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An investigation of perceived exertion via whole body exertion and direct muscle force indicators during the determination of the maximum acceptable weight of lift

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The objective of this study was to identify the perceived exertion mechanisms (direct muscle force and whole body exertion) associated with the decision to change the weight of lift during the determination of the maximum acceptable weight of lift (MAWL). Fifteen males lifted a box of unknown weight at a rate of 4.3 lifts/min, and adjusted the weight until their MAWL was reached. Variables such as the predicted muscle forces and heart rate were measured during the lifting exertion, as well as the predicted spinal loading in three dimensions using an EMG-assisted biomechanical model. Multiple logistic regression techniques were used to identify variables that were associated with the decision to change the weights up and down prior to a subsequent lift. Results indicated that the force in the left erector spinae, right internal oblique, and left latissimus dorsi muscles as well as heart rate were associated with decreases in the weight prior to the next lift. It appears that a combination of local factors (muscle force) and whole body exertion factors (heart rate) provide the feedback for the perceived exertion when decreasing the weight. The up-change model indicated that the forces of the right erector spinae, left internal oblique, and the right latissimus dorsi muscles were associated with the decision to increase the weight prior to the next lift. Thus, local factors provide feedback during the decision to increase the weight when starting from light weights. Collectively, these findings indicate that psychophysically determined weight limits may be more sensitive to muscular strain rather than spinal loading.

1. Introduction

It has been widely reported that 80% of the working population will experience low-back pain (LBP) at some time during their life (Spengler *et al.* 1986). Guo *et al.* (1995) found that the 12-month prevalence of LBP for workers in the USA was 17.6% . Thus, LBP affects a significant proportion of the working population during any one period of time. One approach that has been used to reduce LBP in the workplace is the method of psychophysical maximum acceptable weight of lift (MAWL) (Snook 1978, Snook and Ciriello 1991). Snook and Ciriello (1991) have developed extensive tables of manual materials handling (MMH) tasks that provide magnitudes of loads that are acceptable to a given percentage of the working

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population. These loads were determined by having experienced MMH workers monitor their own feelings of exertion or fatigue and adjust the weight accordingly (Snook 1978). According to Snook (1978), only the individual can integrate the sense of the various stresses associated with MMH tasks into a meaningful response (e.g. perceived exertion). The major assumptions behind the psychophysical approach are that individuals can perceive when a given load will increase the risk of low-back disorders (LBDs) and that individuals can perceive when a load is safe and will reduce the risk of injury (Herrin *et al*. 1986, Gamberale 1990).

Although some evidence exists on the efficacy of the psychophysical approach (Snook 1978), little is known about how the 'acceptable' loads are derived. Thompson and Chaffin (1993) reported that there was no correlation between perceived exertion and compressive force on the lumbosacral joint. Karwowski and Ayoub (1984) reported that the combined effect of biomechanical and physiological stress leads to the overall perception of exertion. Chaffin and Page (1994) and Jorgensen *et al*. (1999) found that the compressive loads on the spine for MAWLs were above the recommended NIOSH tolerance limits (3400 N). Also, Jorgensen *et al*. (1999) found that spinal loads (i.e. compression and shear forces) were not associated with changing the weight toward the MAWL during the psychophysical decision process. It appears that individuals adjusted the weights based on cues other than loading on the spine and thus psychophysically determined limits may not be protective of discogenic injuries.

The lack of association between spinal loading cues and subsequent changing of the weight may be a result of the lack of significant nerve endings (nociceptors) in the disc region as hypothesized by Jorgensen *et al*. (1999), or that the individuals may be regulating the weight based upon 'feelings of sensation and fatigue' of the muscles (Snook 1978). Thus, the determination of the MAWLs may be based upon the perception of muscular force sensation and fatigue. There has been a considerable amount of research on the perception of muscle force, although controversy exists about where the sensation of force originates from within the body.

Some researchers have found support for the existence of a feedforward or central mechanism that corresponds to sensations of whole body exertion (Roland 1975, Gandevia and McCloskey 1977, 1978, McCloskey 1978, Killian *et al*. 1979, Cafarelli 1982, Matthews 1982, Jones and Hunter 1983). According to this model, individuals are able to sense muscle force/fatigue through a central mechanism that reacts to stresses in the cardiovascular system. The central mechanism (whole body exertion) is a direct response to the increased need of oxygen and nutrients, which must be delivered to the muscles through the cardiovascular system (via the blood). This mechanism has typically been evaluated by measuring heart rate, oxygen consumption, or metabolic rate (Ekblom and Goldbarg 1971, Pandolf 1978, 1983, Legg and Myles 1981, Mihevic 1981, Robertson 1982, Nicholson and Legg 1986).

A second hypothesis is that force sensation originates locally (direct muscle force sensation) or via a feedback mechanism (Mihevic 1981, Cafarelli 1982, 1988, Cafarelli and Layton-Wood 1986), where the local sensation mechanism relies upon corollary discharges from muscle spindles and Golgi tendon organs (McCloskey 1978, Mihevic 1981, Gandevia 1982, Pandolf 1983, Cafarelli 1988). These receptors respond to muscle activity, stretching of the muscle, and intramuscular force, where the Golgi tendon organs are more sensitive than muscle spindles to intramuscular force (McCloskey 1978). Thus, the local feedback mechanism relates to the strain in the muscles where the whole body exertion mechanism would encompass stress on the cardiovascular system.

Finally, a third hypothesis is that these sensation mechanisms either work together or separately depending on the muscle group being exerted (Ekblom and Goldbarg 1971, Borg and Noble 1974, Pandolf 1978, 1983, Jones and Hunter 1985). Ekblom and Goldbarg (1971) suggested that the mechanism (whole body exertion versus local) dominating the perception of the exertion depends on the size of the muscle (larger muscles driving whole body exertion mechanism and smaller muscles relying on a direct muscle force sensation mechanism). Robertson (1982) has proposed that the direct muscle force sensation or local factors provide the primary force sensation and the whole body exertion factors (cardiovascular) act as amplifiers of the local signals. It may very well be this combination that individuals are responding to during the determination of MAWLs.

It is hypothesized, therefore, that the psychophysical methodology may be addressing perceived exertions related to sensations with activities of the muscle rather than spinal loading under the assumption that muscle activity is related to the tension in the muscle. Exertions requiring elevated muscle tension have been found to result in muscle damage (Armstrong 1984, 1986, 1990, Armstrong *et al.* 1991). Thus, psychophysically determined acceptable weight limits may be more appropriately used as a control of muscle strain that would indicate acute LBD compared to spinal loading associated with chronic LBD. Therefore, the objective of this study was to identify whether muscle force sensation and/or cardiovascular exertion (as indicated by heart rate) were associated with the MAWL determination process.

2. Methods

2.1. *Participants*

Fifteen male college students participated in this study, with a mean (SD) age of 22.5 (2.0) years, and a mean (SD) height and weight of 109.1 (4.5) cm and 73.4 (6.6) kg, respectively. All participants were inexperienced in manual materials handling and none reported a current episode of low-back pain.

2.2. *Experimental design*

The experimental design consisted of a repeated measures approach, where each participant was subjected to each of the experimental conditions. To address the objective of identifying variables that may influence an individual's decision to either change or not change the weight prior to the next lift, logistic regression techniques were used. Logistic regression techniques are appropriate in this case as it was desired to model a binary dependent variable, such as 'change' or 'no-change' in weight, and the independent variables could be either categorical or continuous. The logistic regression models were restricted to the first eight lifts (trials) for each of the lifting conditions. This range was used since most of the changes of the weight occurred within the first eight lifts, and by not including all the no-change trials (which signalled the end of the lifting condition), the resulting logistic regression models were not artificially influenced by an excessive number of no-change trials. Additionally, to reduce any confounding or masking effect of the independent variables due to the direction of the weight change (up or down), the conditions with the five highest initial weights were used to assess the down changes of weight, and the conditions with the five lowest initial weights were used to assess the up changes of weight. This approach was considered appropriate as most psychophysical studies are carried out by starting participants at both high and low weights, and having them adjust toward their MAWL.

The dependent variable consisted of a dichotomous change of weight variable (i.e. change up and no-change for assessing the increases of the weight, and change down and no-change when assessing the decreases in the weight). The independent variables consisted of the categorized and standardized predicted maximum muscle forces, maximum spinal moments, predicted maximum forces on the L_5/S_1 joint and the heart rate. The standardization and categorization processes are discussed in § 2.6. Additionally, it was hypothesized that the initial weight and lifting trial number might be confounders, as they might be associated with the decision to change or not change the weight, as well as with the resulting magnitudes of the independent variables. Therefore, all logistic regression models were generated while controlling for initial weight and lifting trial effects. Since multiple observations were obtained from each participant, the participant effect was also controlled for when building the logistic regression models.

2.3. *Experimental task*

The study was carried out using a modified psychophysical procedure. Ten lifting conditions were performed by each participant, with each condition beginning at a different initial weight. Five of the lifting conditions began with loads greater than the estimated MAWL for this type of task, while the remaining five lifting conditions began at loads less than the estimated MAWL (Ciriello *et al*. 1990, 1993). The participants were permitted to add or remove as much weight from the box as desired between lift trials, and continued to lift the box until the weight was unchanged for eight consecutive lift trials. For the purposes of this study, this weight was defined as the MAWL. The participants, however, were not aware of this criterion for ending the lifting condition. Since electromyography (EMG) was being used in this study, the use of this modified psychophysical approach reduced the chance for fatigue which, if present, would alter the EMG signal. To simulate an MMH task, the subjects lifted a box from knee height, transferred it a distance of 1.52 m, and placed it on a shelf at elbow height. The lift rate was 4.3 lifts/min, which has been used in previous psychophysical studies (Ciriello *et al*. 1990, 1993, Snook and Ciriello 1991).

2.4. *Apparatus*

The participants moved a box of dimensions $25.4 \times 42.5 \times 32.4$ cm (height × width × depth). The handles were located 20.3 cm from the bottom of the box. The weights consisted of 42 kg of metal filings separated into 0.91 kg packages of similar size and shape. The box was similar in size to the large box used in the studies of Snook and Ciriello (1991).

A Lumbar Motion Monitor (LMM), which is essentially an exoskeleton of the spine, was used to collect three-dimensional kinematic trunk variables (Marras *et al*. 1992, 1993). Participant heart rate was obtained by use of a Polar Favor Heart Rate Monitor (Polar CIC Inc., Port Washington, NY). The monitor transmitted the heart rate to a digital readout on a wrist receiver.

Electromyographic activity was collected through the use of bipolar silver – silver chloride surface electrodes spaced approximately 3 cm apart over 10 trunk muscles (Mirka and Marras 1993). The 10 trunk muscles included: right and left erector spinae (RES and LES); right and left latissimus dorsi (RLAT and LLAT); right and left internal obliques (RIOB and LIOB); right and left external obliques (REOB and LEOB); and right and left rectus abdominis (RABD and LABD).

The EMG-assisted biomechanical model used to predict the muscle forces and spinal loading (Marras and Sommerich 1991a,b, Granata and Marras 1993, 1995, Marras and Granata 1995, 1997) requires calibration exertions using a force plate (Bertec 4060A, Worthington, OH) and an L_5/S_1 locator (Fathallah *et al.* 1997) to determine participant-specific muscle gain. Using methods developed by Fathallah *et al.* (1997), the participant-specific muscle gain was determined. The magnitude of the muscle gain represents 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 forces and moments for the experimental task, and allowed the participants to move without being restricted to a force plate.

All signals from the above equipment (except heart rate as noted below) were collected simultaneously through customized Windows™-based software developed in-house. The signals were collected at 100 Hz and recorded on a 486 portable computer via an analog-to-digital conversion board.

2.5. *Experimental procedure*

Surface electrodes were applied to the trunk muscles specified above using standard placement procedures (Marras 1990). The heart rate transmitter was placed across the participant's chest at the level of the xyphoid process. The participant was then placed in a structure that allowed maximum voluntary contractions (MVCs) of the trunk to be performed in six directions (Mirka and Marras 1993). All subsequent EMG data for the calibration exertions and the experimental tasks were normalized to these MVCs. To reduce fatigue effects, a 2-min rest period was given after every MVC (Caldwell *et al.* 1974).

The LMM was then placed on the participant's back and calibration exertions were performed with the participant standing on the force plate. These sagittally symmetric exertions required the participant to lift a 22.7 kg box from knee height to elbow height. Five calibration exertions were performed at the beginning, at the midpoint (after five lifting conditions), and at the end of the experiment.

After completing the first set of calibration exertions, the participant read the experimental instructions (appendix), which were also repeated verbally to ensure comprehension. A computer-generated tone was a signal to the subject to perform each lift. The participants were able to lift using any style (e.g. freestyle lifting). The box was returned to the starting position by an experimenter, and the participant was permitted to make any desired changes to the weights in the box before the next tone sounded. The heart rate was recorded at the completion of each lift, as well as the amount of weights in the box, which was measured by a force plate.

Each of the 10 lifting conditions began at a different weight. Initial weights of 9.1, 11.8, 14.5, 17.2, 20.0, 29.9, 32.7, 35.4, 38.1 and 41.7 kg were presented in random order to each participant. Participants were required to attempt to lift each weight, even if a lift or placement on the shelf could not be completed. After the attempt, the subject was allowed to change the weight.

2.6. *Statistical analysis*

Descriptive statistics were generated to describe the maximum muscle forces, heart rate and the spinal loading (moments and forces) for each of the 10 lifting conditions across the last eight lifts of each condition. This identifies the magnitude and variability of each dependent variable at the MAWL.

For the logistic regression analyses, each independent variable was categorized by identifying cut-off values that best separated the trials with no-changes from the trials with changes (acceptable and unacceptable categories). In order to identify a common cut-off value that would be independent of inter-participant variability for each independent variable, the data for each independent variable was standardized to the mean and standard deviation of the trials of the respective independent variable at the MAWL, which has been described previously (Jorgensen *et al.* 1999). This standardization was also performed to allow the data to be interpreted in reference to a common point (i.e. the participant's MAWL).

Thus, each independent variable was standardized by the following equation:

$$X_s = \frac{X_{ij} - X_{mawl}}{S_{mawl}}$$

where:

X_s = the standardized independent variable for lifting condition i and participant j;
X_{ij} = the measured independent variable from each participant for lifting condition i and lifting trial j;
X_{mawl} = the mean of the variable across the MAWL trials for each participant; and
S_{mawl} = the standard deviation of the variable across the MAWL trials for each participant.

The standardized variables were then interpreted as follows: values of zero correspond to the mean of the variable at the MAWL trials, while values of + 1.0 represent values that are 1 SD (of the MAWL trials) greater than the mean of the MAWL trials. Similarly, values of – 1.0 represent values that are 1 SD (of the MAWL trials) less than the mean of the MAWL trials. Each independent variable was then categorized by selecting a cut-off value and assigning all standardized values greater than the cut-off a numerical value of 1, and all standardized values less than the cut-off a numerical value of 0. The nine cut-offs for each independent variable were determined by selecting a value ranging from – 2.0 to + 2.0 SDs around the MAWL mean, in 0.5 SD increments. Thus, the resulting lifting trials were categorized into binary 0 and 1 data, for logistic regression purposes.

Initially, univariate logistic regression was performed to assess the individual associations in terms of the odds ratios of changing the weight up or down, independently, versus not changing the weight. Stepwise logistic regression was used to determine which cut-off value was to be used for each independent variable. Wald χ^2 tests were used to assess the significance of each independent variable for the univariate logistic regression models, with a significance level of $\alpha = 0.05$. To build the multiple logistic regression models, Wald χ^2 tests and χ^2 tests on the deviance were used to assess the significance of additional variables entered into the models. The fit of the model was assessed by the Hosmer-Lemeshow goodness-of-fit test (Hosmer and Lemeshow 1989). Additionally, the predictive ability of the final model was determined by evaluating the Goodman-Kruskal γ statistic. All statistical analyses were performed using the SAS Institute (1989) statistical software.

3. Results

3.1. *Descriptive statistics at the MAWL*

The estimated mean (SD) maximum muscle forces at the MAWL as a function of the 10 different lifting conditions are shown in table 1. Generally, the erector spinae and internal obliques resulted in the highest estimates of muscle force, whereas the rectus abdominis and external obliques resulted in the least muscle force. The resulting spinal loading and heart rate at the MAWL are shown for each lifting condition in table 2. Generally, the final mean MAWLs across all 15 participants were very similar for all 10 conditions, ranging from 24.3 to 28.9 kg. The predicted lateral moment was also very similar across the 10 conditions. The mean maximum spinal forces ranged from 561.5 N (9.1 kg initial weight) to 809.9 N (29.9 kg initial weight) for lateral shear, 1091.1 N (9.1 kg initial weight) to 1499.3 N (20.0 kg initial weight) for anterior/posterior (A/P) shear, and 5174.4 N (9.1 kg initial weight) to 5958.8 N (20.0 kg initial weight) for compressive force on the L_5/S_1 intervertebral disc. Mean heart rate was very consistent across all the conditions, ranging from 119 beats per minute (bpm) to 125 bpm.

3.2. *Univariate logistic regression results predicting changes in weight*

Table 3 shows the results of the univariate logistic regression for the odds ratio (OR) of changing the weight up versus no change in the weight while controlling for participant, lifting trial and initial weight effects. All univariate models except for the lateral shear force and heart rate were significant at the $\alpha = 0.05$ level. Odds ratios of less than 1.0 are interpreted as a decrease in the odds that the participant changes the weight up versus not changing the weight prior to the next lift. Interpreted in another way, by taking the inverse of the OR, it can also be interpreted as an increase in the odds of not changing the weight versus changing the weight up prior to the next lift. For example, an odds ratio of 0.13 for right erector spinae force can be interpreted as a decrease in the likelihood of changing the weight up when the muscle force is greater than 2.0 SDs below the MAWL mean. Likewise, by taking the inverse of the 0.13 OR, it can also be interpreted as a 7.69 times increase in the odds of not changing the weight the muscle force is greater than 2.0 SDs below the MAWL mean.

Table 4 shows the results of the univariate logistic regression for the ORs of changing the weight down versus no change in the weight prior to a subsequent lift. All dependent variables were significant at the $\alpha = 0.05$ level. The model variables should be interpreted in the same way as in table 3, except that the ORs represent the likelihood of making a change down versus not changing the weight before the next lift, given the levels of the independent variables.

3.3. *Multiple logistic regression models predicting changes of weights*

As shown in table 5, the final multiple logistic regression model for predicting the down-changes of weight includes maximum muscle forces of left erector spinae, right internal oblique and left latissimus dorsi muscles, and heart rate.

The final multiple logistic regression model for predicting the up-changes of weight included the maximum muscle force for the right erector spinae and latissimus dorsi, the left internal oblique, and the lateral trunk moment (table 6).

The performance and internal validity of the two multiple logistic regression models were evaluated using a rank correlation of the concordant and discordant pairs (Goodman-Kruskal γ), as well a goodness-of-fit test to assess the predictive

Table 1. Mean (SD) maximum muscle force at the MAWL for each initial weight condition.

Condition: initial weight (kg)	MAWL (kg)	Maximum muscle force (N)									
		RLAT	LLAT	RES	LES	RABD	LABD	REOB	LEOB	RIOB	LIOB
9.1	24.9	584.4	472.9	1496.8	1440.6	21.1	31.5	177.2	176.2	1121.0	862.9
		(469.0)	(365.6)	(1032.0)	(602.6)	(19.9)	(41.8)	(109.7)	(180.9)	(1268.1)	(463.2)
11.8	26.9	621.2	455.7	1806.6	1557.7	21.4	26.8	181.9	224.1	1377.2	942.0
		(526.2)	(398.2)	(1173.8)	(623.4)	(17.9)	(26.8)	(100.6)	(197.8)	(1688.8)	(495.2)
14.5	24.3	661.0	427.7	1850.6	1408.4	21.1	27.7	163.0	159.1	1460.7	817.2
		(538.8)	(337.2)	(1246.1)	(586.5)	(20.1)	(30.6)	(93.1)	(87.5)	(1710.7)	(470.3)
17.4	25.3	557.0	442.6	1690.3	1517.1	26.9	32.5	219.4	249.0	1232.3	880.6
		(459.2)	(326.6)	(837.9)	(753.9)	(24.2)	(44.2)	(207.5)	(430.0)	(872.3)	(556.6)
20.0	27.0	929.4	769.0	1788.4	1494.1	40.0	29.4	154.6	297.9	1393.3	838.9
		(1215.9)	(1260.9)	(1126.5)	(657.1)	(69.5)	(26.2)	(69.9)	(827.4)	(1447.3)	(419.7)
29.9	26.1	742.2	417.4	1792.8	1431.0	24.6	33.9	149.6	257.3	1387.6	821.7
		(802.2)	(320.0)	(1107.6)	(591.5)	(19.3)	(40.5)	(78.1)	(729.4)	(1415.2)	(464.2)
32.7	27.2	798.8	438.8	1965.8	1467.6	25.6	27.2	233.2	284.2	1548.5	818.7
		(600.0)	(258.2)	(1211.1)	(691.5)	(23.4)	(21.7)	(299.8)	(508.5)	(1334.6)	(428.7)
35.4	28.9	601.3	351.6	1846.0	1473.8	65.4	26.6	215.7	172.3	1476.9	810.7
		(436.0)	(133.7)	(1278.2)	(642.4)	(175.2)	(25.5)	(286.3)	(98.8)	(1614.8)	(394.6)
38.1	28.6	583.2	546.0	1844.1	1493.9	22.7	29.3	176.9	176.4	1366.8	871.0
		(489.8)	(591.2)	(1160.4)	(681.7)	(19.2)	(23.5)	(90.7)	(101.7)	(1317.8)	(465.9)
41.7	27.1	597.1	392.7	1890.1	1420.8	34.7	32.1	190.1	226.7	1527.4	837.1
		(408.5)	(162.9)	(1187.5)	(667.2)	(54.0)	(40.7)	(139.1)	(186.3)	(1468.9)	(431.1)

RLAT: right latissimus dorsi; LLAT: left latissimus dorsi; RES: right erector spinae; LES: left erector spinae; RABD: right rectus abdominis; LABD: left rectus abdominis; REOB: right external oblique; LEOB: left external oblique; RIOB: right internal oblique; LIOB: left internal oblique.

Table 2. Mean (SD) MAWL, heart rate and spinal loading for each initial weight (mean across 15 subjects) at the eight MAWL trials.

Condition: initial weight (kg)	MAWL (kg)	Heart (bpm)	Spinal loading: Lateral moment (Nm)	Spinal loading: Lateral shear force (N)	Spinal loading: A/P shear force (N)	Spinal loading: Compression force (N)
9.1	24.9	123	84.3	561.5	1091.1	5174.4
	(7.3)	(16)	(87.7)	(366.6)	(683.5)	(2226.2)
11.8	26.9	120	99.5	590.0	1188.1	5697.2
	(7.4)	(17)	(100.2)	(398.8)	(639.9)	(2307.5)
14.5	24.3	122	88.7	645.1	1161.0	5482.9
	(7.4)	(17)	(90.6)	(497.7)	(712.2)	(2556.1)
17.4	25.3	121	75.7	600.7	1158.9	5367.6
	(6.6)	(16)	(62.1)	(427.4)	(606.4)	(2229.8)
20.0	27.0	119	93.8	756.3	1499.3	5958.8
	(7.5)	(19)	(91.3)	(624.0)	(1315.9)	(2675.2)
29.9	26.1	122	92.9	809.9	1185.5	5592.4
	(7.1)	(16)	(93.7)	(656.8)	(695.2)	(2396.6)
32.7	27.2	119	97.8	761.5	1300.2	5947.2
	(6.9)	(15)	(78.0)	(518.5)	(756.9)	(2340.0)
35.4	28.9	121	89.4	693.1	1181.5	5712.1
	(8.6)	(17)	(78.8)	(440.1)	(582.1)	(2324.4)
38.1	28.6	124	95.2	667.9	1196.4	5693.1
	(8.0)	(16)	(78.2)	(446.8)	(602.1)	(2060.2)
41.7	27.1	125	81.5	623.3	1136.1	5563.0
	(9.6)	(17)	(63.4)	(424.9)	(428.5)	(1966.4)

ability of the model. As shown in tables 5 and 6, both the up-change ($\gamma = 0.773$) and down-change ($\gamma = 0.802$) logistic regression models resulted in fairly high γ values indicating that the models resulted in good predictability when applied to the data. However, addition of the biomechanical variables for the up-change model resulted in very little additional predictability (an increase of 10%), as compared to the down-change model after the biomechanical variables and heart rate were added to the model (an increase of 45%). Finally, the internal validity of both multiple logistic regression models was deemed adequate using the Hosmer-Lemeshow goodness-of-fit test (Hosmer and Lemeshow 1989).

4. Discussion

In previous work by Jorgensen *et al.* (1999), a lack of association was found between measures of spinal loading and changing the load weight prior to the next lift during the psychophysical method of determining an 'acceptable' load. The present study also found this result when using a different subset of variables (muscle forces, heart rate and spinal loads). In the present study, the compression and shear forces on the spine were found to be less associated with the changes in weight (both up-change and down-change models) than the muscle forces. If spinal loading is assumed to be the mechanism of injury (NIOSH 1981, Jager and Luttmann 1991, McGill 1997), then this lack of association in combination with high spinal loading at the MAWL (Chaffin and Page 1994, Jorgensen *et al.* 1999) indicates that the psychophysical approach may not be protective of low-back disc injuries.

Table 3. Univariate logistic regression model parameters for the odds of changing up versus the odds of no change of weight, while controlling for participant, lifting trial and intial weight effects. All variables are significant at $p \leqslant 0.05$ *except* for the shaded rows.

Variable	Cut-off value*	Parameter (β)	Standard error	p-value	Odds† ratio
Compression force (N)	– 1.5	– 1.3206	0.3914	0.0007	0.27
Lateral shear force (N)	1.5	– 0.8043	0.5284	0.1280	0.45
Sagittal shear force (N)	0.0	– 0.5215	0.2392	0.0292	0.59
Sagittal moment (Nm)	– 1.5	– 1.4251	0.3458	0.0001	0.24
Lateral moment (Nm)	– 1.0	– 0.7851	0.2843	0.0058	0.46
Twisting moment (Nm)	– 1.0	– 0.9033	0.2900	0.0018	0.41
Resultant moment (Nm)	– 1.5	– 1.6049	0.3855	0.0001	0.20
LLAT force (N)	– 1.5	– 1.0367	0.3475	0.0029	0.36
RLAT force (N)	0.5	– 1.1118	0.2784	0.0001	0.33
LES force (N)	– 1.5	– 1.3665	0.3623	0.0002	0.26
RES force (N)	– 2.0	– 2.0317	0.4633	0.0001	0.13
RABD force (N)	– 1.0	– 0.8663	0.2840	0.0023	0.42
LABD force (N)	– 1.0	– 0.9684	0.2982	0.0012	0.38
LEOB force (N)	– 1.0	– 0.7071	0.2519	0.0050	0.49
REOB force (N)	– 1.5	– 0.8229	0.3116	0.0083	0.44
LIOB force (N)	– 1.5	– 1.6768	0.3395	0.0001	0.19
RIOB force (N)	– 1.0	– 1.0782	0.2372	0.0001	0.34
Heart rate (bpm)	1.0	– 1.1847	0.9365	0.2059	0.31

*Number of standard deviations above or below the average at the MAWL.

†The odds ratio refers to the odds of changing the weight up versus the odds of not changing the weight prior to the next lift. Odds ratio greater than 1.0 indicate an increased likelihood for increasing the weight before the next lift, and an odds ratio of less than 1.0 indicates a decreased likelihood for increasing the weight before the next lift.

LLAT: left latissimus dorsi; RLAT: right latissimus dorsi; LES: left erector spinae; RES: right erector spinae; RABD: right rectus abdominis; LABD: left rectus abdominis; LEOB: left external oblique; REOB: right external oblique; LIOB: left internal oblique; RIOB: right internal oblique.

The final down-change multiple logistic regression model contained the forces from three muscles (left erector spinae, right internal oblique and left latissimus dorsi) and heart rate. The association between decreasing the weight prior to the next lift and the resulting muscle force and heart rate support previous work that suggests that direct muscle force sensation and whole body exertion mechanisms work in conjunction to provide an overall judgement of perceived exertion (Ekblom and Goldbarg 1971, Borg and Noble 1974, Pandolf 1978, 1983, Jones and Hunter 1985).

Similar to the down-change model, the up-change multiple logistic regression model also contained the forces from three muscles (right erector spinae, left internal oblique and right latissimus dorsi). Lateral moment was also found to be associated with increases in the load weight prior to the next lift. This model revealed that the decision process to increase the weight (starting from light weights) versus decreasing the weight (starting from heavy load weights) may rely on different mechanisms. The absence of an association between heart rate and increases of the weight suggests that the perceived exertion may be based solely on direct muscle force sensation rather than whole body exertion mechanisms or the combination of the two. This is consistent with previous results which have found that local factors (i.e. direct

Table 4. Univariate logistic regression model parameters for the odds of changing down versus the odds of no change of weight, while controlling for participant, lifting trial and initial weight effects. All variables are significant at $p \leqslant 0.05$.

Variable	Cut-off value*	Parameter (β)	Standard error	p-value	Odds† ratio
Compression force (N)	0.5	1.2003	0.1986	0.0001	3.32
Lateral shear force (N)	− 1.0	0.8464	0.3563	0.0175	2.33
Sagittal shear force (N)	1.5	1.1593	0.2722	0.0001	3.19
Sagittal moment (N)	1.0	1.4002	0.2047	0.0001	4.06
Lateral moment (N)	− 1.0	1.0683	0.3076	0.0005	2.91
Twisting moment (N)	0.0	0.7274	0.1935	0.0002	2.07
Resultant moment (N)	1.0	1.2584	0.2072	0.0001	3.52
LLAT force (N)	1.5	1.6401	0.2272	0.0001	5.16
RLAT force (N)	− 0.5	1.5762	0.2626	0.0001	4.84
LES force (N)	1.0	1.8803	0.2232	0.0001	6.56
RES force (N)	1.0	1.7748	0.2162	0.0001	5.90
RABD force (N)	− 2.0	− 1.9728	0.6556	0.0026	0.14
LABD force (N)	− 1.5	− 1.7730	0.3485	0.0001	0.17
LEOB force (N)	2.0	0.9646	0.2372	0.0001	2.62
REOB force (N)	0.5	1.0853	0.1965	0.0001	2.96
LIOB force (N)	1.5	1.3152	0.2358	0.0001	3.73
RIOB force (N)	0.0	1.7709	0.2302	0.0001	5.88
Heart rate (bpm)	1.0	1.9239	0.3000	0.0001	6.85

*Number of standard deviations above or below the mean at the MAWL.

†The odds ratio refers to the odds of changing the weight down versus the odds of not changing the weight prior to the next lift. Odds ratio greater that 1.0 indicate an increased likelihood for decreasing the weight before the next lift, and an odds ratio of less than 1.0 indicates a decreased likelihood for decreasing the weight before the next lift.

LLAT: left latissimus dorsi; RLAT: right latissimus dorsi; LES: left erector spinae; RES: right erector spinae; RABD: right rectus abdominis; LABD: left rectus abdominis; LEOB: left external oblique; REOB: right external oblique; LIOB: left internal oblique RIOB: right internal oblique.

Table 5. Multiple logistic regression model parameters for the odds of changing down versus the odds of no change of weight prior to the next lift during the first eight lifting trials (γ= 0.773)

Variable	Cut-off value*	Parameter (β)	Standard error	p-value	Odds† ratio	95% CI for odds ratio†
Intercept	–	− 1.5931	1.0074	0.1212	–	–
Subject	–	− 0.0064	0.0258	0.8066	0.99	0.95–1.05
Initial weight	–	0.0573	0.0276	0.0436	1.06	1.00–1.12
Lift trial	–	− 0.5989	0.0612	0.0001	0.55	0.49–0.62
LES (N)	1.0	1.3916	0.2444	0.0001	4.02	2.49–6.49
RIOB (N)	0.0	1.3393	0.2505	0.0001	3.82	2.34–6.24
Heart rate (bpm)	1.0	1.6851	0.3472	0.0001	5.39	2.73–10.65
LLAT (N)	1.5	1.1228	0.2574	0.0001	3.07	1.86–5.09

*Number of standard deviation above or below the mean at the MAWL.

†The odds ratio refers to the odds of changing the weight down versus the odds of not changing the weight prior to the next lift. Odds ratio greater than 1.0 indicate an increase likelihood for decreasing the weight before the next lift, and an odds ratio of less than 1.0 indicates a decreased likelihood for decreasing the weight before the next lift.

LES: left erector spinae; RIOB: right internal oblique; LLAT: left latissimus dorsi.

Table 6. Multiple logistic regression model parameters for the odds of changing up versus the odds of no change of weight prior to the next lift during the first eight lifting trials (γ= 0.807)

Variable	Cut-off value*	Parameter (β)	Standard error	p-value	Odds† ratio	95% CI for odds ratio†
Intercept	–	9.8090	0.8915	0.0001	–	–
Subject	–	– 0.1841	0.0295	0.0001	0.83	0.79–0.88
Initial weight	–	– 0.1878	0.0327	0.0001	0.83	0.78–0.88
Lift trial	–	– 0.5222	0.0646	0.0001	0.59	0.52–0.67
RES (N)	– 1.5	– 1.3785	0.3278	0.0001	0.25	0.13–0.48
LIOB (N)	– 1.5	– 1.5145	0.3594	0.0001	0.22	0.11–0.45
RLAT (N)	0.5	– 0.9346	0.2937	0.0015	0.39	0.22–0.70
Lateral moment (Nm)	– 1.0	– 0.7878	0.3036	0.0095	0.46	0.25–0.82

*Number of standard deviations above or below the mean at the MAWL.

†The odds ratio refers to the odds of changing the weight up versus the odds of not changing the weight prior to the next lift. Odds ratio greater than 1.0 indicate an increased likelihood for increasing the weight before the next lift, and an odds ratio of less than 1.0 indicates a decreased likelihood for increasing the weight before the next lift.

RES: right erector spinae; LIOB: left internal oblique; RLAT: right latissimus dorsi.

Table 7. Mean (SD) percentage time (% time) into the exertion where the maximum muscle force occurred

Weight change	RLAT	LLAT	RES	LES	RABD	LABD	REOB	LEOB	RIOB	LIOB
Change down	0.62 (0.27)	0.41 (0.32)	0.19 (0.20)	0.17 (0.15)	0.66 (0.19)	0.66 (0.20)	0.61 (0.18)	0.61 (0.21)	0.27 (0.27)	0.22 (0.22)
No change	0.57 (0.30)	0.42 (0.31)	0.21 (0.22)	0.19 (0.17)	0.64 (0.21)	0.62 (0.22)	0.61 (0.21)	0.62 (0.21)	0.24 (0.25)	0.19 (0.18)
Change up	0.56 (0.30)	0.48 (0.32)	0.22 (0.21)	0.17 (0.17)	0.66 (0.21)	0.64 (0.22)	0.59 (0.21)	0.61 (0.21)	0.26 (0.26)	0.16 (0.17)
No change	0.58 (0.31)	0.47 (0.32)	0.18 (0.18)	0.15 (0.13)	0.66 (0.21)	0.66 (0.22)	0.61 (0.20)	0.64 (0.20)	0.22 (0.25)	0.18 (0.19)

RLAT: right latissimus dorsi; LLAT: left latissimus dorsi; RES: right erector spinae; LES: Left erector spinae; RABD: right rectus abdominis; LABD: left rectus abdominis; REOB: right external oblique; LEOB: left external oblique; RIOB: right internal oblique; LIOB: left internal oblique.

muscle force indicators) play the primary role in force sensation (Mihevic 1981, Cafarelli 1982, 1988, Cafarelli and Layton-Wood 1986). The down-change model, however, included the heart rate as well as forces from three muscles, which is consistent with research that suggests that the combination of direct muscle force sensation and whole body exertion factors is responsible for perceived exertion. The difference between the two models indicates that the weight of the load may influence which factors contribute to the perceived exertion. That is, increasing the weight in a psychophysical methodology may rely upon the sensations from the local (direct muscle force sensation) rather than the central (whole body exertion) mechanism.

Inspection of both the up-change and down-change multiple logistic regression models indicates that both models contain an association between changing the weight prior to the next lift and the magnitude of muscle force from similar muscle groups (erector spinae, internal obliques and latissimus dorsi). The local feedback mechanism (direct muscle force) has been hypothesized to be the central nervous system reaction to the discharge of the Golgi tendon organs and to a lesser extent muscle spindles (McCloskey 1978, Mihevic 1981, Cafarelli 1982). Golgi tendon organs and muscle spindles serve as receptors for kinesthetic sensibility, that is, the sense of position and action of various body parts (McCloskey 1978). Muscle activity and muscle stretch cause both Golgi tendon organs and muscle spindles to discharge but Golgi tendon organs have been found to react more to intramuscular force (McCloskey 1978). Thus, given that the initial position of the box remained constant, it is most likely that the Golgi tendon organs rather than muscle spindles are providing the underlying mechanism for the perception of the exertion since several muscle forces were found to be significantly associated with changing the weight of the box.

Furthermore, Ekblom and Goldbarg (1971) suggested that the size of the muscle determined whether or not whole body exertion factors would dominate the perceived exertion. However, the results of the present study do not support this hypothesis, as both models contained forces from the same muscle groups, and the three muscle groups vary considerably in cross-sectional area, with the erector spinae being larger than the latissimus dorsi, and the latissimus dorsi being larger than the internal oblique (McGill *et al.* 1993). The main difference between the two models is that when decreasing the weight, a whole body exertion factor (heart rate) was included in the down-change model. This may indicate that the required level of exertion dictated by the higher starting weight may cause the heart rate to increase, thus contributing to the perceived exertion.

The results may also indicate, however, that the muscle groups primarily responsible for the type of exertion within a complex task may be responsible for the perceived exertion. As shown in table 7, the erector spinae and internal oblique muscles exerted maximally during the first part of the task (about 20% into the task). Thus, the peak forces of the erector spinae and internal oblique muscles were the result of lifting the box off the shelf. As these muscles are responsible for extension of the trunk, the participants may be responding to sensations from these muscles during the lifting phase. Similarly, the maximum muscle force for the right latissimus dorsi occurred more than half-way into the task (about 60% into the task). This indicates that the participants may have been starting to lift the box high enough to place it on the shelf at elbow height, thus using the latissimus dorsi muscles to raise the box further. The left latissimus dorsi muscle had its peak force at approximately the middle of the lift (40 to 48% into the task) indicating that this muscle peaked during the carrying of the load. This muscle force might be a result of stabilizing the trunk during the asymmetric nature of the carrying of the box.

The presence of multiple muscles in the final models for a complex task also indicates that during a psychophysical adjustment of weight, the individual reacts to forces from multiple muscles and not to any single group. This suggests that it is extremely important to consider multiple muscles (instead of a single muscle) when performing task evaluation through EMG. Since the task being performed was complex in nature (lifting, carrying, and lowering with asymmetry), the individual must rely upon multiple muscles to complete the task and thus simple muscle models would drastically underestimate the loads during the task (Marras and Sommerich 1991b).

Finally, when the two multiple logistic regression models are compared, another major difference becomes apparent: the muscles represented in the models are on opposite sides of the body (e.g. right erector spinae in the up-change model versus left erector spinae in the down-change model). This may represent differences in trunk muscle coactivity expected when lifting different magnitudes of weight (Marras and Sommerich 1991b, Granata and Marras 1993, 1995, Marras and Mirka 1993, Mirka and Marras 1993). Based on these studies, it would be expected that the changes in trunk postures and velocities that accompany lower weights (Davis and Marras 1998) may result in different levels of muscle coactivity. This suggests that antagonistic muscle activity may be an important contributor to the decision process, and thus indicates that multiple muscles may contribute to the perception of exertion.

4.1. *Other considerations*

The methodology for determining the MAWL in this experiment was slightly different than that used in previous psychophysical experiments. Whereas other studies have set a time limit for the weight adjustment period (ranging from 20 min to 8 h), this experiment defined the MAWL as the weight lifted for eight consecutive no-changes. The protocol used here was based on a pilot study that indicated that most changes occurred in the first few lifts of the adjustment period, followed by minor oscillatory changes. Additionally, this protocol was also used to minimize the effects that localized muscle fatigue could have on an EMG signal. Thus, it is possible that the MAWLs in this experiment could have been different from those in previous studies, which allow more time for adjustment. However, the MAWLs determined in this study were consistent with those from other studies for a similar task (Ciriello *et al*. 1990).

Second, the variables in the multiple logistic regression models are only applicable to a comparable task. These results may only represent a complex task that occurs 4.3 times per minute. Individuals may respond to different perceived exertion mechanisms under a different set of circumstances; that is, they may respond to a different set of muscles or rely upon the whole body exertion sensation mechanism. Under a purely sagittal lift, the internal oblique and latissimus dorsi muscles may not be associated with the perceived exertion. Under slower lift rates, heart rate may not play a role in the decision process during the decision to change down. However, a faster lifting rate may cause heart rate to play a larger role in the perceived exertion and the decision process since a higher physiological demand would be expected indicating a whole body exertion mechanism of perceived exertion.

The present perceived exertion models reflect the decision process for males and may be less applicable to a female population. Females, who on average have less strength than males, may rely more upon larger muscles (e.g. erector spinae and latissimus dorsi), which may influence their perceived exertion. Additionally, the subjects in this experiment were young college students inexperienced in materials handling. This may have influenced the magnitude of the final MAWLs. Thus, the variables in the models based on this participant population may not be applicable to other populations that may use a different decision process to make changes in the weight of the loads (e.g. experienced materials handlers or older populations).

5. Conclusions

During the decision to decrease the weight of the load prior to the next lift, the participants appear to be perceiving both local (i.e. direct muscle force) and whole body exertion (i.e. heart rate) factors to distinguish perceived exertions. The individuals based their perception of the exertions on multiple muscles (erector spinae, internal oblique and latissimus dorsi) indicating that different parts of the task contributed to the overall perception of exertion. For the increases of weight prior to the next lift, the participants relied solely upon local factors as heart rate was not in the final model. While the same muscle groups were in the up-change model and the down-change model, the muscles were on opposite sides of the body, indicating differences due to changes in trunk kinematics and trunk muscle coactivity. Based on the decision process models, participants seemed to be responding to muscle forces when determining the MAWL, which indicates that the psychophysical methodology may be more sensitive to muscular strain than spinal loading.

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References

Armstrong, R. B. 1984, Mechanisms of exercise-induced delayed onset muscular soreness: a brief review, *Medicine and Science in Sports and Exercise*, **16,** 529 – 538.

Armstrong, R. B. 1986, Muscle damage and endurance events, *Sports Medicine*, **3,** 370 – 381.

Armstrong, R. B. 1990, Initial events in exercise-induced muscular injury, *Medicine and Science in Sports and Exercise*, **22,** 729 – 735.

Armstrong, R. B., Warren G. L. and Warren, J. A. 1991, Mechanisms of exercise-induced muscle fibre injury, *Sports Medicine*, **12,** 184 – 207.

Borg, G. A. V. and Noble, B. J. 1974, Perceived exertion, *Exercise & Sports Science Review*, **2,** 131 – 153.

Cafarelli, E. 1982, Peripheral contributions to the perception of effort, *Medicine and Science in Sports and Exercise*, **14,** 382 – 389.

Cafarelli, E. 1988, Force sensation in fresh and fatigued human skeletal muscle, *Exercise & Sports Science Review*, **16,** 139 – 168.

Cafarelli, E. and Layton-Wood, J. 1986, Effect of vibration force sensation in fatigued muscle, *Medicine and Science in Sports and Exercise*, **18,** 516 – 521.

Caldwell, L. S., Chaffin, D. B., Dukes-Dobos F. N., Kroemer, K. H. E., Laubach, L. L., Snook S. H. and Wasserman, D. E. 1974, A proposed standard procedure for static muscle strength testing, *American Industrial Hygiene Association Journal*, **35,** 201 – 206.

Chaffin, D. B. and Page, G. B. 1994, Postural effect on biomechanical and psychophysical weight-lifting limits, *Ergonomics*, **37,** 663 – 676.

Ciriello, V. M., Snook, S. H. and Hughes, G. J. 1993, Further studies of psychophysically determined maximum acceptable weights and forces, *Human Factors*, **35,** 175 – 186.

Ciriello V. M., Snook, S. H., Blick, A. C. and Wilkinson, P. L. 1990, The effects of task duration on psychophysically-determined maximum acceptable weights and forces, *Ergonomics*, **33,** 187 – 200.

Davis, K. G. and Marras, W. S. 1998, Is changing box weight an effective ergonomic control? *Proceedings of the Human Factors and Ergonomics Society 42nd Annual Meeting* (Chicago: The Human Factors and Ergonomics Society).

Ekblom, B. and Goldbarg, A. N. 1971, The influence of training and other factors on the subjective rate of perceived exertion, *Acta Physiologica Scandinavica*, **83,** 399 – 406.

Fathallah, F. A., Marras, W. S., Parnianpour, M. and Granata, K. P. 1997, A method for measuring external spinal loads during unconstrained free-dynamic lifting, *Journal of Biomechanics*, **30,** 975 – 978.

Gamberale, F. 1990, Perception of effort in manual material handling, *Scandinavian Journal of Work Environment and Health*, **16** (Suppl. 1), 59 – 66.

Gandevia, S. C. 1982, The perception of motor commands or effort during muscular paralysis, *Brain*, **105,** 151 – 159.

Gandevia, S. C. and McCloskey, D. I. 1977, Effects of related sensory inputs on motor performances in man studied through changes in perceived heaviness, *Journal of Physiology*, **272,** 653 – 672.

Gandevia, S. C. and McCloskey, D. I. 1978, Interpretation of perceived motor commands by reference to afferent signals, *Journal of Physiology (London)*, **283,** 493 – 499.

Granata, K. P. and Marras, W. S. 1993, An EMG-assisted model of loads on the lumbar spine during asymmetric trunk extensions, *Journal of Biomechanics*, **26,** 1429 – 1438.

Granata, K. P. and Marras, W. S. 1995, An EMG-assisted model of trunk loading during free-dynamic lifting, *Journal of Biomechanics*, **28,** 1309 – 1317.

Guo, H. R., Tanaka, S., Cameron, L. L., Seligman, P. J., Behrens, V. J., Ger, J., Wild, D. K. and Putz-Anderson, V. 1995, Back pain among workers in the United States: national estimates and workers at high risk, *American Journal of Industrial Medicine*, **28,** 591 – 602.

Herrin, G. A., Jaraiedi, M. and Anderson, C. K. 1986, Prediction of overexertion injuries using biomechanical and psychophysical models, *American Industrial Hygiene Association Journal*, **47,** 322 – 330.

Hosmer, D. W. and Lemeshow, S. 1989, *Applied Logistic Regression* (New York: Wiley).

Jager, M. and Luttmann, A. 1991, Compressive strength of lumbar spine elements related to age, gender, and other influences, *Journal of Electromyography and Kinesiology*, **1,** 291 – 294.

Jones, L. A. and Hunter, I. W. 1983, Effect of fatigue on force sensation, *Experimental Neurology*, **81,** 640 – 650.

Jones, L. A. and Hunter, I. W. 1985, Effect of muscle tendon vibration on the perception of force, *Experimental Neurology*, **87,** 35 – 45.

Jorgensen, M. J., Davis, K. G., Kirking, B. C., Lewis, K. E. K. and Marras, W. S. 1999, The significance of biomechanical and physiological variables during the determination of the maximum acceptable weight of lift, *Ergonomics* (in press).

Karwowski, W. and Ayoub, M. M. 1984, Fuzzy modeling of stresses in manual lifting tasks, *Ergonomics*, **27,** 641 – 649.

Killian, K. L., Cambell, E. J. M. and Howell, J. B. L. 1979, The effect of increased ventilation on resistive load discrimination, *American Review of Respiratory Diseases*, **120,** 1233 – 1238.

Legg, S. J. and Myles, W. S. 1981, Maximum acceptable repetitive lifting workloads for an 8-hour work-day using psychophysical and subjective rating methods, *Ergonomics*, **24,** 907 – 916.

Marras, W. S. 1990, Industrial electromyography, *International Journal of Industrial Ergonomics*, **6,** 89 – 93.

Marras, W. S. and Granata, K. P. 1995, A biomechanical assessment and model of axial twisting in the thoraco-lumbar spine, *Spine*, **20,** 1440 – 1451.

Marras, W. S. and Granata, K. P. 1997, Spine loading during trunk lateral bending motions, *Journal of Biomechanics*, **30,** 697 – 703.

Marras, W. S. and Mirka, G. A. 1993, Electromyographic studies of the lumbar trunk musculature during the generation of low-level trunk acceleration, *Journal of Orthopaedic Research*, **11,** 811 – 817.

Marras, W. S. and Sommerich, C. M. 1991a, A three-dimensional motion model of loads on the lumbar spine. I. Model structure, *Human Factors*, **33,** 123 – 137.

Marras, W. S. and Sommerich, C. M. 1991b, A three-dimensional motion model of loads on the lumbar spine. II. Model validation, *Human Factors*, **33,** 139 – 149.

Marras, W. S., Fathallah, F. A., Miller, R. J., Davis, S. W. and Mirka, G. A. 1992, Accuracy of a three-dimensional lumbar motion monitor for recording dynamic trunk motion characteristics, *International Journal of Industrial Ergonomics*, **9,** 75 – 87.

Marras, W. S., Lavender, S. A., Leurgans, S. E., Rajulu, S. L., Allread, W. G., Fathallah, F. A. and Ferguson, S. A. 1993, The role of dynamic three-dimensional trunk motion in occupationally-related low back disorders, *Spine*, **18,** 617 – 628.

Matthews, P. B. C. 1982, Where does Sherrington's 'muscular sense' originate? Muscles, joints, corollary discharges? *Annual Review of Neuroscience*, **5,** 189 – 218.

McCloskey, D. I. 1978, Kinesthetic sensibility, *Physiology Review*, **58,** 763 – 792.

McGill, S. M. 1997, The biomechanics of low back injury: implications on current practice in industry and the clinic, *Journal of Biomechanics*, **30,** 465 – 475.

McGill, S. M., Santaguida, L. and Stevens, J. 1993, Measurement of the trunk musculature from T_5 to L_5 using MRI scans of 15 young males corrected for muscle fiber orientation, *Clinical Biomechanics*, **8,** 171 – 178.

Mihevic, P. M. 1981, Sensory cues for perceived exertion: a review, *Medicine and Science in Sports and Exercise*, **13,** 150 – 163.

Mirka, G. A. and Marras, W. S. 1993, A stochastic model of trunk muscle coactivation during trunk bending, *Spine*, **18,** 1396 – 1409.

National Institute for Occupational Safety and Health (NIOSH) 1981, Work practices guide for manual lifting, NIOSH Technical Report DHHS (NIOSH) Publication no. 81 – 122.

Nicholson, L. M. and Legg, S. J. 1986, *Ergonomics*, **29,** 903 – 911.

Pandolf, K. B. 1978, Influence of local and central factors in dominating rated perceived exertion during physical work, *Perceptual & Motor Skills*, **46,** 683 – 698.

Pandolf, K. B. 1983, Advances in the study and application of perceived exertion, in R. Terjung (ed.), *Exercise and Sport Sciences Reviews* (Philadelphia: Franklin Institute Press), 118 – 158.

Robertson, R. J. 1982, Central signals of perceived exertion during dynamic exercise, *Medicine and Science in Sports and Exercise*, **14,** 390 – 396.

Roland, P. E. 1975, Do muscular receptors in man evoke sensations of tension and kinaesthesia? *Brain Research*, **99,** 162 – 165.

SAS Institute Inc. 1989, *SAS User's Guide* (Cary, NC: SAS Institute).

Snook, S. H. 1978, The design of manual handling tasks, *Ergonomics*, **21,** 963 – 985.

Snook, S. H. and Ciriello, V. M. 1991, The design of manual handling tasks: revised tables of maximum acceptable weights and forces, *Ergonomics*, **34,** 1197 – 1213.

Spengler, D. M., Bigos, S. J., Martin, N. A., Zeh, J., Fisher, L. and Nachemson, A. 1986, Back injuries in industry: a retrospective study. I. Overview and cost analysis, *Spine*, **11,** 241 – 245.

Thompson, D. D. and Chaffin, D. B. 1993, Can biomechanically determined stress be perceived? *Proceedings of the Human Factors and Ergonomics Society 37th Annual Meeting* (Santa Monica, CA: Human Factors Society), 789 – 792.

Appendix

We want you to imagine that you are working on a job where you are getting paid for bulk. The job would be conducted over an 8-h shift that allows you to go home not feeling exhausted. We want you to work as hard as you can *without straining yourself, or without becoming unusually tired, overheated, or out of breath*.

The task will consist of one lifting frequency of 4 lifts/min. You will be lifting a box at knee height to a position marked at about elbow height. The load will be returned to the original position by one of the experimenters.

YOU WILL ADJUST YOUR OWN WORKLOAD AS YOU FEEL APPROPRIATE. You will lift when the computer-generated tone signals the start of the lift. Your job will be to adjust the load according to how you feel. This part of the task will not be easy. Remember, only you know how you feel. You will be able to adjust the weight by adding or removing masses from the box.

If you feel you are working too hard, reduce the load. But we don't want you loafing either. If you feel you can work harder, as you might on piece work, increase the load. Don't hurry your lift. Feel free to adjust the load as many times as necessary. Remember, we are not interested in how much you are capable of lifting but rather the maximum amount that you would like to handle if you were actually performing the task at work.