Practice-related changes in lumbar loading during rapid voluntary pulls made while standing

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Abstract

Objective. To determine if five days of practice on a novel dynamic, multi-joint pulling task resulted in lower magnitudes of lumbar loading or a more consistent relationship between pulling force and lumbar loading.

Design. A repeated measures design compared how practice influenced the magnitude of lumbar torque and the correlations between lumbar torque and pulling force.

Background. Previous studies suggest that practice can decrease the magnitude of lumbar loading on simple manual material handling tasks, but it is unknown whether practice reduces lumbar loading for more complex tasks. Neither is it known whether the consistency of lumbar loading increases with practice.

Methods. Ten healthy adults practiced impulse-like horizontal pulls to targets equaling 20, 40, and 80% of their estimated maximal dynamic pulling force over 5 days. Movements were unrestrained, other than keeping the feet flat on the ground. We used a four-segment, sagittal plane inverse dynamics model to compute lumbar, hip, knee, and ankle torques on Day 1 and Day 5 from ground reaction forces and moments, pulling forces, and kinematics.

Results. An analysis of variance showed significant practice-related changes in lumbar torque at the time of peak pulling force (lumbar $\text{torque}_{\text{peakPF}}$). The lumbar $\text{torque}_{\text{peakPF}}$ decreased for the 20% pulls, did not change for the 40% pulls, and increased for the 80% pulls. Two subjects showed a significant decrease in lumbar $\text{torque}_{\text{peakPF}}$ for all three force levels. Coefficients of determination between pulling force and lumbar torque ($r^2_{PF,LT}$: a measure of the consistency of the relationship between these two variables) were significantly higher on Day 5 than Day 1.

Conclusions. Practice on a novel pulling task changed the magnitude of lumbar torques and increased their correlation with pulling force, suggesting that subjects learned strategies that improve motor control of lumbar torques.

Relevance

The study showed that the magnitude and consistency of lumbar loading changed spontaneously as subjects practiced a novel multijoint pulling task. Such changes may decrease the risk of low back injury.

Keywords: low back, lumbar spine, biomechanics, learning, practice, training

Introduction

Excessive or repeated mechanical stresses on the lumbar spine are thought to be both aggravating and initiating factors in low back disorders. Studies using cadaver motion segments demonstrated that cartilage endplate and trabecular bone of the vertebral body failed under one large loading or repetitive loadings[1,2]. Manual material handling (MMH) tasks like lifting, pulling and pushing are often associated with low back injury, because they impose significant repetitive loads on the lumbar spine[3-5]. Recently, it has been proposed that motor control errors in counteracting lumbar loading contribute to low back injury during MMH tasks[6].

Improved motor control might reduce the risk of low back injury during MMH tasks in at least two ways. First, people could organize their movements and external load position to decrease the lumbar loading. Second, people could learn to control their movements to more consistently load the lumbar spine, which would facilitate activating the trunk muscles more effectively, without sometimes overloading the lumbar spine[7-9]. Both strategies represent motor control mechanisms that could be learned. To clarify whether people can spontaneously learn to use these mechanisms during complex MMH tasks, we examined how lumbar loading changed with five days of practice on a novel, dynamic pulling task that incurs substantial lumbar loading. Specifically, we investigated whether subjects spontaneously learn strategies to reduce the magnitude or increase the consistency of lumbar loading.

The first motor control mechanism that could protect the lumbar spine during MMH tasks involves learning to reduce lumbar loading. This idea is consistent with the hypothesis that the central nervous system learns to re-organize movements and muscular activities to minimize joint torque or jerk cost[10,11]. In the context of MMH tasks, it has been hypothesized that practice allows subjects to learn how to decrease lumbar loading[12-14]. This hypothesis has some experimental support. For example, experience and instructions to use specific lifting techniques can alter trunk muscle activation patterns and lower lumbar loading during bimanual lifting[15-17]. Moreover, practice on MMH tasks can reduce lumbar loading. Patterson[18] observed that experienced lifters had lower lumbar torque when compared to inexperienced persons. Similarly, Gagnon[15] as well as Lavender and colleagues[19] found that subjects with task experience and practice had significantly lower spinal compression force during repeated imposed trunk loading. While these studies are consistent with the idea that learning reduces lumbar loading, few subjects were evaluated and the dynamics of the loading task were limited. Moreover, the effects of practice on lumbar control have been studied primarily during imposed loads.

The second mechanism would be to learn to control movements so that the lumbar spine is more consistently loaded in a given task, thereby enabling subjects to more reliably and effectively stabilize the spine[7-9]. Unexpected sudden trunk load was reported to be one of the risk factors for low back injury[20,21]. Studies have reported that trunk muscle activation can be delayed, augmented or prolonged when loads are unexpectedly imposed on the trunk[9,22,23]. Such varied muscular responses to external loads represent inconsistent control strategies that could lead to greater trunk accelerations and mechanical loading on lumbar ligaments, facet joints, and intervertebral discs. Although the consistency of lumbar loading has not yet been examined directly, previous studies have shown that conditions which facilitate the anticipation of trunk loads can be beneficial. For example, lumbar torques were lower when subjects knew the magnitude of the external load[7,8,18]. Moreover, Lavender and colleagues[19] found that four subjects who experienced repeated imposed trunk loads over five days showed less spinal compression force after practice. Those reductions could be mediated partly by subjects learning more consistent control strategies for counteracting the external load. Improvements in consistency seem even more likely to characterize the learning of MMH tasks that involve voluntary force production and lumbar loading (e.g., pulling or lifting). Early in practice, subjects might experience variable lumbar loading for a given external load, especially in tasks that involve relatively unrestrained motions and have complex effects on lumbar loading. As skill improves, lumbar loading could become more consistent, which would manifest as higher correlations between the task-related external load and lumbar torque.

This study tested two hypotheses about how lumbar loading might change spontaneously over five days' practice on a novel dynamic pulling task that involves rapid and large movements of the lower extremities and trunk [24,25]. Subjects pulled to three peak pulling force targets, set to low, medium, and high percentages of each subject's estimated maximum pulling force, based on height and weight[25]. Practice effects were assessed by comparing L5-S1 torque on Day 1 and Day 5. The first hypothesis was that lumbar torque would be lower after practice. The second hypothesis was that the correlation between lumbar torque and peak pulling force would be higher after practice, reflecting a more consistent relationship between these variables.

Methods

Subjects

Eight female and two male subjects without any history of low-back pain, neurological, or musculoskeletal disorders (mean ± SD: age = 27.7 ± 9.3 yr; height = 1.65 ± 0.12 m; weight = 56.0 ± 7.0 kg) participated. Subjects signed informed consent forms complying with federal and university guidelines.

Task

Freely standing subjects made brief, bilaterally symmetrical pulls on a rigid handle at elbow height (Figure 1). Subjects were told to produce pulls to three peak force targets, as accurately as possible. The force targets equaled 20, 40, and 80% of each subject's estimated maximum pulling force, based on a regression equation using the person's height and weight. Subjects stood barefoot on a force platform with arms aligned with the trunk and forearms parallel to the ground. They were told to make an impulse-like backward pull and then immediately let the pulling force return to zero. Subjects were told they could move their bodies as they wished to produce the target forces, providing they kept both feet flat on the force platform and tried to keep the forearms parallel to the floor. These constraints did not prevent subjects from making substantial body movements, as illustrated in Figure 1 and described elsewhere[25,26]. Each pull was self-paced and completed within two seconds.

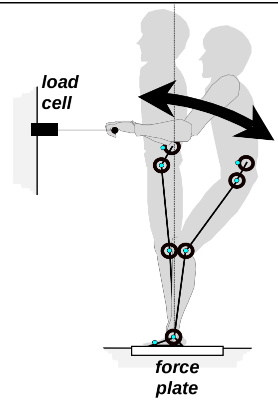

Figure 1. Posture before starting the pull and at a time near the peak pulling force. All subjects had the same start posture but individual subjects moved differently during the pull (see text for references). The double-ended arrow indicates that the posterior motion during pulling and the anterior motion during balance recovery are similar. Large circles indicate joints used in the model; small circles indicate the location of motion analysis markers used with force platform data in the inverse dynamics computation of joint torques and forces.

Protocol

Subjects participated in five sessions over one week. Each session consisted of 12 blocks of 9 pulling trials (3 trials for each of the 20, 40 and 80% targets, presented sequentially). On Day 5, they made pulls to 5, 10, 20, 40, 60, 80 and 95% of their estimated maximal pulling force (total = 135 trials; the additional trials were related to separate questions addressed elsewhere[24,25]). Pulls were brief (time to peak pulling force < 250 ms). Subjects received verbal feedback about the accuracy of their pulls on Days 1 to 4. That feedback was given after every trial on Day 1 (100% feedback) and reduced on subsequent days (given on 66%, 50%, 33% of the trials). No feedback was given on Day 5. No feedback was given about body motion.

Instrumentation

Pulling force, ground reaction forces and moments, and movement kinematics were recorded on Day 1 and 5. A uniaxial load cell (1100 N maximum; Sensotec , Columbus, USA) was used to record the pulling force. The handle was connected to the load cell by an inelastic but flexible steel cable (1.2m). The load cell was bolted, at elbow height, to a rigid metal frame secured to the floor (Figure 1). An AMTI™ (Watertown, USA) force platform was used to record ground reaction forces and moments (100 Hz), from which center of pressure (COP) records were computed. A target region for the COP was centered on midline and at 40% of the distance from the rear to the front of the base of support, defined respectively by the posterior aspect of the lateral malleoli and the interphalangeal joint of the first toes. Feedback about the COP and its target region were shown on a monitor to enable the subject to correctly position the COP before each pull. A two-camera ELITE™ motion analysis system (Milan, Italy) recorded 3D joint coordinates at 50 Hz. Reflective hemispheric markers (1 cm) were taped over the left side of the body at the head of the fifth metatarsal, lateral malleolus, lateral femoral condyle, greater trochanter and iliac crest. Kinematic marker trajectories were conditioned with a linear phase FIR filter[27] and interpolated to 100 Hz before computing kinematics and joint torques.

*Inverse dynamic model*We assumed a sagittal 4-link model for estimating joint reactive torques, similar to models used and validated by others[3,5,28,29]. Segments were rigid and connected by frictionless pin joints. Markers on the iliac crest, greater trochanter, lateral femoral condyle and lateral malleolus were used to approximate the centers of LS/S1, hip, knee, and ankle joint rotation (see Figure 1). A Newtonian formulation was used to develop the equations of motion for determining net intersegmental joint reaction forces and torques. Analysis started with measured forces and moments at the feet and continued to each proximal segment in succession[3,5,15].

Dependent Measures

Our analysis of lumbar loading focused on the pulling phase (Figure 2: time t_1 to t_2), when lumbar torque is greatest. We evaluated lumbar loading during this phase with three measures: (a) lumbar torque at the time of peak pulling force (time t_2); (b) peak lumbar torque; and (c) lumbar torque cost function from t_1 to t_2 [$\frac{1}{2}\int_{t_1}^{t_2} \left(\frac{T_{L5/S1}}{dt}\right)^2 dt$]. Correlations showed that these three measures were highly related (median r = 0.83), and the conclusions of statistical tests using the three measures were the same. Therefore, we only report the results using lumbar torque at the time of peak pulling force (lumbar torque_{peakPF}).

Statistical analysis

The first hypothesis, that lumbar torque_{peakPF} would decrease after practice, was tested with a 2 (day of practice) x 3 (pulling force level) repeated measures ANOVA, using the Huyn-Feldt corrected degrees of freedom which accommodates deviations of data from the normal distribution. Post-hoc 95% confidence interval tests were used to determine how each subject's lumbar torque_{peakPF} changed with practice. Trials were blocked by actual pulling force into three pulling force levels. Trials with peak pulling forces ranging from 10 to 30% of estimated maximal pulling force comprised the 20% pulls block. The same principle was used to block 40 and 80% pulls. Means and standard deviations of the lumbar torque_{peakPF} were computed for each subject and pulling force level, after normalizing to body weight and height.

The second hypothesis, that the correlation between peak pulling force and lumbar torque_{peakPF} would be higher after five days of practice, was evaluated by a paired t-test on the coefficients of determination between actual peak pulling force and lumbar torque_{peakPF} (squared Pearson correlation coefficients, $r^2_{PF,LT}$). Similar correlations were evaluated between peak pulling force and the hip, knee, and ankle torques, all measured at the time of peak pulling force.

Results

General features of pulling force, body motion and lumbar torque

Figure 2 shows one subject's averaged time histories of pulling force, center of mass movement in the anterior-posterior direction (CM_{ap}), and lumbar torque for 20, 40, and 80% pulls on Day 1 and Day 5, with trials matched by actual peak pulling force. The mean duration of the pull was less than 1.0 sec. The CM_{ap} moved posteriorly (up to 0.15 m for 80% pulls), developing momentum before force was exerted on the handle. The direction of CM_{ap} motion was reversed while force was exerted on the handle. Finally, the CM_{ap} returned approximately to its initial position. Lumbar torque exceeded above baseline throughout the movement, always in the direction of external flexion torque, and reached its peak near the time of the peak pull. This subject had lower lumbar torque at the time of peak pulling force on Day 5 than Day 1 for all three force levels. The lower lumbar torque for the 20% pulls was consistent with the group results (described below); the decrease for the 80% pulls was not.

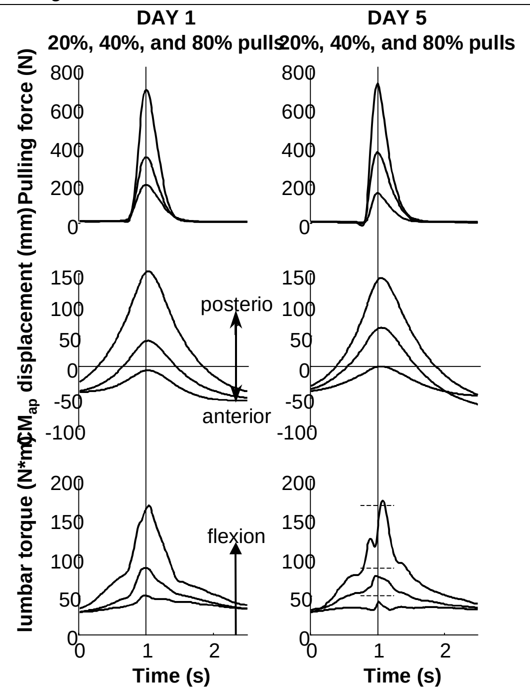

Figure 2. Ensemble-averaged time histories of pulling force, CMap and lumbar torque of one subject on Day 1 and Day 5, with overlaid records from 20%, 40%, and 80% pulls. Vertical lines represent events within the pulling movement. t1 = pulling force onset, t2 = peak pulling force. For averaging, trials were synchronized to the time of the peak pulling force (time t2). Lumbar torque was analyzed for the pulling phase (time t1 to t2). To facilitate comparison across days, the horizontal dashed lines on the Day 5 lumbar torque plot represent the lumbar torque at the time of peak pulling force (lumbar torquepeakPF) on Day 1 and the circles represent the lumbar torquepeakPF on Day 5 at the same pulling force levels.

Practice-related changes in the magnitude of lumbar torque

The results of the ANOVA and post-hoc tests partly supported the hypothesis that the lumbar loading would decrease with practice. Practice had a significant effect on lumbar torque_{peakPF}, but the direction of the effect differed with pulling force (ANOVA interaction term, $F(1.11, 12.17) = 7.08$, $P = .019$). Figure 3 shows group means and 95% confidence intervals of normalized lumbar torque_{peakPF} for 20%, 40%, and 80% pulls, before and after practice. Lumbar torque_{peakPF} decreased only for the 20% pulls (paired t-test, $t(9) = 2.55$, $P = .02$), increased for the 80% pulls (paired t-test, $t(9) = -1.73$, $P = .06$), and did not statistically change for the 40% pulls. Thus, practice significantly influenced but did not always reduce lumbar loading.

The wide confidence intervals in Figure 3 indicate large inter-subject variation in how practice influenced the lumbar torque. Figure 4 shows each subject's lumbar torque_{peakPF} before and after practice, for each pulling force. Even when most subjects showed similar effects at a given pulling force (e.g., decreased lumbar torque_{peakPF} for 20% pulls), a few subjects showed no change or changed in the opposite direction. The 95% confidence interval tests on absolute values of the change in lumbar torque_{peakPF} showed that practice significantly altered lumbar torque in 83% of the thirty between-day comparisons (3 pulling force levels x 10 subjects). Specifically, lumbar torque_{peakPF} changed significantly in seven subjects for the 20% pulls, in nine subjects for the 40% pulls, and in eight subjects for the 80% pulls. Two subjects significantly decreased lumbar torque_{peakPF} for all three force levels.

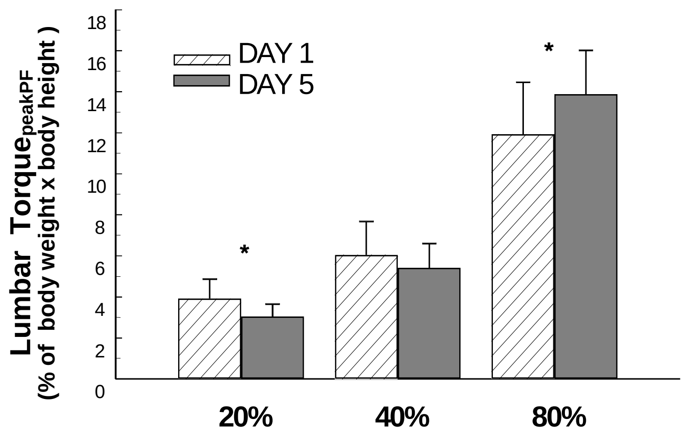

Figure 3. Group means and 95% confidence intervals for lumbar torquepeakPF, on Day 1 and Day 5, for 20, 40, 80% pulls. Torque units (y-axis) are expressed as a percentage of body weight and height.

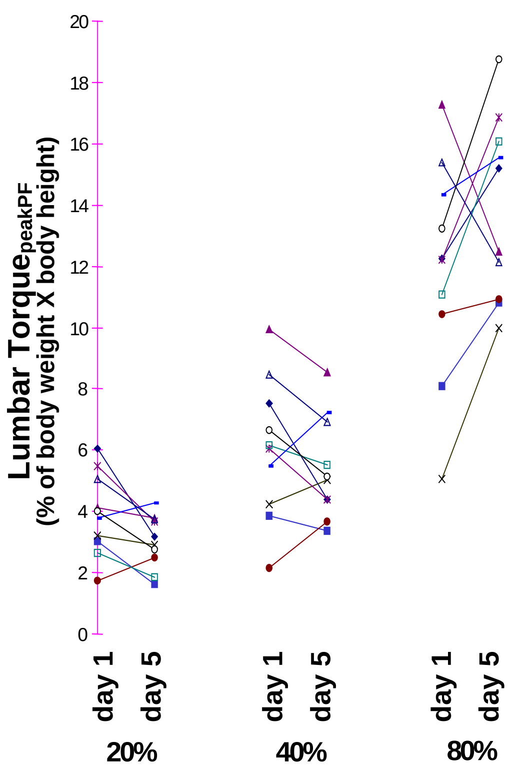

Figure 4. Mean lumbar torquepeakPF for each subject, on both days and all three pulling forces. Each subject's data are represented by unique symbols connected by a line between Day 1 and Day 5 for each pulling force level.

Relationship between peak pulling force and lumbar torque

The results supported the hypothesis that pulling force and lumbar torque$_{peakPF}$ would be more strongly correlated after practice. Coefficients of determination between peak pulling force and lumbar torque$_{peakPF}$ were significantly higher on Day 5 than Day 1, t(9) = -2.03, P = .04 (Day 5 mean $r^2_{PF,LT}$ = 0.90, range = 0.81 to 0.95; Day 1 mean $r^2_{PF,LT}$ = 0.81, range = 0.22 to 0.96). Figure 5a shows the scatterplot of peak pulling force vs lumbar torque$_{peakPF}$ before and after practice for subject D, who had the greatest change in $r^2_{PF,LT}$ between Day 1 and Day 5. Four subjects had $r^2_{PF,LL}$ that exceeded 0.90 on Day 1 (including the two subjects whose lumbar torque$_{peakPF}$ for 80% pulls decreased with practice), leaving little room for further increases. However, even these subjects showed practice effects, because their high correlations emerged between the first and second halves of Day 1. The scatterplot of peak pulling force vs lumbar torque$_{peakPF}$ for the first and last three blocks of trials on Day1 for subject F exemplifies an increased correlation within Day 1 (Figure 5b).

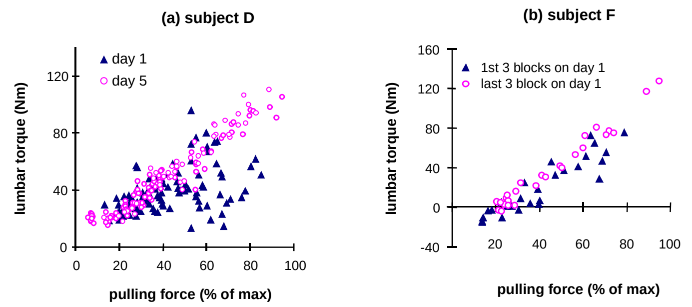

Figure 5. (a) Scatterplot of pulling force vs lumbar torquepeakPF on Day 1 and Day 5 for subject D. Correlations were higher on Day 5 than Day 1. (b) Scatterplot of pulling force vs lumbar torquepeakPF on the first and last 3 blocks of Day 1 for subject F. Correlations were higher for the last than first 3 blocks.

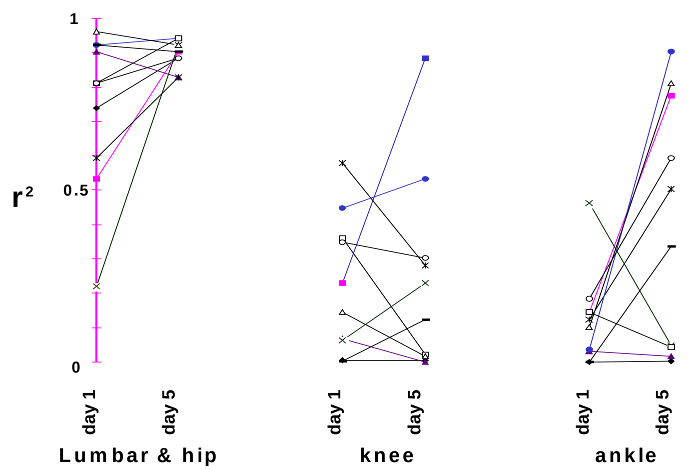

Figure 6. Coefficients of determination (r2PF,LT) between peak pulling force and joint torque at the time of peak pulling force for the lumbar, hip, knee, and ankle joints. Each subject is represented by a unique symbol, with a line connecting the Day 1 and Day 5 data points.

Practice-related increases in the correlation between pulling force and joint torque could be selective for the lumbar joint. Alternatively, increases also could occur for all lower extremity joints. Increased correlations between pulling force and all joint torques would suggest that subjects simply learned to use more consistent movement patterns, not motor control strategies specific to lumbar loading. To determine if practice effects were selective or general, we evaluated how practice altered the correlations between pulling force and hip, knee, and ankle torques, as well as lumbar torque. Figure 6 shows the r^2 between pulling force and lower extremity joint torques, measured at the time of peak pulling force. Lumbar and hip torque correlations with pulling force are presented as one category, because they were always highly correlated ($r^2 > 0.98$). The r^2 between pulling force and knee torque was low on both days. The r^2 between pulling force and ankle torque increased with practice (t(9) = 2.19, P < .05), but less than for lumbar torque. Only two subjects had r^2 between pulling force and ankle torque after practice that exceeded 0.81, which was the lowest value for $r^2_{PF,LT}$ on Day 5. Thus, subjects selectively increased the correlation between lumbar torque and peak pulling force.

Discussion

This study showed that persons without history of low back pain changed the magnitude and consistency of their lumbar loading after five days of practice on a novel pulling task that involved relatively unrestrained body motion[25,26] and highly nonlinear dynamics[10]. Subjects spontaneously learned to alter at least two aspects of motor control related to lumbar loading without any instructions or feedback about their movements. These observations extend previous reports that experience and practice influence lumbar loading during trunk loads imposed upon subjects whose leg and pelvic motions were restrained[9,19,22,23,30]. The present study also revealed individual variations in how practice influenced lumbar loading.

Practice-related changes in the magnitude of lumbar torque

The hypothesis that subjects would learn to reduce lumbar loading was only partly supported. After practice, lumbar torque was significantly lower for the 20% pulls, suggesting that subjects learned to organize those movements to minimize lumbar loading. Lumbar torque did not change consistently for the 40% pulls, although 60% of the subjects had lower lumbar torque after practice. Lumbar torque increased for the 80% pulls. That increase was unexpected, because the largest pulls have the greatest potential to overload the lumbar spine and intuitively would seem to be where subjects might try most to decrease lumbar loading.

The finding that practice influenced lumbar torques more for small than large external loads is consistent with previous reports that evaluated lumbar torque changes during MMH tasks of varying dynamics. For example, lifting techniques influenced peak lumbar torque only for low and moderate speed but not fast lifts[5]. Similarly, Hall[31] reported no significant differences in the L5/S1 torque for very fast lifting movements with 40, 60, and 80% of maximum loads. Moreover, the studies of lifting that found lower lumbar torques in experienced than inexperienced subjects[15,18,19] used loads that were equal to or less than the 20% pulling forces used in our study. No previous studies on practice used loads as great as our average 40% target pulling forces**.** One reason that practice and other experiential factors do not consistently reduce lumbar torques for highly dynamic MMH tasks may be that such tasks are tightly constrained by the task requirements. During standing pulls, for example, the constraints of pulling force accuracy and balance recovery during the 80% pulls might be major determinants of lumbar torque, leaving little room for practice to reduce lumbar loading. Nonetheless, task constraints for the 80% pulls did not *totally* prevent a reduction in lumbar loading, because two subjects reduced lumbar torque for the 80% pulls after practice.

The increased lumbar loading after practice for the 80% pulls in most subjects does not necessarily imply that motor control related to lumbar loading was worse after practice for those pulls. As discussed next, the lumbar loading was more consistent after practice, effectively reducing the motor control 'error' of making occasional pulls with extreme lumbar loading.

Relationship between peak pulling force and lumbar torque

Previous studies reported that experience and the predictability of upcoming external loads influence lumbar loading, the latencies of reactive responses in trunk muscles, and trunk kinematics[7,8,15,18,19,22]. Such findings suggest that the lumbar spine is controlled better after subjects have experience with and presumably can better anticipate the external load. The present finding that practice increased the consistency of lumbar loading (shown by higher correlations between peak pulling force and lumbar loading after practice) supports the idea that experience enhances lumbar control, for the following reason. Early in practice, most subjects had lumbar torques that varied considerably for each pulling force as shown by relatively low correlations between peak pulling force and lumbar torque$_{peakPF}$. After practice, subjects had higher correlations between peak pulling force and lumbar torque$_{peakPF}$, as well as greater accuracy in producing the pulls[32]. The higher correlations could, in principle, enable subjects to better anticipate and counteract the effects of the lumbar loading for an intended pull.

As was the case with practice-related changes in the magnitude of lumbar loading, subjects differed in how much practice they needed before the correlations between peak pulling force and lumbar torque increased. The two subjects whose lumbar torques were lower after practice for all three pulling force levels showed an early emergence of high correlations between peak pulling force and lumbar torque$_{peakPF}$ on Day 1. Such high correlations could have enabled them to anticipate what lumbar loading would occur during a given pull even on Day 1, hypothetically freeing them to use the remaining practice to learn how to minimize lumbar loading for the high as well as low pulling forces. In contrast, subjects who achieved high correlations between pulling force and lumbar torque only after five days had higher lumbar torques for 80% pulls on Day 5 than Day 1.

Such observations suggest that there may be a preferred sequence for learning to accommodate lumbar loading. First, people may learn to organize their movements to achieve high correlations between peak pulling force and lumbar torque, potentially making the lumbar loading easier to anticipate and control. Once that relationship is established, people could learn to better counteract the external load and reduce lumbar torque. This hypothesis could be tested by evaluating whether subjects who keep practicing high force pulls after achieving high correlations between peak pulling force and lumbar torque$_{peakPF}$ eventually reduce lumbar torque.

An alternative explanation for the increased correlations between peak pulling force and lumbar torque$_{peakPF}$ is that subjects learn to more consistently load all joints. While this possibility cannot be completely ruled out, it seems unlikely because the correlations of the knee joint torque and pulling force did not change with practice, and because the changes in correlations between ankle joint torque and pulling force were smaller and less consistent than for lumbar torque. A second alternative is that the timing of lumbar muscle activation might have changed, resulting in greater synchronization between the lumbar torque and peak pulling force time histories. Future studies should test these alternative explanations.

Individual differences in practice effects

The finding that individuals differed in how practice influenced lumbar torque is consistent with a report by Lavender and colleagues[19], who reported differences in how practice influenced trunk muscle and postural responses for four subjects who were exposed to repeated trunk loading. Factors such as gender, anthropometrics, strength, flexibility, and previous experience could influence how subjects learn to control lumbar torques as they practice MMH or other tasks that load the lumbar spine. For example, persons with previous experience with activities that rapidly load the trunk might learn to modify lumbar loading faster than subjects with little or no experience. This idea is supported by studies that found different kinematics and lumbar loading between novice and expert subjects on MMH tasks[15,18]. Further studies are needed to clarify how experience and other individual factors influence practice-related changes in lumbar torques.

Limitations of the study

It was beyond the scope of this study to evaluate other practice-related changes in the motor control of lumbar loading, such as L5/S1 joint shear and compression forces, even though such changes are important in understanding lumbar control mechanisms. We focused on lumbar torques because they are one of the biomechanical risk factors for low back injury[33,34] and because they could be estimated with reasonable accuracy using kinematic and force plate data and inverse dynamics. Two factors contributed to our decision not to estimate shear and compression forces at L5/S1. First, such estimations require accurate measures of the spatial orientation of the vertebra, which could change rapidly as the pelvis and trunk translate and rotate during the brief pull. The accuracy of such measures is questionable using camera-based motion analysis of surface markers, when whole-body kinematics is recorded. Second, models that have been used to estimate lumbar shear and compression forces make numerous assumptions[29,35] that have not been validated for the very brief, highly dynamic standing pulls. Consequently, model-based estimates of L5/S1 shear and compression forces could not be made with confidence, even if multiple trunk muscle recordings had been made. A second limitation is that the study did not directly evaluate how the rate of lumbar loading changed with practice. A lower rate of loading could reduce the chances of injury during pulling. A third limitation is that the restrictions on foot placement and the immobility of the load may have precluded subjects from adopting some strategies to further alter lumbar loading in beneficial ways. Future studies thus should assess if practice influences L5/S1 shear and compression forces and rates of change of lumbar loading, as well as lumbar torque, in tasks where movements are somewhat restricted (e.g., standing pulls) and in MMH task without such restrictions.

Conclusions and future directions

This study showed that subjects given five days of practice on the novel dynamic pulling task spontaneously changed the magnitude and consistency of their lumbar torques. The specific neuromuscular processes that mediate those motor control changes remain to be determined. Altered timing of muscle activation patterns may mediate some of the practice-related changes in lumbar torque. One specific hypothesis is that before practice, novices activate their trunk muscles after starting to pull, reflecting reliance on reactive motor control; whereas after practice, they might activate their trunk muscles before pulling, reflecting an increased reliance on predictive control. Changes in the magnitudes, durations and rates of change of trunk muscle activity also could mediate effects of practice on lumbar loading. Future studies should determine how variations in kinematics and muscle activation patterns during pulling, and changes in those patterns with practice, influence lumbar loading.

Finally, practice resulted in individually varied changes in the magnitude and consistency of lumbar loading. Such variations suggest that workplace trainees might benefit by practicing handling, at their own pace, the full range of loads they will have to deal with on the job. However, people should be warned to avoid maximum loads during unsupervised practice. Some benefits of practice may accrue spontaneously, even without standard "back school" programs (which do not always reduce the probability of injury in workers performing MMH[36,37]). Moreover, it may be appropriate to determine training time on an individual basis, perhaps quantified by reductions in lumbar torque for the expected range of loads to be handled, or by changes in the consistency of lumbar loading during various external loads. Allowing workers enough time to practice novel tasks may help them develop motor control strategies for handling expected loads, thus reducing the risk of low back injury.

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References

1. Brinckmann P, Biggemann M, Hilweg D. Prediction of the compressive strength of human lumbar vertebrae. Clinical Biomechanics 1989;4.
2. Hutton WC, Adams MA. Can the lumbar spine be crushed in heavy lifting? Spine 1982;7(6):586-90.
3. de Looze MP, Kingma I, Bussmann JBJ, Toussaint HM. Validation of a dynamic linked segment model to calculate joint moments in lifting. Clinical Biomechanics 1992;7:161-169.
4. Lee KS, Chaffin DB, Waikar AM, Chung MK. Lower back muscle forces in pushing and pulling. Ergonomics 1989;32(12):1551-63.
5. Bush-Joseph C, Schipplein O, Andersson GB, Andriacchi TP. Influence of dynamic factors on the lumbar spine moment in lifting. Ergonomics 1988;31(2):211-6.
6. McGill SM. The biomechanics of low back injury: implications on current practice in industry and the clinic. Journal of Biomechanics 1997;30(5):465-75.
7. Commissaris DA, Toussaint HM. Load knowledge affects low-back loading and control of balance in lifting tasks. Ergonomics 1997;40(5):559-75.
8. Butler D, Andersson GB, Trafimow J, Schipplein OD, Andriacchi TP. The influence of load knowledge on lifting technique. Ergonomics 1993;36(12):1489-93.
9. Marras WS, Rangarajulu SL, Lavender SA. Trunk loading and expectation. Ergonomics 1987;30(3):551-62.
10. Schneider K, Zernicke RF, Schmidt RA, Hart TJ. Changes in limb dynamics during the practice of rapid arm movements. Journal of Biomechanics 1989;22(8-9):805-17.
11. Uno Y, Kawato M, Suzuki R. Formation and control of optimal trajectory in human multijoint arm movement. Minimum torque-change model. Biological Cybernetics 1989;61(2):89-101.
12. Andersson G, Winters J. Role of muscle in postural tasks: Spinal loading and postural stability. In: Winters J, Woo S-Y, editors. Multiple Muscle System: Biomechanics and Movement Organization. New York: Springer-Verlag; 1990. p. 377-393.
13. Gracovetsky S, Farfan H. The optimum spine. Spine 1986;11(6):543-73.
14. Schultz AB, Andersson GB. Analysis of loads on the lumbar spine. Spine 1981;6(1):76-82.
15. Gagnon M. Box tilt and knee motions in manual lifting: Two differential factors in expert and novice workers. Clinical Biomechanics 1997;12:7-8.
16. Potvin JR, McGill SM, Norman RW. Trunk muscle and lumbar ligament contributions to dynamic lifts with varying degrees of trunk flexion. Spine 1991;16(9):1099-107.
17. Troup JD, Leskinen TP, Stalhammar HR, Kuorinka IA. A comparison of intraabdominal pressure increases, hip torque, and lumbar vertebral compression in different lifting techniques. Human Factors 1983;25(5):517-25.
18. Patterson P, Congleton J, Koppa R, Huchingson RD. The effects of load knowledge on stresses at the lower back during lifting. Ergonomics 1987;30(3):539-49.
19. Lavender SA, Marras WS, Miller RA. The development of response strategies in preparation for sudden loading to the torso. Spine 1993;18(14):2097-105.
20. Garg A, Moore JS. Epidemiology of low-back pain in industry. Occupational Medicine 1992;7(4):593-608.
21. Manning DP, Mitchell RG, Blanchfield LP. Body movements and events contributing to accidental and nonaccidental back injuries. Spine 1984;9(7):734-9.
22. Wilder DG, Aleksiev AR, Magnusson ML, Pope MH, Spratt KF, Goel VK. Muscular response to sudden load. A tool to evaluate fatigue and rehabilitation. Spine 1996;21(22):2628-39.
23. Lavender SA, Mirka GA, Schoenmarklin RW, Sommerich CM, Sudhakar LR, Marras WS. The effects of preview and task symmetry on trunk muscle response to sudden loading. Human Factors 1989;31(1):101-15.
24. Lee WA, Patton JL. Learned changes in the complexity of movement organization during multijoint, standing pulls. Biological Cybernetics 1997;77(3):197-206.
25. Lee WA, Russo AM. Constraints and coordination in whole body action. In: Swinnen SP, Heuer H, Massion J, Casaer P, editors. Interlimb Coordination: Neural, Dynamical and Cognitive Constraints. New York: Academic Press; 1994. p. 537-569.
26. Michaels CF, Lee WA, Pai Y-C. The organization of multisegmetal pulls made by standing humans: I. Near-maximal pulls. Journal of Motor Behavior 1993;25(2):107-124.
27. D'Amico M, Ferrigno G. Technique for the evaluation of derivatives from noisy biomechanical displacement data using a model-based bandwidth-selection procedure. Medical & Biological Engineering & Computing 1990;28(5):407-15.
28. Andres RO, Chaffin DB. Validation of a biodynamic model of pushing and pulling. Journal of Biomechanics 1991;24(11):1033-45.
29. McGill SM, Norman RW. Partitioning of the L4-L5 dynamic moment into disc, ligamentous, and muscular components during lifting. Spine 1986;11(7):666-78.
30. Callaghan JP, McGill SM. Muscle activity and low back loads under external shear and compressive loading. Spine 1995;20(9):992-8.
31. Hall SJ. Effect of attempted lifting speed on forces and torque exerted on the lumbar spine. Medicine & Science in Sports & Exercise 1985;17(4):440-4.
32. Lee WA, Russo A, Pai Y-C, Schena D. Schema learning in multijoint pulling. In: Neuroscience Abstract 1994;20:995.
33. Jorgensen MJ, Davis KG, Kirking BC, Lewis KE, Marras WS. Significance of biomechanical and physiological variables during the

determination of maximum acceptable weight of lift. Ergonomics 1999;42(9):1216-32.
34. Marras WS, Lavender SA, Leurgans SE, Fathallah FA, Ferguson SA, Allread WG, et al. Biomechanical risk factors for occupationally related low back disorders. Ergonomics 1995;38(2):377-410.
35. McGill SM, Norman RW, Cholewicki J. A simple polynomial that predicts low-back compression during complex 3-D tasks. Ergonomics 1996;39(9):1107-18.
36. Daltroy LH, Iversen MD, Larson MG, Lew R, Wright E, Ryan J, et al. A controlled trial of an educational program to prevent low back injuries. New England Journal of Medicine 1997;337(5):322-8.
37. Turner JA. Educational and behavioral interventions for back pain in primary care. Spine 1996;21(24):2851-7.