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Within-Subject Variability in Low Back Load in a Repetitively Performed, Mildly Constrained Lifting Task

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Study Design. A repeated-measures *in vivo* experiment.

Objective. To describe within-subject variability of spinal compression in repetitive lifting.

Summary of Background Data. Epidemiology and failure mechanics suggest that peak loads may be more predictive of injury than average loads. Nevertheless, biomechanical studies usually focus on the latter.

Methods. Ten healthy males performed 360 lifts in 1 hour of a 45-L crate, weighted with a stable 10-kg mass on 1 day and with an unstable mass (10 kg of water) on another day. The maximum compression force in each lift was estimated, using a simple inverse dynamics model and a single equivalent muscle model.

Results. The individual distributions of maximum compression force were slightly skewed to the right (average skewness 0.67). Median and 95th percentile values were used to characterize the distribution. The median (50th percentile) compression ranged from 3375 to 6125 N, and from 3632 to 6298 N in the stable and unstable load conditions, respectively. The within-subjects peak (95th percentile) compression forces were from 405 to 1767 N and from 526 to 2216 N, respectively, higher than the median values. The peak values differed significantly between conditions, whereas the difference in medians did not reach significance. Only a limited trendwise (fatigue-related) variance could be demonstrated.

Conclusion. Peak spinal compression by far exceeds median compression in repetitive lifting and can be affected by task conditions independently from the median. Therefore, the variability of spinal loads needs to be taken into consideration when analyzing and redesigning tasks that can cause spinal injuries. [Key words: compression, lifting, variability, injury prevention] **Spine 2001;26:1799–1804**

It has been well established that repeated performance of the same motor task by an individual entails substantial, apparently random, variation in the kinematics of the successive trials.^{2,5} It is therefore to be expected that the mechanical loads on the musculoskeletal system also vary considerably between trials. Surprisingly, although variance in task performance has received ample attention in the motor control literature, it is often overlooked in studies on mechanical loading induced by the performance of motor tasks. The latter studies typically tend to focus on average values of parameters of mechanical

load to obtain a best estimate. Although the maximum within a trial is usually selected for further analysis, maximum values are subsequently averaged within and between subjects (Examples of low back load in lifting were presented previously.^{4,15,18}) However, both biomechanical¹⁰ and epidemiologic^{14,19,26} models predict that the peak of the time-series of force acting on the spine strongly determines the probability of injury. In other words, given a specific injury threshold, the peak load determines whether and when injury will occur (Figure 1).

It is conceivable that task conditions affect the peak load independently from the average load. For instance, when lifting objects with variable mechanical properties in contrast to objects with constant properties, the variance in back load may increase around a constant average. In addition, the study participant's status may affect peak load differently from the average. For example, because fatigue has been shown to increase kinematic and kinetic variance,²⁷ the effect of fatigue on low back load may be more apparent in an increase of the peak load over a number of lifts rather than in the average. This might explain why several studies have found only moderate effects of fatigue on low back load in lifting.^{11,12,32}

To date only a few studies have addressed the issue of variance in low back loading.^{13,23,24} In these studies only a limited number of trials per study participant was available, limiting the possibilities of drawing inferences with respect to the distribution of peak loads across repeated trials. In addition, these studies reported the total variance in the data set without separating within- and between-subject components. Because anthropometric characteristics and injury thresholds also vary between subjects, within-subject variance is more directly relevant and separation of the two components is needed. For this reason the present study focused on within-subject variability of back load during lifting. The aims were to describe the distribution of maximum compression forces on the spine in repetitive lifting and to explore underlying causes of variance in compression. Special attention was given to variable mechanical properties of the object handled and fatigue as potential causes of variance.

Methods

Ten male study participants with an average ($\pm$ SD) age of 23.2 $\pm$ 2 years, mean height of 1.82 $\pm$ 0.07 m, and mean body mass of

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Figure 1. Schematic model of injury causation. A constant individually determined threshold at which the spine is injured is assumed. Spinal load in a number of repeated trials of a motor task varies in an apparently random fashion. Although on average spinal load remains below the threshold, a relatively high load incurred during one of the repeats does exceed the threshold and causes injury.

72 ± 10 kg participated in the experiment. All of the study participants declared to have been free from low back trouble and signed informed consent forms before participation. Light-emitting diodes (LEDs) were taped over the T10 and T12 spinous process, on the left and right iliac crests, and the superior posterior iliac spines. The positions of these LEDs were recorded at 100 Hz (Optotrak, Northern Digital Inc. Waterloo, Ontario, Canada).

Surface EMG recordings were made using bipolar disposable Ag-AgCl electrodes (Medi-Trace [Graphic Controls Corp., Buffalo, NY] pellet electrodes ECE 1801, lead-off area 1.0 cm^2 , interelectrode distance 2.5 cm). Signals were amplified 20 times (Porti-17, TMS, Enschede, The Netherlands; input impedance $> 10^{12} \Omega$, CMRR > 90 dB), band-pass filtered (10–400 Hz), and A-D converted (22 bits) at 1000 Hz. Activity was recorded bilaterally from the erector spinae muscle (ES: 3 cm lateral of the L3 spinous process) and external oblique muscle (EO: umbilical level above the anterior superior iliac spine). Ground reaction forces were recorded by means of a custom-made strain gauge force platform ($1.0 \text{ m} \times 1.0 \text{ m}$). The analogue force signals were amplified, filtered (10 Hz, fourth-order Butterworth filter), sampled (100 Hz), and stored.

On two separate occasions, at least 2 days apart, the study participant performed a repetitive lifting task with either a stable or an unstable object. The object handled was a 45-L crate ($0.5 \times 0.5 \times 0.3 \text{ m}$) with handles at the top. Stable load conditions were created by taping a 10-kg weight to the bottom of the empty crate, and unstable conditions were created by filling the crate with 10 kg of water. The crate had to be transported between a shelf at 0.7 m height and 0.5 m in front of the study participant's toes and a shelf just in front of the study participant at 0.1 m above the level of the forceplate. A metronome was used to pace the study participants. Lowering and placing the box on the low shelf and lifting and placing the box on the high shelf took 5 seconds, after which 5 seconds of rest in the upright position was allowed. The study participant was instructed to use a so-called back technique (*i.e.*, to keep the knees extended as much as possible during lifting). After re-

peated lifting for 10 minutes (60 lifts) the study participant stepped from the forceplate (to allow resetting) and performed an isometric test contraction (holding a 5-kg weight for 5 seconds with the arms extended horizontally) to obtain stationary EMG signals for fatigue monitoring. Lifting was then resumed after approximately 30 seconds. In total 6 bouts of 10 minutes each were performed.

Before the actual experiment study participants performed three attempted maximum isometric contractions of the trunk extensor (manually resisted trunk extension from a prone position) and flexor muscles (manually resisted trunk flexion from a supine position). After 15 minutes of rest an isometric test contraction as described above was performed and subsequently the first lifting bout was started.

LED positions on the pelvis were used to reconstruct a reference frame with its origin at L5–S1, the x-axis in the frontal and transversal planes, and the y-axis in the sagittal and transversal planes. The ground reaction force was projected on this reference frame and the three-dimensional moment around L5–S1 was directly estimated on this basis. The trunk angle was defined as the angle of the line through the T10 and T12 LEDs with the z-axis of the pelvis reference system. Thus, a larger lumbar angle coincides with an increase in flexion.

EMG data obtained during maximum isometric contraction trials and lifting were filtered with a 30-Hz high-pass finite impulse response filter to reduce ECG contamination,³⁰ full-wave rectified and normalized to the maximum found for each muscle in the maximum isometric tests. The level of abdominal co-activity was defined as the sum of left and right normalized EO EMG divided by the sum of left and right ES EMG. To minimize the effect of the electromechanical delay EMG signals were shifted in time by 120 msec with respect to kinematic and kinetic data.⁹ The mean power frequency of the EMG signals obtained from the ES during the isometric test contractions was determined by means of fast Fourier transformation (1024-point Hanning windows, with 800-points overlap).⁸

Maximum net moments in each lift and the concomitant lumbar angle were determined. From these compression forces estimates were obtained through a model as described by Dieën and Looze.⁶ In short, origins and insertions of 114 muscles crossing the L5–S1 joint were averaged after weighting for cross-sectional area and in case of the abdominal muscles for the level of co-activity. This procedure yields a single equivalent muscle model for the trunk muscles at the given lumbar angle. Given the maximum net moment and an estimated inclination angle of the L5–S1 disc with respect to pelvis, the maximum compression force in each lift could be estimated.

To describe the within-subject distribution of peak compression forces, the skewness, the median (50th percentile), and 95th percentile of the 360 values obtained for each study participant in each condition were calculated. The 95th percentile of spinal compression was used as an indicator of the peak load rather than the absolute maximum value because the latter is more likely to be a consequence of a random error in the data. Linear regression of the mean power frequency values of the EMG against time was performed to assess the occurrence of back muscle fatigue. Subsequently, linear regression analysis of compression against the number of the lift was used to quantify the extent back load varied with time and hence possibly with fatigue. The effects of the stability of the object on median and 95th percentile compression values were tested separately by means of Wilcoxon's test for matched samples. To obtain insight in the causes of the variance in compression force, esti-

Figure 2. An example of a time series (upper panel) and concomitant histogram (lower panel) of the peak compression forces in one study participant (patient 3) while lifting the stable load.

mations of compression force were recalculated three times, each time keeping two of three input parameters to the model (net moment, trunk angle, level of co-activity) constant at their respective average values.

Results

The individual distributions of compression force were in general slightly skewed to the right (Figure 2, Table 1). The within-subject variance in peak compression values was considerable (Figure 2, Table 1). The within-subject median (50th percentile) compression ranged from 3375 to 6298 N across load conditions. The within-subject peak (95th percentile) compression forces were from 405 to 2216 N (or 9–39%) higher than the median values.

The mean power frequency of ES EMG significantly decreased in both conditions in all study participants but one, indicating that the extensor musculature became fatigued by the experimental task. Overall, the decrease in mean power frequency averaged -2.0 Hz per lifting bout (SD 1.2 Hz per bout). The decrease was not significantly affected by the load condition.

Fatigue-related changes did not account for a substantial part of the variance in compression because on average the variance explained by the regression of lift number *versus* peak compression was only 10% (range 7–39%). Nevertheless, a statistically significant increase in compression over the time-series could be detected in most study participants (Table 2). When performing regression analysis of the 10-minute bout number against the median or 95th percentile compression force in each bout, significant increases in compression could be detected in some study participants (Table 2). This was the case in more study participants for the median level than for the 95th percentile level. However, Wilcoxon's matched-pairs test revealed no significant differences between the slopes of the median and 95th percentile values, suggesting that the whole distribution of compression values shifted toward higher values. Also, the effect of load condition on these slopes was not significant.

The difference between the 95th percentile and the median compression was larger (median 987 *vs.* 699 N) in the unstable load condition than in the stable condition, which was found to be significant ($P = 0.007$). Although the median compression was higher in the unstable condition (median 4788 *vs.* 4644 N), this difference was not significant ($P = 0.053$). In contrast, the difference between the 95th percentile values (median 5587 *vs.* 5182 N) was significant ($P = 0.042$).

Although all of the input parameters of the model used to estimate compression forces showed considerable within-subject variance (Figure 3), variance of the

Table 1. Skewness, Median (50th Percentile) Values, and 95th Percentile Values of the Within-Subject Distributions of Peak Compression Forces in Repeated Lifting of Stable (Fixed 10-kg Mass) and Unstable (10-kg Water) Loads

Subject	Skewness		Median Compression		95th Percentile Compression		95th Percentile – Median*	
	Stable	Unstable	Stable	Unstable	Stable	Unstable	Stable	Unstable
1	0.98	1.64	4900	3895	6667	5416	1767	1521
2	1.69	1.46	6125	6298	7816	8514	1691	2216
3	1.15	0.77	4152	4788	4853	5587	700	799
4	-0.76	1.08	3665	3654	4264	4729	599	1075
5	0.60	-0.14	4848	5218	5545	6319	697	1101
6	0.54	1.56	3375	3632	3781	4370	405	738
7	0.05	0.58	4759	4918	5484	5743	726	824
8	-0.28	0.10	4507	4758	4921	5284	414	526
9	0.55	0.51	4644	5278	5182	6178	538	899
10	0.49	0.85	4286	4845	4987	5939	701	1094
Mean	0.50	0.84	4526	4729	5350	5808	824	1079
SD	0.71	0.60	755	824	1160	1128	491	481

* The difference is an indicator of variability.

Table 2. Individual Coefficients of Correlation Between the No. of the Lift and Peak Compression Force and Between the No. of the Lifting Bout and the Median and 95th Percentile Compression per Bout

Subject	Lift vs. Compression (n = 360)		Bout vs. Median Compression (n = 6)		Bout vs. 95th Percentile Compression (n = 6)	
	Stable	Unstable	Stable	Unstable	Stable	Unstable
1	0.10	-0.27*	0.46	-0.66	0.09	-0.55
2	0.03	0.13*	-0.26	0.69	0.60	0.50
3	0.35*	-0.04	0.83*	-0.08	0.76	0.38
4	0.18*	0.45*	0.82*	0.72	0.11	0.77
5	0.19*	0.39*	0.77	0.89*	0.53	0.93*
6	0.25*	-0.09	0.97*	-0.29	0.55	-0.39
7	0.63*	0.47*	0.93*	0.96*	0.92*	0.89*
8	0.49*	0.41*	0.95*	0.74	0.92*	0.58
9	0.03	0.12*	0.22	0.84*	-0.73	-0.05
10	-0.25*	-0.39*	-0.74	-0.84*	-0.05	-0.56
Mean	0.20	0.12	0.50	0.30	0.37	0.25
SD	0.25	0.31	0.59	0.69	0.52	0.59

* Significant value at $P < 0.05$.

net moment and the level of co-activity were the main contributors to variance in the peak compression estimates. Varying the net moment only reduced the difference between the 95th percentile and the median compression estimate from a median value of 768 (405–2216) N to 562 (308–1301) N. Varying the level of co-activity only reduced the difference to 450 (92–1734) N and varying the lumbar angle only reduced it to 94 (30–636) N. Only the lumbar angle showed a clear trendwise variation, accounting for on average 33% (range 0.2–82%) of the total variance. In general, the lumbar angle was found to increase (more flexion) over time (Table 3). In line with this finding, varying this input parameter while keeping the other two constant produced a clear increase in estimated compression in most study participants (Table 3).

Figure 3. An example of the time series of the maximum net moment (upper panel), lumbar angle (middle panel), and level of co-activity at the instant of the maximum net moment (lower panel). The data correspond with those presented in Figure 2.

Discussion

The present study was designed to describe within-subject variance of spinal loading in repeated performance of a lifting task. Even though the task was fairly constrained in terms of lifting technique and speed, considerable variation was present, with 95th percentile values of compression being on average 952 N or 20% higher than the median values. In a study by Granata et al,¹³ the overall average compression force was at 5790 (SD 1480) N, well comparable with our results. In their data 14% of the total variance could be ascribed to within-subject variance. On this basis and assuming a normal distribution, the 95th percentile value of the within-subject distribution can be estimated to be 1.19 times the mean. Given slightly skewed distributions, as found in the present study, this fits excellently with the present results.

The method used in the present study to estimate spinal compression consisted of two simplified biomechanical models. The simplifications were needed to allow data acquisition and analysis of a large number of trials in each study participant. Although the fact that compression force estimates compare well with the literature^{5,13,22} illustrates that no major systematic errors were introduced, it is necessary to consider the effects this may have had on the estimates of variability. The first model used standard inverse dynamics to estimate the net moment about the L5–S1 joint. It was simplified in that it treated the legs and extremities essentially as one stationary segment, thus ignoring inertial forces acting on the constituting segments. This allowed a major reduction in the number of LEDs to be tracked and has been shown not to cause large errors in net moments in back lifting.³³ The second model reduced the trunk musculature producing this moment to a single equivalent muscle. However, because we have previously shown⁶ that the validity of this approach strongly depends on assumptions regarding co-activity of abdominal muscles,

Table 3. Individual Coefficients of Correlation Between the No. of the Lift and Peak Moment, and Flexion Angle and Level of Coactivity at the Instant of Peak Moment and Individual Coefficients of Correlation Between the No. of the Lift and Peak Compression Force Obtained While Varying Only One of Three Input Parameters (Net Moment, Lumbar Angle, Coactivity)

Subject	Lift vs. Net Moment		Lift vs. Lumbar Angle		Lift vs. Coactivity		Lift vs. Compression Estimate Varying Net Moment		Lift vs. Compression Estimate Varying Lumbar Angle		Lift vs. Compression Estimate Varying Coactivity	
	Stable	Unstable	Stable	Unstable	Stable	Unstable	Stable	Unstable	Stable	Unstable	Stable	Unstable
1	0.00	-0.25*	-0.55*	0.05	-0.07	0.19*	0.00	-0.25*	0.37*	0.05	0.02	-0.24*
2	0.45*	0.28*	0.70*	0.75*	0.32*	0.06	0.45*	0.28*	0.70*	0.75*	-0.26*	-0.06
3	0.10	-0.36*	0.46*	-0.17*	-0.26*	-0.38*	0.10	-0.36*	0.46*	-0.19*	0.33*	0.29*
4	-0.25*	0.40*	0.62*	0.74*	-0.10	0.07	-0.25	0.40*	0.65*	0.72*	0.13*	-0.04
5	0.04	0.42*	0.27*	-0.15*	-0.26*	-0.46*	0.04	0.42*	0.27*	-0.15*	0.18*	0.40*
6	0.16*	-0.21*	0.54*	0.77*	-0.31*	0.08	0.16*	-0.21*	0.54*	0.77*	0.27*	-0.06
7	0.56*	0.37*	0.91*	0.89*	0.08	-0.14*	0.56*	0.37*	0.91*	0.89*	-0.15*	0.09
8	0.38*	0.40*	0.55*	0.75*	-0.11*	-0.05	0.38*	0.40*	0.80*	0.75*	0.09	-0.01
9	-0.28*	0.20*	0.37*	-0.58*	-0.38*	-0.17*	-0.28*	0.20*	0.37*	-0.58*	0.34*	0.06
10	-0.33*	-0.39*	-0.36*	0.16*	-0.06	0.22*	-0.33*	-0.39*	-0.40*	0.17*	0.09	-0.23*
Mean	0.08	0.09	0.35	0.32	-0.11	-0.06	0.08	0.09	0.47	0.32	0.10	0.02
SD	0.09	0.12	0.44	0.35	-0.12	-0.08	0.09	0.12	0.48	0.35	0.11	0.05

* Significant value at $P < 0.05$.

we included a measured level of co-activity in the model. As the level of co-activity was based on the most active muscle at the initiation of a lifting movement,¹⁷ this may represent a systematic overestimate. In addition, the model accounted for the lumbar angle, which through its effect on the muscle lever arms and their orientation with respect to L5–S1 also substantially influences compression estimates.⁶ Other important influences on compression estimates, such as lumbar curvature and anthropometry,⁶ are rather constant subject properties. Because these factors were not expected to have a strong effect on within-subject variance of compression forces, no attempt was made to account for them. Errors introduced by the estimation of the L5–S1 position (through palpation of landmarks) and orientation (fixed angle with respect to the pelvis) do not affect within-subject variance in one session, which was the main topic of this article. The first may increase variance between sessions and thus obscure to some extent differences between the two experimental conditions. The second source of error is expected to reduce between-subject variance to an unknown extent. Finally, the approach chosen assumes negligible moments in the frontal and transversal planes. This assumption agreed with the net moment estimates obtained (data not presented).

The results of the present study suggest that variance in movement execution and thus ultimately in the net moments produced contributes substantially to variance in spinal loading. In line with this, Mirka and Marras²⁴ found the peak compression force to be only 6% above the mean, in tasks where kinematics and net moments were strictly controlled. This small variance suggests that, in these highly constrained tasks, co-activity induced only limited variance in spinal loading. In contrast, in the present study the varying level of co-activity

was also shown to be able to substantially contribute to variance in spinal loading. This disparity may be explained by co-activity being more constant in more constrained tasks.

Although the small difference in P values of the effects of load condition on peak and median compression does not warrant dismissing one while accepting the other, the effects would conventionally be interpreted differently because only the former was significant. In part this finding was anticipated because the sloshing content of the box in the unstable load condition can both increase and decrease the moment arm of gravity acting on the box with respect to the spine. Nevertheless, the effect on the median load, which almost reached significance, might be expected as a consequence of increased co-activity to deal with this unstable situation.

A limited part of the variance in compression could be explained by the number of the lifting trials and is thus probably related to fatigue. The occurrence of extensor muscle fatigue was confirmed by a decrease of the frequency content of the EMG signals of these muscles, as in previous studies on repetitive lifting.^{12,28,29} The trendwise increase of compression is caused by the increase in the lumbar angle, which affects the extensor muscles' moment arms. This increase in flexion from muscle fatigue and possibly flexion creep²¹ is in line with previous studies on repetitive lifting.^{3,12,32}

In conclusion, trial-to-trial variance of spinal compression in a repetitive lifting task is considerable, and the variance is affected by some task conditions more clearly than the median. It is to be expected that in real-life tasks the variance in loading is even larger than in the relatively constrained experimental task studied here. The implications of this finding are therefore considerable. Conventionally in injury prevention, the focus is on

average load levels. Although cumulative loading also constitutes a factor in the causation of injury risk,^{10,16,26,31} the peak level probably is the most important determinant of injury risk as a result of compression.^{10,14,19,26} The focus on average loads may therefore be misleading. Not only does the peak load exceed the median substantially, it was also found to be to some extent independent from it. However, the present results should not be interpreted as a plea for task design, which constrains task performance, with the aim of reducing peak loads. Several considerations support the notion that variability in task performance is beneficial, both with respect to health aspects^{7,20} and optimization of task performance.^{1,2} This study showed that such variability implies that injury prevention considerations in task design need to encompass the variability of the mechanical load in repeated performance of a task.

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■ Key Points

- Ten study participants repetitively lifted a crate 360 times in approximately 1 hour, once with a stable content, once containing water.
- The maximum compression force in each lift was estimated.
- The within-subject 95th percentile of the compression estimates substantially exceeded the median value.
- When lifting the crate filled with water the 95th percentile compression was higher than when lifting the crate with stable content, whereas median compression did not differ significantly between conditions.

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