Assessment of the Relationship between Box Weight and Trunk Kinematics: Does a Reduction in Box Weight Necessarily Correspond to a Decrease in Spinal Loading?

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Typically, the simplest and most cost-efficient ergonomic solution to offset the rising costs of low back injuries is to reduce the box weight that is lifted. However, there is limited research on how a worker interacts with the box. In the present study, we quantify the utility of reducing the weight that is lifted – specifically, how changes in the box weight affect trunk kinematics, trunk moments, and ultimately, spinal loads. In the experiment, 15 participants lifted a variety of box weights (from 9.1 to 41.7 kg) from knee height, carried it a distance of 5 feet (1.5 m), and placed it on a shelf at elbow height. For the lower weights, small increases in box weight (3–9 kg) were offset by the trunk dynamics (sagittal velocity), resulting in no difference in spinal loads. At the same time, spinal loads were found to be significantly higher for weights above 25 kg. Thus, when making ergonomic changes (reduction of box weight), it is important to consider how workers will interact with the box. These results indicate that purely weight-based ergonomic controls might not sufficiently reduce the risk of low back disorders. Furthermore, this study provides additional evidence of the utility of using more complex spinal load models (dynamic, multiple muscle models) when evaluating highly dynamic and complex tasks.

INTRODUCTION

Although it is evident that manual material handling (MMH) tasks are associated with many lower back injuries (Bigos et al., 1986; Snook, Campanelli, & Hart, 1978), these jobs are common in warehousing and remain a cost-effective method of material transfer (Drury et al., 1989). MMH tasks often require workers to lift, bend, twist, lateral bend, maintain static postures, carry heavy loads, and perform combinations of these tasks (Bigos et al., 1986; Fathallah, Marras, & Parnianpour, 1998; Kelsey et al., 1984; Snook et al., 1978). For manual handling to be cost effective and most efficient, a worker must lift as much weight as physically possible but remain below the tolerance limit of the lower back. In other words, the effectiveness of manually transferring material depends on the handling costs as well as the injury (medical) costs. Whereas increases in box weight would reduce handling costs, it would also be expected to increase medical costs.

Higher weights would be expected to correspond to higher external trunk moments. Several researchers have found increases in trunk moments to be associated with increases in muscle activity (Andersson, Ortengren, & Nachemson, 1976; Dolan & Adams, 1993; Marras & Mirka, 1990, 1992, 1993; Seroussi & Pope, 1987) and subsequent increases in three-dimensional spinal loads (Chaffin & Park, 1973; Fathallah et al., 1998; Granata & Marras, 1995a, 1995b; Kumar, 1994; Marras & Sommerich, 1991b; McGill & Norman, 1986; Schultz, Andersson, Ortengren, Haderspeck, & Nachemson, 1982). Marras, Granata, Davis, Allread, and Jorgensen (1999) found

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that an increase in box weight of 4.5 kg resulted in higher three-dimensional loads during a depalletizing task. However, those authors limited their scope to only three weights (18.2, 22.7, and 27.3 kg).

Another factor that affects the external moment is the horizontal moment arm, which is the distance between the center of the load and the spine (L_5/S_1; Schipplein, Reinsel, Andersson, & Lavender, 1995). The influence of horizontal distance has been demonstrated by Chaffin and Page (1994), who found that increases in the horizontal moment arm significantly increased the estimated spinal compression. The importance of the combination of weight and moment arm distance (static trunk moment) has also been demonstrated by Marras, Lavender, et al. (1993, 1995), who found that static trunk moment was the single most predictive variable of high-risk jobs and remained an important factor in the multivariate model. Thus it appears that a change in box weight might not necessarily result in changes in spinal loads (e.g., lower weights with higher moment arms) and risk of a low back disorder (LBD).

More in-depth analyses have found that changes in the weight of the box affect trunk kinematics (Allread, Marras, & Parnianpour, 1996; Ferguson, Marras, & Waters, 1992). Allread et al. found that increases in box weight (9.1 kg) reduced bending and trunk velocity in the sagittal and lateral planes. Participants in that study lifted the boxes while standing on a force plate that held the maximum horizontal moment arm constant. By not allowing the participants' feet to move, those authors lost some level of realism, which might have led to the small differences between the weights for sagittal flexion and lateral bending. Similarly, Ferguson et al. (1992) found that increases of 6 kg in box weight resulted in significantly lower sagittal plane motion; in that study, participants were allowed to move their feet. However, those researchers limited their scope to the resulting trunk kinematics.

Trunk kinematics have been found to influence the estimates of trunk moments. Trunk moments have been found to be underestimated by 20% to 30% when dynamics of lifting are ignored (McGill & Norman, 1985). Increased trunk motions also increase muscle activities and spinal loads. Marras and Mirka (1992) found that increased sagittal flexion corresponded to increased muscle activities. Increased sagittal flexion is expected to increase spinal loads by either increasing trunk moment (e.g., increased moment and moment arm) or muscle activity (as found by those researchers). Other researchers have found that increased muscle coactivity (Kim & Marras, 1987; Marras & Mirka, 1992, 1993) and spinal loads (Granata & Marras, 1995a, 1995b; Marras & Granata, 1997b; Marras & Sommerich, 1991b) are associated with increased sagittal trunk velocity. This suggests that trunk kinematics can be altered to adjust for changes in trunk moment (box weight), and the resulting spinal loads might be affected. At present, little information is available regarding the interaction among box weight, horizontal moment arm, and trunk kinematics and how these variables together affect spinal loading and, ultimately, the risk of LBD.

Thus the objective of this study was to evaluate how lifting boxes of different weights while allowing for the movement of feet affects trunk kinematics, horizontal moment arm, and resulting spinal loads.

METHOD

Participants

The participants in this study were 15 men who were college students and had a mean age, height, and weight of 22.5 (*SD* = 2.0) years, 109.1 (*SD* = 4.5) cm, and 73.4 (*SD* = 6.6) kg, respectively. All participants were inexperienced in manual material handling, and none reported a current episode of low back pain.

Experimental Task

In order to better simulate a realistic manual material handling task, the participants lifted a box from knee height, carried it a distance of five feet (1.5 m), and placed it on a shelf at elbow height. The participants moved a box with the dimensions of 25.4 × 42.5 × 32.4 cm (height × width × depth). The handles were located 20.3 cm from the bottom of the box. The box weights evaluated in this study were 9.1, 11.8, 14.5, 17.2, 20.0, 29.9, 32.7, 35.4, 38.1, and 41.7 kg. The participants transferred the boxes at a rate that corresponded to 4.3 lifts/min (once every 14 s). The current study was part of a larger psychophysical study, in that only the first lift of each adjustment session was evaluated. The remainder of the results are reported elsewhere (Davis, Jorgensen, & Marras, 2000; Jorgensen, Davis, Kirking, Lewis, & Marras, 1999). The participants were able to adopt any type of lifting style but were instructed to complete the transfer of the box before a computer tone signaled the start of the next lift.

Apparatus

A lumbar motion monitor (LMM) was used to collect three-dimensional trunk kinematic variables (Marras, Fathallah, Miller, Davis, & Mirka, 1992). The horizontal moment arms between the approximate location of L_5/S_1 and the hands were measured with a tape measure for both the origin and destination of the lift (Marras, Lavender, et al., 1993). Electromyographic (EMG) activity was collected through the use of bipolar silver-silver chloride surface electrodes spaced approximately 3 cm apart over five pairs of trunk muscles: (right and left) erector spinae, latissimus dorsi, internal obliques, external obliques, and rectus abdominis (Mirka & Marras, 1993).

The EMG-assisted biomechanical model used to estimate spinal loading (Granata & Marras, 1993, 1995a; Marras & Granata, 1995, 1997b, 1997c; Marras & Sommerich, 1991a, 1991b) was calibrated using a force plate and an L_5/S_1 locator (Fathallah, Marras, Parnianpour, & Granata, 1997) to determine participant-specific muscle gain. The magnitude of the muscle gain represented the force output of the muscle per cross-sectional unit area for that particular participant. This gain factor was then used to calculate the internal moments and forces for the experimental task that allowed the participants to move without being restricted to a force plate.

All signals from the aforementioned equipment were collected simultaneously through customized Windows™-based software developed in the Biodynamics Laboratory of Ohio State University. The signals were collected at 100 Hz and recorded on a portable computer via an analog-to-digital board.

Experimental Procedure

Surface electrodes were applied to the trunk muscles specified previously using standard placement procedures (Marras, 1990). The participant was then placed in a structure that allowed maximum voluntary contractions (MVCs) of the trunk to be performed in six directions (Marras & Mirka, 1993), which were used to normalize all subsequent EMG data. The MVC structure, which is made of rigid pipe, restricts lower body motion by fixing the pelvis while the upper body is allowed to exert against a shoulder restraint. The MVC exertions were extension while flexed at 20 , flexion in the upright posture, right and left lateral flexion in the upright posture, and right and left twisting in the upright posture. To minimize the possibility of fatigue, a 2-min rest was given between exertions (Caldwell et al., 1974).

The LMM was then placed on the participant's back, and calibration exertions were performed while the participant stood on the force plate. These sagittally symmetric calibration exertions required the participant to lift a 22.7-kg box from knee height to elbow height. After this set of calibration exertions was completed, the experimental lifts were performed; each of the box weights was lifted once by the participants in random order. Upon completion of the lift, a lab assistant returned the box to the starting position.

Data Analyses

Voltages were collected from the LMM and were converted into trunk angles, velocities, and accelerations through customized conversion software. The trunk kinematic data, horizontal moments (the combination of box weight and horizontal moment arm), and lift rate (4.3 lifts/min, or 258 lifts/h) were used as inputs into the multiple logistic regression model from Marras, Lavender, et al. (1993). This model predicts the "probability of high-risk group membership" (hereafter referred to as *LBD Risk Index*). The logistic regression model was developed to predict the odds of a low back injury when comparing the high-risk group (12 or more reported LBD cases per 200 person years) with the low-risk group (no LBD cases per 200 person years). The LBD Risk Index model was based on more than 400 industrial MMH jobs similar to the task evaluated in the current study and thus gives one assessment of potential risk. The index provides an estimate of how similar a task is to the jobs in the high-risk group, which had a significantly higher number of reported injuries than did the low-risk jobs.

Sagittal trunk flexion and horizontal moment arms at the start of the transfer as well as the box weight were used as inputs into the two-dimensional static strength prediction model (University of Michigan, Ann Arbor) to predict statically determined anterior-posterior (A-P) shear and compression forces for the lifts. The static posture at the beginning of the lift was used to minimize the effects attributable to off-plane motion (i.e., lifts were almost pure sagittal motion). This posture also corresponded to the maximum estimated static moments and loading during the lift, because it had the largest moment arm between the box and spine as well as the highest sagittal flexion. After the maximum trunk sagittal position was entered into the model, the positions of the segments for the arms and legs were adjusted until the appropriate distance between the box and L_5/S_1 was obtained. Given that different combinations of these segments can result in the same moment arm distance, several iterations were completed to ensure that peak loads were obtained. The maximum values of these combinations were used as the estimate of the static spinal forces.

The EMG and kinematic data were imported into an EMG-assisted spinal loading model to predict spinal forces and moments in the three cardinal planes on the L_5/S_1 joint (Granata & Marras, 1993, 1995a; Marras & Granata, 1995, 1997b, 1997c; Marras & Sommerich, 1991a, 1991b). Descriptive statistics (peak values) were generated to describe the trunk kinematics, workplace factors (horizontal moment arm and moment), and spinal loading (moments and forces) for each of the box weight conditions. Trunk kinematics, spinal force, and trunk moments were determined for the portion of the transfer associated with the lifting of the loads (e.g., the acceleration of the load), given that this part of the lift was expected to produce the greatest risk for this specific task. Repeated-measures multivariate analyses of variance (MANOVAs) and univariate analyses of variance (ANOVAs) were performed followed by Fisher least significant difference (LSD) multiple pairwise comparisons to determine whether any significant differences were present across the 10 weight conditions.

RESULTS

The results from the MANOVA procedure indicated that a change in weight had a significant global affect on the dependent variables ($p = .0001$). For the ANOVA procedures, we found that several peak kinematic variables were significantly affected by the changes in box weight (p values, means, and standard deviations in Table 1), but none of the peak trunk position variables were found to be affected by weight. Sagittal velocity was significantly ($p = .0001$) affected by weight. The two lowest weights (9.1 and 11.8 kg) had the highest sagittal trunk velocities, whereas the three highest weights (35.5, 38.2, and 41.8 kg) were associated with the lowest sagittal velocities. Overall, there was a downward trend in sagittal velocity with increased box weight. A similar trend was found for maximum sagittal acceleration ($p = .002$), except that the 38.2-kg weight was not different from the first two weights. An interesting trend was found for peak lateral acceleration, in that the higher weights were found to have larger values, with the largest differences found among three of the four lowest weights (9.1, 11.8, and 17.3 kg) and two of the three highest weights (35.5 and 41.8 kg). Thus there appeared to be a trade-off between sagittal and lateral acceleration when box weight was changed.

The horizontal distances and static trunk moments were found to be significantly altered by changes in box weight as assessed by ANOVA (p values, means, and standard deviations in Table 2). Although no difference was found for the horizontal distances at the origin across box weights, participants typically held the box farther away from their bodies when lifting lighter boxes. As expected, the static trunk moments at the origin and destination increased with increases in box weight. However, there were a few weights for which no significant differences were found (e.g., between 35.5 and 38.2 kg at the origin).

TABLE 1: Descriptive Statistics of Trunk Kinematics as a Function of Box Weight

		Box Weight (kg)									
		9.1	11.8	14.5	17.3	20.0	30.0	32.7	35.5	38.2	41.8
		Maximum Trunk Positions (°)									
Sagittal (p = .09)	mean	38.4^{A}	39.2^{A}	37.7^{A}	38.6^{A}	40.2^{A}	40.3^{A}	41.6^{A}	39.5^{A}	39.5^{A}	40.8^{A}
	SD	8.6	8.3	9.7	9.0	8.0	8.0	7.9	8.5	9.4	10.2
Lateral (p = .07)	mean	3.8^{A}	4.7^{A}	4.4^{A}	4.1^{A}	4.4^{A}	4.5^{A}	4.6^{A}	4.4^{A}	5.2^{A}	5.3^{A}
	SD	1.8	1.7	1.9	1.9	1.8	1.4	1.3	1.6	2.4	1.7
Twisting (p = .39)	mean	5.6^{A}	6.6^{A}	7.0^{A}	5.8^{A}	5.5^{A}	5.5^{A}	5.6^{A}	5.0^{A}	5.6^{A}	5.0^{A}
	SD	3.8	4.3	4.2	3.0	4.1	3.9	3.4	3.8	2.2	3.3
		Maximum Trunk Velocities (°/s)									
Sagittal (p = .0001)	mean	41.7^{A}	41.6^{A}	37.1^{AB}	38.9^{AB}	39.2^{AB}	34.7^{BC}	34.6^{BC}	30.7^{C}	29.7^{C}	28.7^{C}
	SD	11.5	11.9	12.0	11.1	9.3	6.1	12.5	8.6	12.6	9.2
Lateral (p = .09)	mean	9.6^{A}	10.1^{A}	10.9^{A}	9.5^{A}	10.7^{A}	11.1^{A}	11.6^{A}	11.0^{A}	11.6^{A}	13.4^{A}
	SD	2.9	2.3	3.5	3.5	3.1	3.5	4.4	3.8	4.7	6.1
Twisting (p = .47)	mean	10.4^{A}	10.7^{A}	12.7^{A}	9.4^{A}	11.9^{A}	10.1^{A}	10.3^{A}	9.2^{A}	11.2^{A}	10.0^{A}
	SD	6.7	8.2	8.0	4.9	9.2	8.3	6.0	6.8	7.4	7.6
		Maximum Trunk Accelerations (°/s^2)									
Sagittal (p = .002)	mean	112.8^{A}	107.2^{A}	97.4^{AB}	96.2^{AB}	100.2^{AB}	86.3^{BC}	87.2^{BC}	76.3^{C}	94.2^{ABC}	76.6^{C}
	SD	37.8	37.1	28.5	20.8	22.7	17.4	33.9	27.4	37.5	34.5
Lateral (p = .03)	mean	42.6^{A}	44.9^{AB}	50.4^{ABC}	44.6^{AB}	50.6^{ABC}	54.6^{ABCD}	55.9^{BCD}	58.5^{CD}	55.9^{BCD}	63.6^{D}
	SD	10.5	9.3	17.6	19.5	16.4	20.5	24.8	23.5	20.6	31.5
Twisting (p = .21)	mean	41.6^{A}	45.7^{A}	48.9^{A}	36.8^{A}	53.4^{A}	38.5^{A}	37.7^{A}	36.3^{A}	47.5^{A}	43.3^{A}
	SD	31.7	37.3	34.4	24.3	46.4	30.0	26.0	26.4	41.4	36.7

Note: Values of p indicate significance results from the analysis of variance tests.
$^{A, B, C, D}$ Different alpha characters indicate significant differences at $\alpha \leq .05$ as found by LSD tests.

When the trunk kinematics (sagittal flexion, lateral velocity, and twist velocity) and static trunk moments were entered into the LBD risk model (Marras, Lavender, et al., 1993), there was an overall increase in the LBD Risk Index ($p = .0001$). However, an increase of about 3 kg did not necessarily correspond to an increase in the LBD Risk Index. Also, box weights greater than 30 kg were found to have LBD risk values above .6, the level of probability that almost guarantees "high-risk group membership" (Marras, Allread, & Reid, 1999). Those authors found that 97% of the jobs above .6 were high-risk jobs.

As with static trunk moment, the static compression and A-P shear forces were found to steadily increase with increases in box weight (*p* values, means, and standard deviations in Table 3). Again, for a few of the weights, no differences in static loads were found for an increase of 3 kg (e.g., 11.8 and 14.5 kg, 35.5 and 38.2 kg for A-P shear, and 32.7 kg and 35.5 kg for compression). Hence the values of static moment and spine loading reflect mainly the magnitude of the box weight and fail to account for trunk kinematics. Table 3 also shows that there was an increase in the dynamic sagittal trunk moment with increases in box weight. Four levels of dynamic sagittal trunk moments were found to statistically differ (Table 3). Whereas the static and dynamic sagittal trunk moments increased with weight, dynamic resultant trunk moments among the five lowest weights as a group did not differ significantly, and the five highest weights as a group did not differ significantly. Thus the dynamic trunk moments represented the interaction between trunk kinematics and box weights, whereas the static moments appeared only to reflect changes in box weight.

The trends in spinal loads were found to be similar to the trends in trunk kinematics, in that increases in weight did not necessarily correspond to increases in dynamic loading (*p* values, means, and standard deviations in Table 3). With respect to maximum dynamic lateral shear loads, there were no differences among the five lowest weights as a group, and there were no significant differences among the five highest weights as a group. There were five significant differences between the various weights for dynamic A-P shear force, with substantial differences found only among the 9.1 kg box and the two highest box weights (38.2 and 41.8 kg). In general, there was an increasing trend in dynamic A-P shear load with increases in weight. Whereas there were four levels of significant differences among the dynamic compressions, the largest difference occurred among the five lowest weights as a group and the five highest weights as a group, similar to the findings with the dynamic trunk moments. An example of the general trend for the dynamic moments and spinal forces is shown in Figure 1.

DISCUSSION

Small increases in box weight were found to have limited effect on the dynamic spinal loads. Typically, the increases in box weight were offset with increases in trunk kinematics. It appears that there was a change in lift dynamics between 20 and 30 kg. The 20-kg boxes were found to have a probability of LBD risk of .49, whereas the 30-kg box weight had values of .71 (above the .6 level, which almost guarantees high-risk group membership). To further investigate the box weights between 20 and 30 kg as well as the possibility of a threshold weight level, we asked the participants to lift boxes weighing 23, 25, and 27 kg. The LBD Risk Index was .56, .59, and .59 for the 23-, 25-, and 27-kg weights, respectively, whereas the dynamic spinal loads resulted in lateral shear forces that ranged from 620 to 740 N, A-P shear forces that ranged from 1220 to 1370 N, and compression forces that ranged from 5680 to 5880 N.

Based on the spinal loading and LBD Risk Index, it appears that weights above 25 kg resulted in a substantial increase in risk of low back injury. In addition, the sagittal trunk velocity for the three weights (23, 25, and 27 kg) was found to be 40.3°/s, 37.2°/s, and 35.9°/s, respectively. Thus, as found in the original study, the decreasing trend in velocity with increases in box weight was found for these weights, and it appears that the 25-kg weight was the level at which participants slowed down during the lifting task.

The utility of this weight threshold is even more apparent when considering the spinal

TABLE 2: Descriptive Statistics for Workplace Factors as a Function of Box Weight

		Box Weight (kg)									
Workplace Factors		9.1	11.8	14.5	17.3	20.0	30.0	32.7	35.5	38.2	41.8
Moment arm at origin (cm; p = .14)	mean	45.6^{A}	46.1^{A}	46.7^{A}	46.9^{A}	46.4^{A}	45.9^{A}	45.9^{A}	44.2^{A}	46.4^{A}	45.4^{A}
	SD	4.1	3.4	3.8	3.6	3.5	3.5	4.3	3.7	4.7	3.8
Moment arm at destination (cm; p = .0001)	mean	45.6^{A}	44.5^{ABC}	44.5^{ABC}	45.9^{A}	44.9^{AB}	42.0^{CD}	41.1^{D}	41.3^{D}	42.7^{BCD}	40.3^{D}
	SD	5.0	5.8	5.0	7.8	8.6	4.8	5.7	4.4	7.3	3.9
Static moment at origin (Nm; p = .0001)	mean	49.3^{A}	60.6^{B}	74.3^{C}	91.8^{D}	98.0^{D}	139.8^{E}	150.5^{F}	155.9^{FG}	164.5^{G}	190.3^{H}
	SD	8.2	7.8	9.1	22.0	9.0	11.1	19.7	16.0	31.8	20.6
Static moment at destination (Nm; p = .0001)	mean	49.6^{A}	59.4^{A}	71.6^{B}	92.4^{C}	95.9^{C}	130.4^{D}	137.3^{DE}	148.3^{EF}	154.3^{F}	170.9^{G}
	SD	8.2	9.9	10.9	33.2	19.7	17.7	26.6	22.7	40.9	22.4
LBD risk index (p = .0001)	mean	$.23^{A}$	$.26^{A}$	$.36^{B}$	$.47^{C}$	$.49^{C}$	$.71^{D}$	$.75^{DE}$	$.76^{DE}$	$.80^{E}$	$.87^{F}$
	SD	.06	.07	.09	.17	.13	.12	.12	.11	.16	.06

Note: Values of p indicate significance results from the analysis of variance tests.
$^{A,\ldots,H}$ Different alpha characters indicate significant differences at $\alpha \leq .05$ as found by LSD tests.

TABLE 3: Descriptive Statistics of Static Spinal Loads, Dynamic Trunk Moments, and Dynamic Spinal Loads as a Function of Box Weight

		Box Weight (kg)									
		9.1	11.8	14.5	17.3	20.0	30.0	32.7	35.5	38.2	41.8
Maximum Static Spinal Loads (N)											
A-P shear (p = .0001)	mean	326.3^{A}	349.1^{B}	365.4^{B}	387.8^{C}	410.6^{D}	482.2^{E}	503.0^{F}	525.4^{G}	543.7^{G}	575.9^{H}
	SD	32.6	32.3	34.2	33.9	34.4	41.0	49.6	38.1	41.2	46.1
Compression (p = .0001)	mean	2244.9^{A}	2433.0^{B}	2564.7^{C}	2756.5^{D}	2927.6^{E}	3458.4^{F}	3609.2^{G}	3688.8^{G}	3887.9^{H}	4058.6^{I}
	SD	344.8	282.8	303.5	295.7	300.0	342.4	430.0	370.1	399.7	435.2
Maximum Dynamic Trunk Moments (Nm)											
Sagittal (p = .0001)	mean	165.1^{A}	178.4^{A}	187.1^{AB}	192.7^{AB}	206.8^{B}	259.3^{C}	278.1^{C}	275.0^{C}	284.6^{CD}	309.9^{D}
	SD	82.1	97.2	96.7	90.7	82.2	99.8	112.2	100.3	114.0	123.5
Resultant (p = .0001)	mean	191.8^{A}	215.6^{A}	225.7^{A}	227.1^{A}	233.1^{A}	307.2^{B}	323.8^{B}	318.7^{B}	335.2^{B}	352.7^{B}
	SD	114.6	159.2	152.7	112.5	90.5	157.4	145.2	143.8	156.4	156.3
Maximum Dynamic Spinal Loads (N)											
Lateral shear (p = .0001)	mean	416.1^{A}	425.0^{A}	498.8^{A}	557.0^{AB}	472.2^{A}	711.6^{BC}	774.1^{C}	730.1^{BC}	738.7^{C}	758.7^{C}
	SD	308.3	312.3	296.9	352.2	229.8	521.8	582.6	466.1	547.1	652.6
A-P shear (p = .0001)	mean	1068.6^{A}	1166.4^{ABC}	1135.7^{AB}	1166.1^{ABC}	1209.3^{ABCD}	1286.9^{BCDE}	1367.4^{DE}	1329.2^{CDE}	1380.7^{E}	1437.5^{E}
	SD	208.4	341.2	258.5	245.6	379.0	424.4	517.6	407.8	552.8	492.2
Compression (p = .0001)	mean	4331.4^{A}	4644.2^{AB}	4671.9^{AB}	4814.1^{AB}	5062.6^{B}	6054.2^{C}	6490.0^{CD}	6315.5^{C}	6491.5^{CD}	6962.5^{D}
	SD	1705.1	2144.3	2024.6	1870.4	1750.0	2098.8	2296.2	2133.1	2409.0	2569.8

Note: Values of p indicate significance results from the analysis of variance tests.
$^{A,\ldots,I}$ Different alpha characters indicate significant differences at $\alpha \leq .05$ as found by LSD tests.

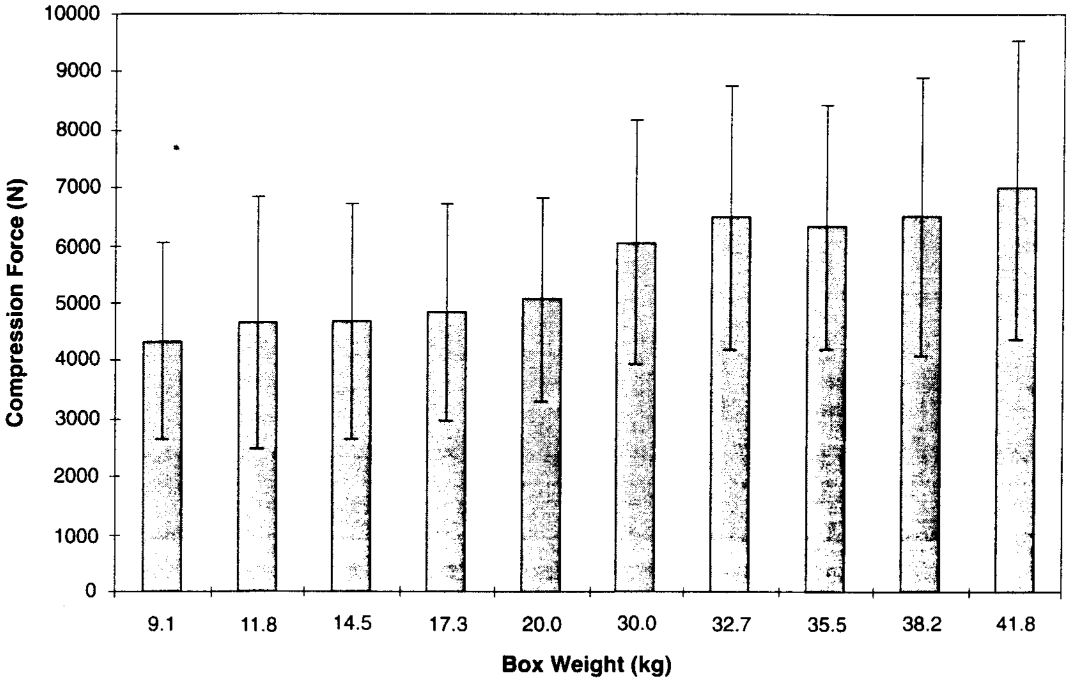

Figure 1. Maximum dynamic compression force as a function of box weight.

tolerance limits for compression and shear forces. There are two commonly accepted tolerance limits for compression: one at 3400 N, at which some individuals begin to have vertebral endplate microfractures, and one at 6400 N, at which 50% of workers would be expected to have vertebral endplate microfractures (NIOSH, 1981). The spinal tolerance limits for shear are not as well defined, but McGill (1996) estimated the shear tolerance limits (both lateral and A-P shear) to be around 1000 N. At this level, there is an increased probability of tears in the annulus fibrosis.

Weights above 25 kg were found to have dynamic compression values that approached the upper tolerance level and dynamic shear loads that were at or above the shear tolerances. Combination loads might result in higher probabilities of LBD (Shirazi-Adl, 1991). It is not recommended that 25 kg be used as a threshold because of uncertainty of applicability to the general population and because the loads exceed tolerance limits for compression (3400 N limit) and A-P shear (1000 N) forces. Thus, even lower weights should be investigated to determine whether a safe load can be lifted under similar conditions. The dynamic spinal loads in the present study were similar to those found by Marras, Granata, et al. (1999) for similar box weights and a depalletizing task.

Spinal loads were affected by several factors. There seemed to be a relationship between changes in weight and changes in trunk dynamics and horizontal moment arm. Trunk dynamics have been found to increase trunk muscle coactivity and spinal loads (Granata & Marras, 1993, 1995a; Kim & Marras, 1987; Marras & Mirka, 1992, 1993; Mirka & Marras, 1993). In the present study we found that the muscle activities of the extensor muscles (erector spinae and internal oblique muscles) were significantly higher for the five highest weights as compared with the five lowest weights (see Figure 2), basically mirroring the dynamic loading trends. The muscle activity of the flexor muscles (left rectus abdominus and both external oblique muscles) was also found to be higher for the five highest weights (Figure 3). Thus as weight increased, muscle coactivity increased. Whereas the general trend in muscle activity increased with increased box weight, several muscles appeared not to be sensitive to small changes in box weight. This might be a direct result of the complex

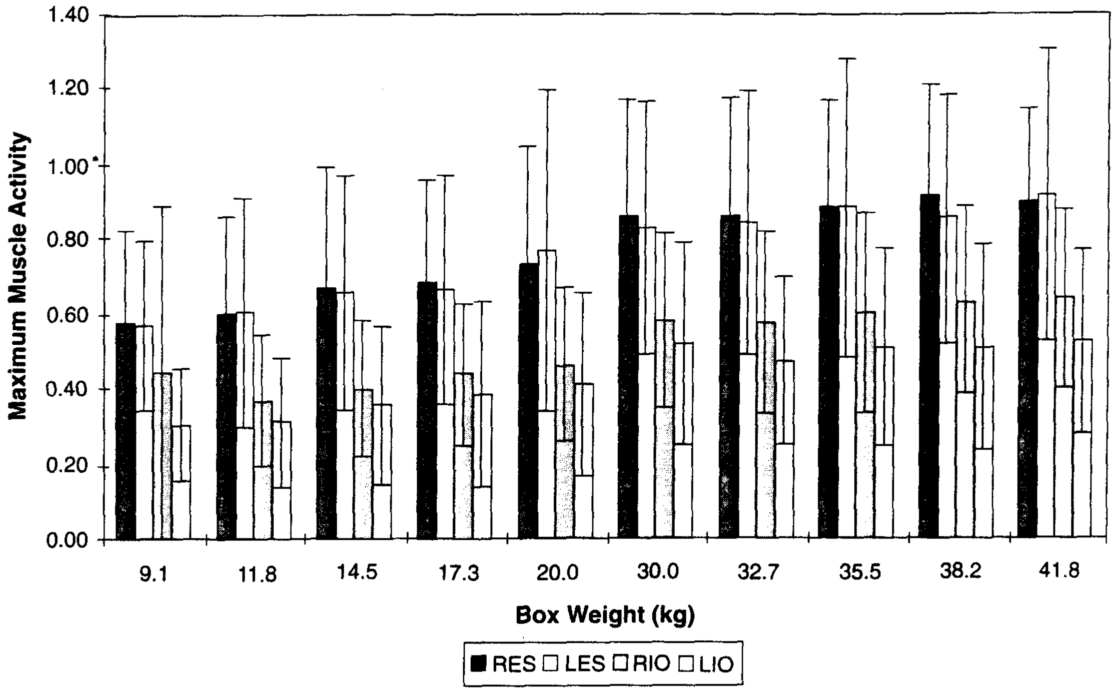

Figure 2. Maximum muscle activity of the extensor muscles as a function of box weight (RES = right erector spinae, LES = left erector spinae, RIO = right internal oblique, LIO = left internal oblique).

relationship among box weight, horizontal moment arm, and trunk kinematics.

Several authors have illustrated the importance of considering muscle coactivity and trunk dynamics in the estimation of spinal loads. Static models that neglect trunk dynamics have been found to drastically underestimate the spinal loads by as much as 22.5% to 60% (Marras & Sommerich, 1991a). In addition, models that neglect trunk muscle coactivity have been found to underpredict spinal loads by as much as 45% (Granata & Marras, 1995b). Both of these underpredictions have been estimated for sagittally symmetric lifts and thus probably represent a best-case scenario. Given that lighter weights were found to have higher sagittal velocities (more muscle coactivity) and larger horizontal moment arms, the influence of box weight on the estimation of spinal loads could be significantly affected by the model used to estimate them (e.g., static single equivalent muscle models vs. dynamic multiple muscle models). Using a static single equivalent muscle model, the static compression and A-P shear forces were estimated for each of the lifting tasks.

As with external static trunk moment, the static compression and A-P shear force values increased almost uniformly with weight. The spinal loads predicted by the static model were compared with those from the dynamic model by computing ratios. The compression ratios ranged from 1.9 (light weights) to 1.65 (heavy weights). This indicates that in a highly dynamic lift that contains complex trunk motion, the compressive values would be underpredicted by 60% to 90% when using a static model as compared with a multiple-muscle dynamic model. The importance of dynamics was even more apparent when the A-P shear force ratios were evaluated (Figure 4). The ratios ranged from 3.3 (lowest weights) to 2.5 (highest weights). This indicates that static models underpredict A-P shear by 150% to 230%, depending on the weight that is lifted. It is apparent that the lighter weights were influenced more by the dynamics of the lift and the resulting coactivity than were the heavier weights. These results show why trunk kinematics and muscle coactivity must be considered when estimating spinal loads and evaluating MMH tasks.

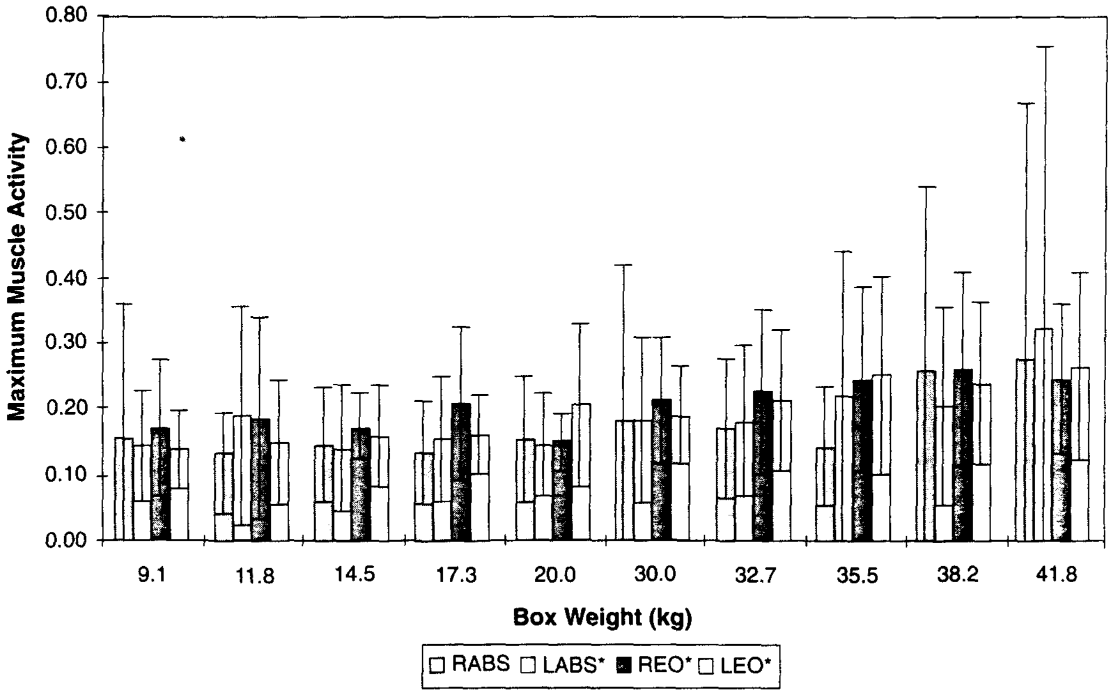

Figure 3. Maximum muscle activity of the flexor muscles as a function of box weight (RABS = right rectus abdominus, LABS = left rectus abdominus, REO = right external oblique, LEO = left external oblique; * indicates significant effect of box weight).

Several potential limitations must be considered when interpreting the results of this study. The range of the box weights used in this study was limited to weights above 9.1 kg. Future studies should be performed to determine whether the spinal loads would be lower for even lighter weights and result in spinal loads that would be considered safe (below all three spinal tolerance limits). Based on data gathered by Marras and Davis (1998), sagittal symmetric lifts of boxes weighing 3.4 and 6.8 kg were found to have sagittal velocities (49.7°/s and 44.6°/s, respectively) that were greater than those for the 9.1 kg weights in this study. Whereas the spinal loads are not directly comparable, the levels for the lower weights were found to be significantly lower than those for the 9.1 kg weight, with lateral shear force less than 200 N, A-P shear force less than 820 N, and compression force less than 3600 N. These results provide some indication that lower weights might be further offset by changes in trunk dynamics, indicating that a change in weight might not result in a resultant change in spinal loads.

The MMH task evaluated in this study was a combination task that included lifting, carrying, and lowering boxes. Although different tasks might result in slightly different results, the current task represented a complex task that is commonly found in industry. Simpler tasks, such as pure sagittal lifting, would yield lower spinal loads as well as less underprediction of spinal loads, but these tasks are not typically found in industry (Marras, Lavender, et al., 1993).

The results would be most applicable to lifts performed at a rate of 4.3 lifts/min. Slower lift rates would be expected to result in lower spinal loads because trunk kinematics would be expected to be lower, whereas faster lift rates would probably result in more trunk motion and higher spinal loads. Future research is needed to investigate the interaction between lift rate and box weight.

In addition, these results are most applicable to a male student population. The trunk motions and spinal loads might be different in other populations, such as women and experienced manual material handlers. Experienced

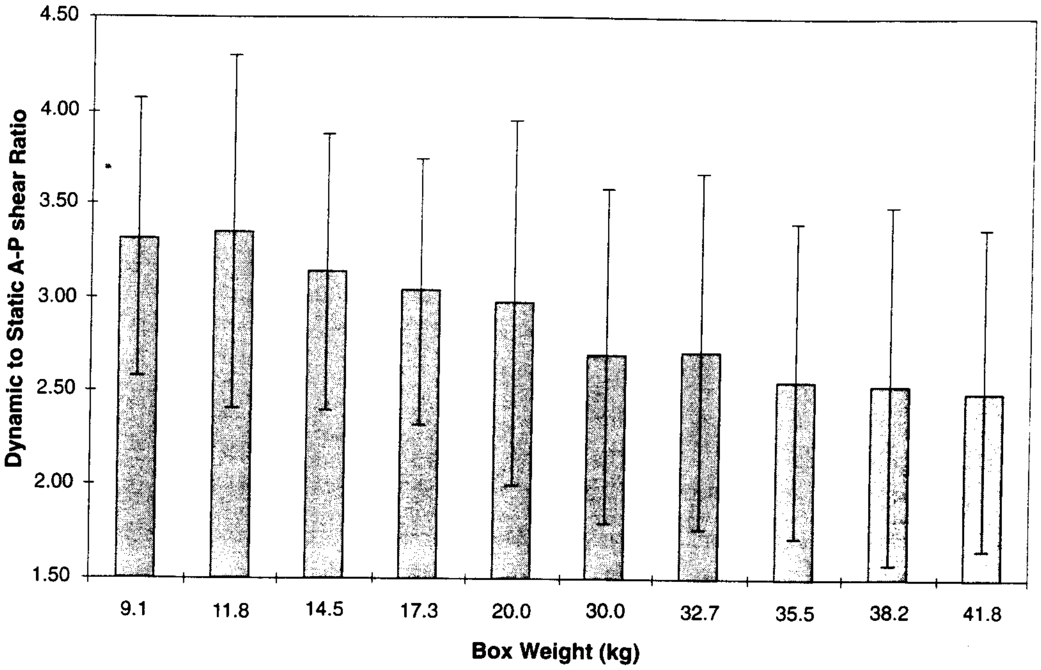

Figure 4. Ratio of dynamic A-P shear force to static A-P shear force as a function of box weight.

workers would be expected to move differently as well as to have different trunk muscle recruit patterns, which might result in different spinal loads (Granata, Marras, & Davis, 1999). The spinal loads might also be different for women, given that their muscle cross-sectional area and lines of action have been found to be significantly different from those of men (Jorgensen, Marras, Granata, & Wiand, 2000; Marras, Jorgensen, Granata, & Wiand, 2000).

The results of the static spinal loading should be considered as "best" estimates because the methods used to predict these values were based on estimates of body posture. Although every effort was made to accurately represent the postures of upper extremities, no actual measurements were taken. However, the estimate of static loads was based on the maximum value obtained for all of the various combinations of upper body links that kept the same moment arm and trunk angle. Thus any misrepresentation of these upper extremity links would reduce the estimates for static loading.

Finally, the effect of repetitive lifting was not evaluated in the present study because the participants completed each weight condition only once. It would be expected that additional changes in trunk motion, muscle activity, and subsequent spinal load would result from the accumulation of fatigue over an entire workday. Marras and Granata (1997a) found that trunk kinematics (e.g., sagittal posture and velocity) and sagittal trunk moment decreased as participants became fatigued over a 5-h study session. They also found a change in the loading patterns, in that compression forces decreased whereas A-P shear forces increased, resulting from changes in the muscle activity patterns. Although that study provides an indication about how trunk kinematics and spinal loads were altered under fatiguing conditions, future studies are needed to evaluate longer periods of time that are representative of an entire work day (e.g., 8–10 h of lifting).

CONCLUSION

The present study quantifies the utility of reducing the weight that is lifted and indicates the importance of considering trunk kinematics and muscle coactivity. Small changes in weight were found to influence three-dimensional spinal loads only slightly. Changes in box weight appeared to be offset by changes in trunk kinematics (sagittal velocity) and the accompanying changes in muscle coactivity. In general, two levels of spinal loads were found: one for the five lowest weights (9.1–20 kg) and one for the five highest weights (30–41.8 kg). Thus, purely weight-based analyses might not be sensitive enough to spinal load and, ultimately, risk of LBD. There appeared to be a weight threshold at 25 kg at which spinal loads became increasingly risky. The accuracy of the spinal load estimates was found to be influenced by trunk dynamics and trunk muscle coactivity. Models that do not account for trunk dynamics and corresponding muscle coactivity might underrepresent the spinal loads during various lifting exertions. Thus, it is important to consider the interaction between trunk kinematics and the weight that is lifted when estimating spinal loads.

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