numbers in parentheses represent the standard deviation.

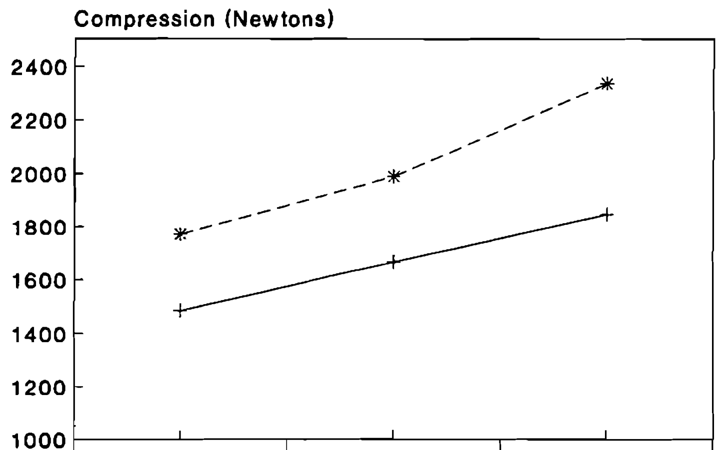
	LLD	RLD	LES	RES	LEO	REO	LRA	RRA
Kneeling								
350 mm	28·7	52·6	641·9	746·6	121·8	149·9	20·5	50·1
15 kg	(12·1)	(29·6)	(219·7)	(214·0)	(65·4)	(72·1)	(7·9)	(47·3)
350 mm	36·0	56·0	704·5	849·4	130·5	166·0	23·9	63·9
20 kg	(12·2)	(30·6)	(192·3)	(229·6)	(66·4)	(75·3)	(10·7)	(76·3)
350 mm	40·7	60·5	856·6	982·9	178·6	215·3	22·8	52·5
25 kg	(17·6)	(32·1)	(212·1)	(210·5)	(113·7)	(117·8)	(9·1)	(38·6)
700 mm	42·2	68·4	735·1	910·1	161·6	188·3	25·0	47·5
15 kg	(18·3)	(36·0)	(155·8)	(204·2)	(98·6)	(85·3)	(14·1)	(34·3)
700 mm	48·5	69·1	875·3	1007·8	173·6	210·2	21·5	64·2
20 kg	(22·8)	(33·4)	(176·7)	(167·9)	(90·2)	(70·4)	(6·3)	(66·9)
700 mm	48·1	67·6	976·1	1161·5	259·5	253·2	26·1	65·7
25 kg	(22·4)	(33·3)	(152·0)	(162·8)	(137·6)	(109·8)	(8·5)	(53·2)
Stooped								
350 mm	54·5	64·3	587·9	493·2	89·6	137·6	50·1	52·8
15 kg	(23·5)	(37·8)	(179·9)	(263·7)	(57·3)	(63·2)	(33·8)	(27·2)
350 mm	65·9	66·1	629·7	554·8	102·7	162·5	40·1	67·0
20 kg	(21·4)	(45·8)	(226·4)	(271·3)	(70·7)	(60·6)	(32·9)	(45·9)
350 mm	63·1	60·5	751·5	617·7	104·7	179·7	39·2	57·0
25 kg	(28·2)	(32·1)	(223·3)	(243·8)	(48·0)	(55·3)	(23·5)	(31·3)
700 mm	64·2	62·5	682·4	561·8	94·9	204·3	44·5	57·9
15 kg	(20·5)	(32·8)	(227·7)	(266·1)	(47·5)	(75·3)	(37·9)	(37·9)
700 mm	70·4	72·1	783·6	620·8	115·0	232·2	39·2	70·5
20 kg	(21·2)	(31·7)	(267·7)	(296·4)	(65·3)	(94·9)	(16·4)	(47·9)
700 mm	86·0	69·2	861·6	669·1	128·2	276·8	42·1	62·6
25 kg	(40·4)	(22·9)	(303·5)	(334·1)	(65·8)	(89·3)	(19·0)	(36·2)

Note: See table 2 for muscle abbreviations.

when lifting to the 700 mm shelf compared to the 350 mm lift. Peak anterior shear increased 32% for the high lifts. For the same increase in H, torque about X increased by 18%. These results reveal the increased muscular responses to overcoming the force of gravity and to maintain control of the trunk during high lifts in restricted postures. This finding provides biomechanical support to numerous studies that have shown that an individual's lifting capacity decreases as vertical distance of lifting increases (Martin and Chaffin 1972, Garg and Chaffin 1975, Ayoub *et al.* 1978, Snook 1978). However, previous biomechanical models have not been able to directly quantify the increased strain as reflected in increased recruitment of the musculature of the trunk, as evidenced by EMG. In a related finding on the effects of vertical distance of lift, Gallagher (1991) found in a combined manual materials handling task that psychophysical lifting capacity in restricted postures was decreased 5% when lifting to a 600 mm shelf compared to a 350 mm height.

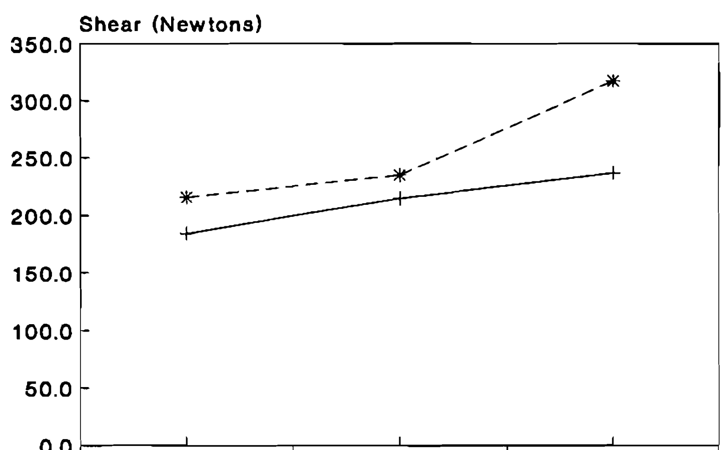
4.2.3. *Effects due to box weight*: Increasing the weight of the lifting box significantly increased peak lumbar compression ($F_{2,22} = 34·33$, $p < 0·001$), A–P shear

($F_{2,22} = 29{\cdot}19$, $p < 0{\cdot}001$), and forward-bending (X) torque ($F_{2,22} = 29{\cdot}19$, $p < 0{\cdot}001$). However, A–P shear was also influenced by a significant $P * W$ interaction ($F_{2,22} = 6{\cdot}41$, $p < 0{\cdot}01$). Figure 5 depicts the interaction of posture and box weight on peak anterior shear. As seen in this figure, peak anterior shear values are about 32 N higher in the kneeling posture when lifting 15 kg. This difference decreases to about 20 N when lifting 20 kg. However, there is a marked increase in anterior shear in the kneeling posture when lifting 25 kg compared to the stooped posture, where the increase in anterior shear remains almost linear.



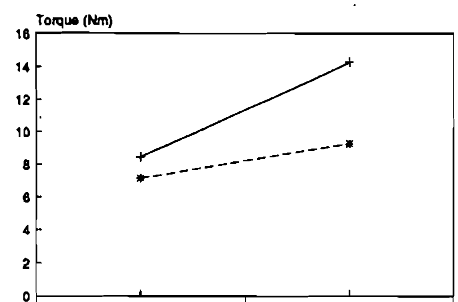
	15 kg	20 kg	25 kg
Stooping —+—	1484	1666	1847
Kneeling —✳·	1771	1989	2337

Figure 3. Interaction of posture and weight of the lifting box on peak compression.

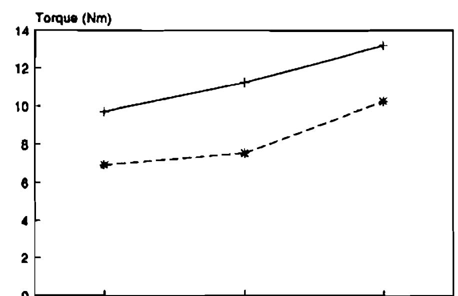


	15 kg	20 kg	25 kg
Stooping —+—	184.2	215.1	237.0
Kneeling —✳·	215.9	235.0	317.6

Figure 4. Posture by box weight interaction on peak anterior shear.



	35 cm	70 cm
Stooping —+—	8.47	14.29
Kneeling —*·	7.18	9.31



	15 kg	20 kg	25 kg
Stooping —+—	9.7	11.23	13.2
Kneeling —*·	6.91	7.58	10.24

Figure 5. *Top*. Interaction of posture and lifting height on torque about the Z axis. *Bottom*. Interaction of posture and box weight on Z torque.

4.2.4. *Muscle forces*: Table 4 presents estimated peak muscle forces for all treatment combinations, averaged over the twelve subjects. As was the case in symmetric tasks, the MANOVA on model estimates of muscle forces evidenced a complex method of muscle recruitment for asymmetric lifting in restricted postures. The $P*H*W$ interaction was significant ($F_{16,28} = 31{\cdot}80, p < 0{\cdot}001$), once again suggesting unique recruitment schemes for each specific treatment combination. Figure 6 presents normalized EMG activity (for peak EMGs) in each condition. This figure demonstrates that, similar to the symmetric lifting study described in part 1, activity of the erectores is significantly greater when lifting in the kneeling pose, while activity of the latissimus dorsi, external obliques, and rectus abdominis are increased when engaged in stooped lifting. In the stooped posture, the asymmetric nature of the lifting task is reflected in activity of the erectores spinae and both pairs of abdominal muscles. The left erectores spinae and right side of both abdominal pairs demonstrated consistently increased activity compared to the contralateral muscles. In the kneeling posture, all four muscle pairs manifested asymmetric activity. In contrast with the stooped posture, the right erectores spinae uniformly demonstrated greater activity when kneeling. The right side of the other three muscle pairs also exhibited increased activity compared to the left side of these pairs when kneeling.

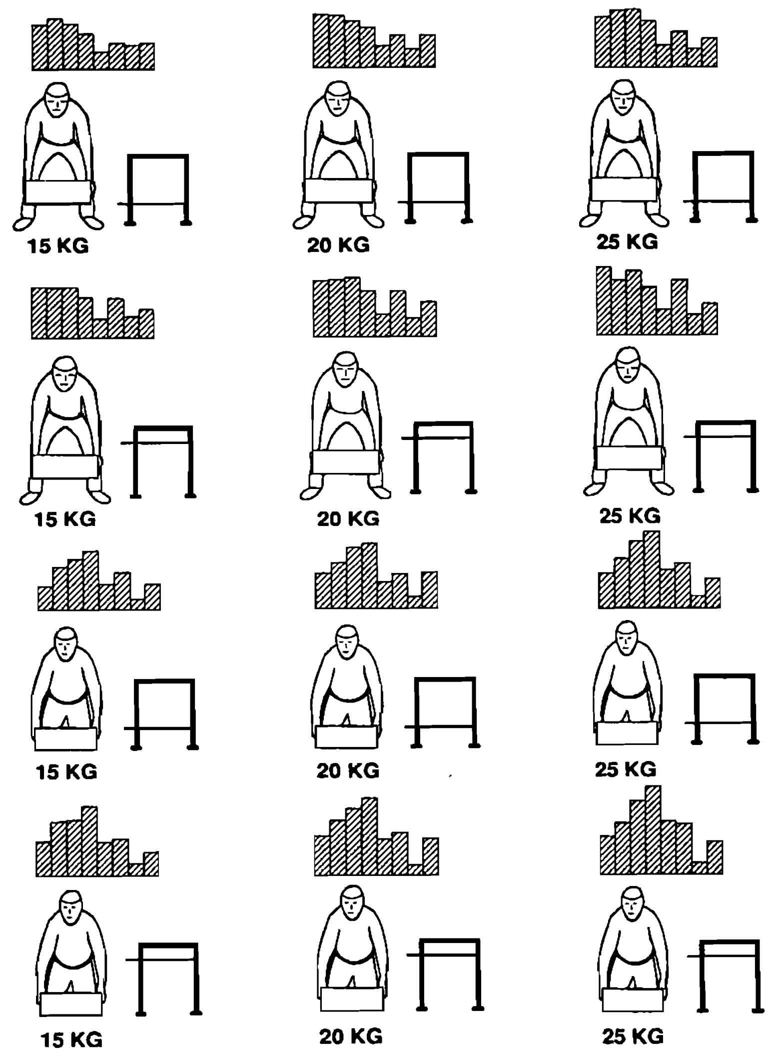


Figure 6. Normalized peak EMG for asymmetric tasks, averaged over the 12 subjects. From left to right, muscles represented in histogram icons are l. and r. latissimus dorsi, l. and r. erector spinae, l. and r. external oblique, and l. and r. rectus abdominis.

4.3. *Asymmetric versus symmetric lifting*

Paired *t*-tests were performed to determine whether significant differences existed between model estimates of compression, shear, torque, and muscle forces for symmetric lifting results (described in part 1) compared to the results of asymmetric lifting. These tests indicated no significant difference in peak compression between symmetric and asymmetric conditions; however, average compression was significantly higher in symmetric lifting ($t_{143} = 2.05$, $p < 0.05$). On the other hand, peak R–L shear was confirmed to be higher under asymmetric conditions ($t_{143} = -3.10$, $p < 0.01$). Asymmetric lifting also resulted in greater peak ($t_{143} = -7.08, p < 0.001$) and average ($t_{143} = -6.41$, $p < 0.001$) A–P shear. Torque about X was significantly increased in symmetric lifting for both peak ($t_{143} = 5.15$, $p < 0.001$) and average ($t_{143} = 4.10$, $p < 0.001$) values. As would be presumed, peak ($t_{143} = -6.36, p < 0.001$) and average

($t_{143} = -2{\cdot}88$, $p < 0{\cdot}01$) torque about Y (lateral bending torque) were both greater in the asymmetric handling activities. Finally, both peak ($t_{143} = -7{\cdot}56$, $p < 0{\cdot}001$) and average ($t_{143} = -4{\cdot}50$, $p < 0{\cdot}001$) axial (Z) torque were significantly increased in asymmetric tasks.

5. Regression models

The following regression models were developed for peak compression, peak shear, and peak torque for symmetric tasks:

$$\mathrm{PKCOMP} = 397{\cdot}0{*}P + 6{\cdot}2{*}H + 41{\cdot}9{*}W + 526{\cdot}2 \quad (R^2 = 0{\cdot}95)$$

$$\mathrm{PKLATSH} = -1{\cdot}40{*}P + 20{\cdot}7 \quad (R^2 = 0{\cdot}24)$$

$$\mathrm{PKANTSH} = 0{\cdot}9{*}H + 4{\cdot}9{*}W + 31{\cdot}6 \quad (R^2 = 0{\cdot}59)$$

$$\mathrm{PK}X\mathrm{TORK} = 27{\cdot}0{*}P + 0{\cdot}2{*}H + 1{\cdot}4{*}W + 17{\cdot}2 \quad (R^2 = 0{\cdot}97)$$

$$\mathrm{PK}Y\mathrm{TORK} = -3{\cdot}4{*}P + 15{\cdot}8 \quad (R^2 = 0{\cdot}33)$$

$$\mathrm{PKZTORK} = 5{\cdot}5 \quad (R^2 = 0{\cdot}0)$$

The following regression models apply to asymmetric lifting tasks:

$$\mathrm{PKCOMP} = -41{\cdot}1{*}P + 9{\cdot}3{*}H + 15{\cdot}9{*}W + 20{\cdot}4{*}P \times W + 452{\cdot}9 \quad (R^2 = 0{\cdot}97)$$

$$\mathrm{PKLATSH} = 24{\cdot}7 \quad (R^2 = 0{\cdot}00)$$

$$\mathrm{PKANTSH} = -53{\cdot}6{*}P + 1{\cdot}9{*}H + 0{\cdot}4{*}W + 4{\cdot}9{*}P \times W + 8{\cdot}9 \quad (R^2 = 0{\cdot}91)$$

$$\mathrm{PK}X\mathrm{TORK} = 18{\cdot}1{*}P + 0{\cdot}3{*}H + 1{\cdot}7{*}W + 5{\cdot}1 \quad (R^2 = 0{\cdot}95)$$

$$\mathrm{PK}Y\mathrm{TORK} = 19{\cdot}8 \quad (R^2 = 0{\cdot}00)$$

$$\mathrm{PKZTORK} = 2{\cdot}8{*}P + 0{\cdot}3{*}H + 0{\cdot}4{*}W - 0{\cdot}1{*}P \times H - 0{\cdot}02{*}P \times W - 4{\cdot}4 \quad (R^2 = 0{\cdot}90)$$

where:

PKCOMP = Peak compression force (N)
PKLATSH = Peak lateral shear (N)
PKANTSH = Peak anterior shear (N)
PKXTORK = Peak torque about X (Nm)
PKYTORK = Peak torque about Y (Nm)
PKZTORK = Peak torque about Z (Nm)
P = Posture (Stooping = 0, Kneeling = 1)
H = Height (mm)
W = Box weight (kg)

6. Discussion

Analysis of the biomechanical data collected in these studies illustrate the trade-offs in lumbar loading that are made when various trunks positions are adopted or when lifting asymmetrically as opposed to symmetrically. The following discussion describes the ergonomic impact of these findings for improving the design of manual handling tasks.

A comparison of asymmetric versus symmetric lifting illustrates how forces acting on the lumbar spine are reorganized as the plane of a lifting task is altered. Data from these studies indicate that lumbar compression is lower when a lifting task is asymmetric; however, the cost of this decrease in compression is that both A–P and R–L shear forces acting on the lumbar spine are increased. Previous asymmetric lifting studies (e.g., Kromodihardjo and Mital 1985) also found decreased compression with asymmetric lifts but attributed the results to the lower psychophysical estimates of lifting capacity observed in asymmetric conditions. However, the present study indicates that compression remains lower in asymmetric tasks even when subjects lift *equivalent* weights in symmetric and asymmetric conditions, at least in restricted lifting postures. Data from these studies would tend to support the notion that tolerance to shear may be a constraint in determining psychophysically acceptable loads (Ayoub and Mital 1989).

It is noteworthy that while lumbar compression is decreased in asymmetric lifts, this type of exertion is associated with a large proportion of low back pain cases (Snook *et al.* 1978). This would imply that the traditional biomechanical parameter used for ergonomic design of lifting tasks (compression) may not relate to the higher incidence of low back pain associated with asymmetric motion. Data from the present studies suggest that increased shear forces present in asymmetric lifting motions may be a factor influencing the higher incidence of low-back pain associated with this type of exertion. Unfortunately, the role of shear forces in production of low-back pain remains equivocal. The best ergonomic advice that can be given at present is to minimize shear forces (i.e., reduce bending and twisting motions) in the design of manual materials handling tasks.

The current studies allow comparison of muscle recruitment in symmetric and asymmetric lifting while lifting equivalent loads. The finding reported elsewhere (Kumar 1980) that lifting asymmetrically entails greater erectores spinae activity did not receive backing by the current studies. Erectores spinae activity in the present study was slightly greater (2·5%) for the symmetric lifting tasks. This discrepancy may reflect the difference in use of the long extensors in restricted conditions compared to unrestricted postures in Kumar's study. Results of the present studies do support the finding of increased external oblique activity in asymmetric tasks, as reported by others (Kumar and Davis 1983). The external obliques were 31·5% more active in the asymmetric conditions than when lifting symmetrically. This finding bolsters the proposition that in asymmetric tasks, the demands of lifting may be shifted to ancillary muscles with smaller cross-sectional areas (Marras and Mirka 1990). These smaller muscles may be at greater risk of injury during MMH tasks.

The effects of the posture adopted for the lifting task also demonstrated a trade-off of forces acting on the lumbar spine. Model estimates indicated that active compression was substantially higher when the kneeling pose was used; however, shear forces were significantly higher when stooping. On the average, peak compression was 22–24% greater when kneeling. This is probably due to two major factors, both of which deal with the erectores spinae. The first is that these muscles are relatively quiescent in the stooped posture, while they may assume increased responsibility for lifting in the kneeling posture. With regard to the first point, it is well-known that the EMG of the long extensors of the spine become silent when flexion of the trunk becomes extreme (Floyd and Silver 1955). In trunk flexion, use of the ligaments to stabilize the spine is much more efficient (owing to their longer lever arm) than use of the supporting musculature. Use of the ligaments in this posture (at least in a mechanically sound column) act to decrease the shear force across the disk and the facet joints. Thus, the 'muscle sparing' function of the ligaments leads to decreased muscle forces and less compression on the discovertebral joints.

The second reason for increased estimates of compression in the kneeling posture is that the erectores spinae may assume a more significant role as an agonist when lifting in this posture. When lifting objects from the floor in an unrestricted standing posture, the gluteal muscles and other strong hip extensors provide a large portion of the forces required to perform a lift (Farfan 1988). However, in the kneeling posture, these powerful extensors appear less capable of providing substantive forces to assist with the lift. Therefore, the erectores spinae appear to be relied upon more heavily in the kneeling posture, especially when contrasted with the action of these muscles in the stooped posture. Of course, these compressive forces acting on the disc lead to tensile forces on the annulus fibrosus, whose fibres may weaken and tear as a result.

Earlier in the discussion the proposition was forwarded that compression could not be relied upon to explain the relationship between asymmetry and low-back pain. A similar argument can be made regarding flexion of the trunk. Stooping has been identified as a risk factor in production of LBP (Lawrence 1955, Partridge and Duthie 1968), yet compressive forces are often calculated to be lower in this posture than when using alternative lifting strategies (Park and Chaffin 1974, Garg and Herrin 1979). The same result was obtained in the current studies. However, higher shear forces are consistently predicted by biomechanical models when analysing this posture (Park and Chaffin 1974, Garg and Herrin 1979), including the present EMG-assisted model. Once again, increased shear forces acting on the lumbar spine may be implicated as a possible factor related to the higher incidence of low back pain in this position. These findings point to our lack of knowledge regarding the effects of shear on the lumbar spine, and suggest that lifting limits in asymmetric and forward flexed postures may ultimately have to be based (at least in part) on the shear tolerance of the lumbar spine.

The lack of data on the shear tolerance of the lumbar spine is only one of the difficulties in quantifying low-back stresses associated with stooped lifting. The primary ergonomic parameters at our disposal for determining lifting limits (compression, physiological cost, and psychophysical estimates of lifting capacity) may all favour the stooped posture over kneeling (Gallagher and Hamrick 1991). In spite of such data, the authors are reluctant to endorse stooping as the preferable lifting technique under constrained conditions. Recent studies have suggested that the reason that stooping may be hazardous may be beyond the scope of currently accepted ergonomic design parameters. For example, studies by Adams and Hutton (1981) have indicated that in forward flexion the disk is less tolerant to compression, suggesting that perhaps different compression lifting limits should be set according to the amount of trunk flexion required in a lifting task. Furthermore, recent studies by Pearcy and Hindle (1991), and Shirazi-Adl and Drouin (1987) have demonstrated that in forward flexion, the facet joints may disengage, allowing greater torsion to be experienced by the intervertebral disk, placing the disks fibres in a position much more vulnerable to injury. It may be noted that IAP does predict increased stress in the stooped posture (Davis and Troup 1966, Davis and Ridd 1981, Ridd 1985); however, use of IAP as a parameter for setting lifting limits has come under question of late (Andersson 1982, McGill and Norman 1987).

The interest in using an EMG-assisted biomechanical model was due to several reasons. First, this technique is the only one available that is sensitive to the changes in muscle action that occur with changes in spinal curvature (Potvin *et al.* 1991).

Furthermore, the authors were interested in investigating the time-varying three-dimensional loading on the lumbar spine in restricted postures. In addition, this model yields information on peak loading of the spine. It would be expected that trauma is most likely to occur under peak loading conditions.

As stated previously, the predictive validity of the current model has not been established for full flexion lifting tasks. However, it should be noted that the face validity of this EMG-assisted model appears to be excellent. Changing experimental conditions generally resulted in the expected manner in both studies. For example, changing from a kneeling to a stooped posture resulted in lower model estimates of compression, but higher shear forces, a result that conforms both to one's expectations and previous modelling efforts (Park and Chaffin 1974, Garg and Herrin 1979). Increases in box weight or lifting height resulted in greater compression and shear estimates given by the model. Comparison of symmetric and asymmetric lifting clearly showed the increased shear forces associated with lifting asymmetrically. In addition, the model clearly demonstrated the complexity of trunk loading, as trunk muscles were recruited in a unique manner for each treatment combination studied. Hopefully, use of EMG data will allow future biomechanical models to address the effects and implications of such complicated loading patterns, not just in terms of muscle activity, but of the temporal aspects of the muscle recruitment.

Results reported in these papers demonstrate several important differences in these lifting postures that are not readily apparent, and which would not have been uncovered with a static analysis of the tasks. For example, the model allows information on co-contraction of antagonistic muscles, which significantly affects the estimates of the compression and shear forces acting on the lumbar spine. The effects of co-contraction during lifting tasks is as yet poorly understood; however, data gathered using this type of model may help to increase our understanding of muscular coordination during lifting tasks. Ultimately, this should permit more realistic and accurate models of trunk loading in occupational work settings. Model outcomes also indicate that certain factors (such as posture and weight combinations in asymmetric lifting) may interact, influencing the compression and shear loading on the spine. Traditional static or optimization models may be unable to identify many of these interactions. Overall, it can be stated that the present model reported no results that would be counter-intuitive, and often identified results that may not have been anticipated, but which are easy to interpret.

The limitations of this model should be recapitulated. As the model requires sophisticated instrumentation, its use is restricted primarily to laboratory settings. In addition, while the model appears to give a good picture of the muscular loading on the spine, it is unable to address other biomechanical factors that may influence production of low back pain. For instance, when the spine is in kyphosis (as in the stooped posture investigated here), a large portion of the restorative moment to maintain the position of the trunk may be provided through the muscle-sparing action of the posterior ligaments, as well as passive loading by the posterior musculature. The strains on ligaments and other posterior structures (such as the apophyseal joints) are apt to be considerable, but unfortunately remain difficult to quantify, and are not addressed by the current model. However, the assumption that concentric muscular activity was present in the stooped posture appeared to be reasonable in the present study, thus the estimates of loading due to muscular exertion should be valid. Additional research is needed so that a better understanding of the passive loading on the spine in flexion can be obtained and incorporated in future biomechanical models.

7. Conclusions

Based on the use of an EMG-assisted biomechanical model during lifting tasks in restricted postures, the following conclusions were drawn:

1. Asymmetric lifting results in decreased compression, but increased anterior–posterior and lateral shear forces compared to symmetric lifting. Furthermore, demands of lifting asymmetrically are shifted to muscles with smaller cross-sectional areas, which may be at greater risk of injury.
2. Compression on the lumbar spine due to active muscle action is increased in the kneeling posture compared to stooping; however, anterior–posterior and lateral shear forces are higher when stooping.
3. Increased height of lift and increased weight of the lifting box increased compression on the lumbar spine as well as A–P shear.
4. A complex relationship exists between lifting posture, lifting height, and box weight and the muscle forces exerted by the eight trunk muscles studied. A unique trunk muscle recruitment profile was identified for each treatment combination.

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