

Electromyographic Studies of the Lumbar Trunk Musculature During the Generation of Low-Level Trunk Acceleration

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Summary: An understanding of how the support mechanisms of the spine behave during lifting may yield insight into the loading of the spine under occupational conditions and help shed light on the etiology of low-back disorders. Previous controlled laboratory studies of spinal loadings have been limited to isometric and isokinetic conditions. To evaluate the behavior of the trunk during acceleration, we recorded intra-abdominal pressure and trunk muscle activities during low-level acceleration. Twenty subjects performed controlled accelerations of the trunk under different trunk loading conditions. Muscle activity decreased as acceleration increased; however, the rate of decrease differed among muscles (mean decrease, $\leq 1\%$ of maximum per $10^\circ/s^2$ increase in acceleration), with the activity of the erector spinae muscles decreasing the most (1.88% of the maximum per $10^\circ/s^2$ increase in acceleration). No changes in intra-abdominal pressure were found as a function of acceleration. Relative coactivation of the muscles increased; however, this was a function of increases in trunk velocity and torque.

In early studies of trunk muscle activity, electromyographic activity of the trunk muscles was observed while subjects assumed various isometric lifting postures (1-3). The findings showed that muscles behave as a function of the magnitude of the load lifted, the distance of the load from the spine, and the position of the trunk during lifting. Under these isometric conditions, the agonist muscles were active, and little antagonist coactivation was observed. Later studies explored the activity of the trunk muscles during the generation of isokinetically controlled trunk velocity. Such studies showed that electromyographic activity of trunk muscles and coactivity of muscles increased as trunk velocity increased (8,10,14). The effect of this increase in coactivity was an increase in spinal loading as a result

of an increase in the magnitudes of compressive and shear forces (12,13).

Under true dynamic lifting conditions, angular acceleration of the trunk also is present. We hypothesize that, for the trunk to accelerate, a reduction in the coactivity and stiffness of the trunk muscles is necessary so that the trunk can move more ballistically. Only one study (11) documented the activity of the trunk musculature during constant acceleration of the trunk in three dimensions; with increasing acceleration, coactivation of muscles increased, and the trunk muscles with the greatest mechanical advantage had the most increase in activity. However, due to technical constraints with the dynamometer used to test acceleration, there were two limitations to the study that may alter the coactive behavior of the musculoskeletal system. First, the effects of trunk acceleration could not be separated from the effects of trunk velocity and muscle lengthening, because different subjects produced approximately constant accelerations at subjectively determined levels of

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low, medium, and high rates of acceleration. Under these conditions, it was difficult to determine how muscle activity related to muscle force. Second, the effects of trunk acceleration could not be examined while the subjects were generating a torque about the trunk, as would be the case during lifting. Therefore, it is not known if the trunk musculature would exhibit the same coactivation characteristics under more realistic lifting conditions, in which acceleration is developed while trunk torque is produced.

Current advances in dynamometer control now provide a means to overcome these problems partially. Therefore, the objectives of the current study were to determine if the muscles of the trunk behave differently during generation of controlled acceleration within the trunk under load and, if they do, to determine how.

METHODS

The subjects were 20 male volunteers, ranging in age from 21-34 years. The average height and weight were 182 cm (range: 163-197 cm) and 83 kg (range: 60-127 kg). The average length of the spine (from L5-S1 to C1) was 60.3 cm (range: 53.0-70.5 cm). The average trunk breadth was 30.5 cm (range: 21-38 cm) and the average trunk depth, 22.5 cm (range: 20-26 cm). None of the subjects had had a significant low-back disorder. Their experience in manual materials-handling varied.

The experimental design controlled four motion components (independent variables) while the behavior of 11 internal trunk structures (dependent variables) was observed. The factors controlled in this study consisted of (a) trunk position, (b) external trunk torque, (c) angular trunk velocity, and (d) angular trunk acceleration.

Trunk position was defined as the position in space of the trunk relative to the pelvis. Six combinations of trunk positions in the sagittal (20 and 40° bend) and transverse (0, 15, and 30° twist) planes were determined so that the effect of muscle length could be controlled. Trunk torque was the external torque that was exerted by the subject on the dynamometer lever arm in contact with the upper back. The levels of exertion were 0, 54, and 108 Nm of torque.

Angular trunk extension velocity was controlled by an isokinetic dynamometer and was defined as the angular bending velocity of the trunk about L5-S1. The levels of this variable were 0, 15, and 30°/s. Angular trunk extension acceleration was controlled

by the same dynamometer and was defined as the angular acceleration of the trunk about L5-S1. This was made possible by new software developed for this research that precisely controls angular acceleration. The levels of this variable were 0, 20, and 40°/s².

Thus, the experiment was designed so that subjects experienced different levels of acceleration as the trunk moved through the same positions in space at the same velocities and under the same torque conditions. In this manner, the effects of acceleration could be distinguished from the effects of the other motion variables. To ensure that combinations of the specific levels of angular velocity and acceleration occurred at the specific trunk angles, the starting position was calculated and controlled for each exertion. For example, if a given trial dictated that the subject was to be moving at 30°/s at the sagittal trunk angle of 20° with acceleration of 20°/s², then the acceleration began at a trunk angle of 42.5°. In this way, all requirements of position, velocity, and acceleration were met for a given trial.

The dependent variables measured were intra-abdominal pressure and the normalized processed electromyographic activities of the 10 trunk muscles identified as the major contributors to spinal loading (20): the right and left latissimus dorsi, the right and left erector spinae, the right and left rectus abdominis, and the right and left external and internal oblique muscles. In the reference frame dynamometer (KINCOM; Chattanooga, Hixson, TN, U.S.A.) (Fig. 1) used to control trunk motion and monitor trunk torque, the axis of rotation of L5-S1 was aligned with the axis of rotation of the dynamometer. This positioning was performed so that the dynamometer was aligned with L5-S1 in both symmetric and asymmetric conditions.

Two load cells were placed on the ends of a horizontal bar that was in contact with the subject's back. This bar was connected to the dynamometer lever arm and permitted the measurement of bending moment as well as twisting moment throughout trunk extension. The subjects were able to monitor the production of torque continuously through a feedback system consisting of a computer screen that graphically displayed the current level of torque production online. A target torque was shown on the computer screen, as was a tolerance of $\pm 10\%$ about the target torque. The dynamometer limited velocity and acceleration. Pilot studies indicated that position, velocity, and acceleration are controlled within

a 1% tolerance. Thus, during the experimental task, the subjects needed to control only torque, and the dynamometer limited the amount of velocity and acceleration that could be produced for as long as torque was applied.

The processed electromyographic activities of the trunk muscles were monitored by surface electrodes attached to small, lightweight preamplifiers. The preamplifiers were mounted on a belt that was fitted around the subject's waist. This configuration was used in an attempt to minimize the amount of external noise in the recorded signal. The pre-

The dynamometer signals (position, velocity, acceleration, and torque), processed electromyographic signals, and the intra-abdominal pressure signal all were digitized with an analog-to-digital converter at 100 Hz. The multichannel analog-to-digital system interfaced with a 386-based microcomputer to collect, display, and store the data.

The experimental procedure involved a practice session to familiarize the subject with the experimental task of controlling torque by interfacing with the computer feedback system. Experimental sessions were performed on the day subsequent to the prac-

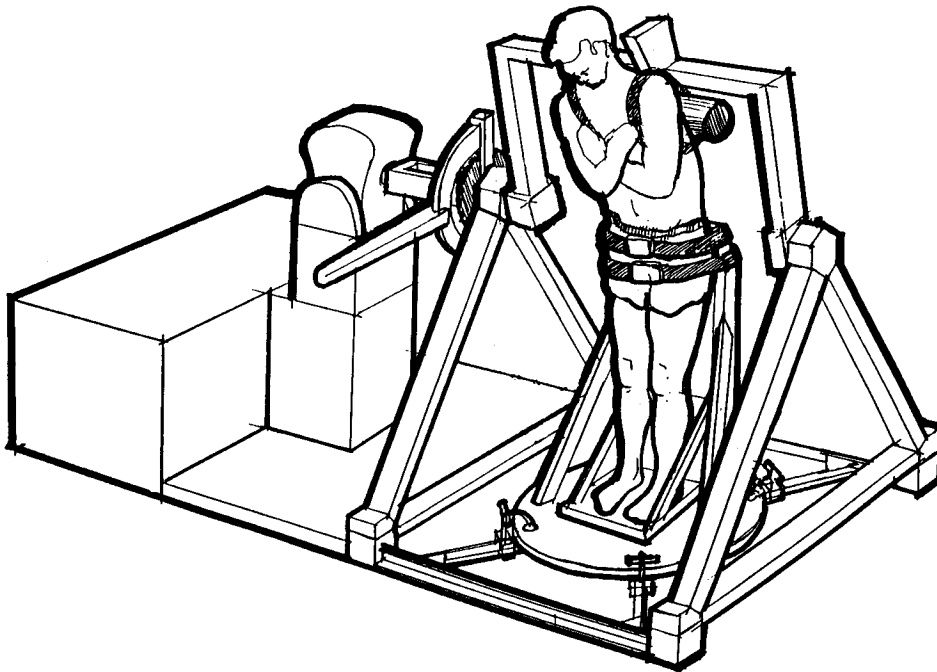


FIG. 1. Reference frame and dynamometer used to control trunk position, velocity, and acceleration.

amplifiers were connected to electromyographic amplifiers, filters, and processing units. The electromyographic signal was low-pass filtered at 1,000 Hz and high-pass filtered at 80 Hz. The signal was rectified and averaged with a time-constant window of 20 ms; this served as the processed signal. The intra-abdominal pressure was monitored with a telemetric pressure transducer pill (Remote Control Systems, London, England) inserted rectally. Rushmer (19) found that intrarectal pressure was essentially equivalent to intra-abdominal pressure. An antenna worn around the subject's trunk received the intra-abdominal pressure signal. The range of pill pressure was ± 13.3 kPa, and drift was <0.03 kPa.

tice day. Prior to testing, maximum and minimum activities of each muscle in each trunk position were collected for electromyography normalization purposes. The experimental task required the subjects to control the trunk torque within the $\pm 10\%$ torque tolerance throughout the exertion (defined on the computer screen) under each condition. If the subject failed to maintain an exertion within the $\pm 10\%$ torque tolerance, the trial was repeated.

During data analysis, each dependent measure was evaluated as the trunk moved through a "trunk position window," to control for the effects of changes in muscle length. These windows consisted of the forward trunk angle (20 or 40°) $\pm 1^\circ$ of sagittal mo-

tion. Each value, therefore, represented the mean activity as the trunk passed through a 2° range of motion. All data were normalized with respect to the static maximum and minimum electromyographic activity of the muscle, collected while the subject assumed that specific combination of trunk angle and asymmetry.

Multivariate analysis of variance (ANOVA) was employed to determine whether acceleration influenced the collective activity of the trunk muscles, which would indicate a change in coactivity. Univariate ANOVA was used as a follow-up procedure to

affected (Fig. 2). However, the rates of decrease varied among muscles. The decrease in the signal amplitude of the erector spinae muscles was 1.88% of the maximum for each $10^\circ/\text{s}^2$ increase in acceleration. The rates of reduction in signal amplitude for the internal oblique and latissimus dorsi muscles were 0.78 and 0.98%, respectively, of the maximum for each $10^\circ/\text{s}^2$ increase in acceleration. No statistically significant change in intra-abdominal pressure was observed as acceleration increased.

In addition, the combined effect of torque and velocity influenced the coactive behavior of the

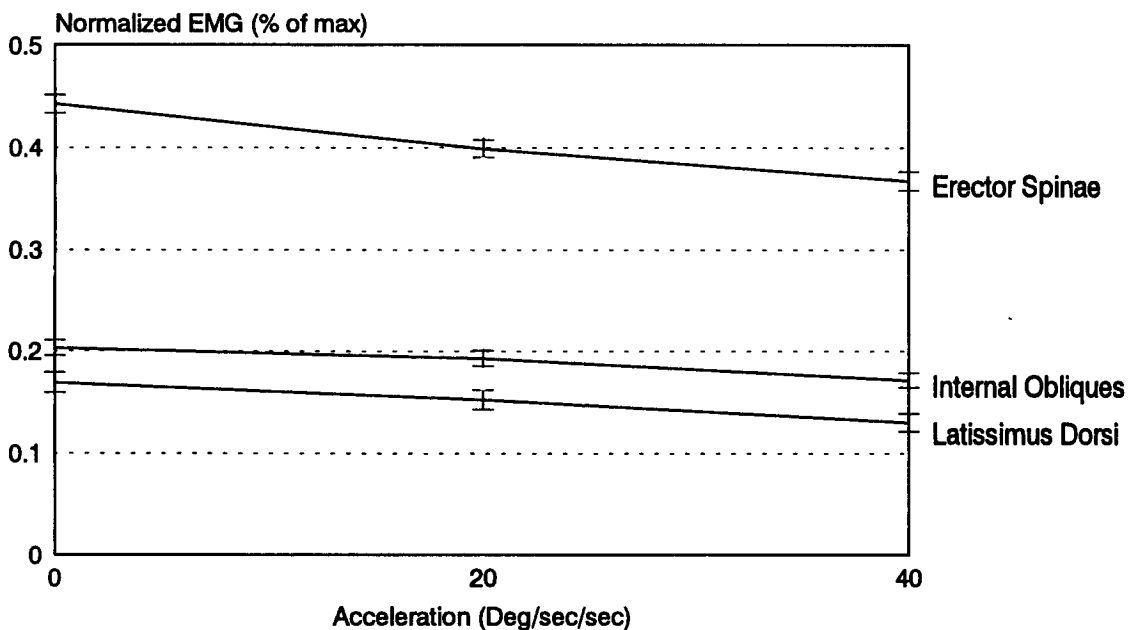


FIG. 2. The effects of increased acceleration on the activity of trunk muscles. Vertical bars indicate SD. EMG = electromyography.

identify changes in the activity levels of the individual muscle groups or in intra-abdominal pressure, or in both, as acceleration increased.

RESULTS

As trunk acceleration increased, a change in coactivation among the muscles was noted ($F = 3.778$; $p < 0.0001$). Subsequent univariate analyses revealed that the activities of most muscles were affected by acceleration (only the left rectus abdominis and left external oblique muscles were not affected). Post hoc analyses indicated that increased acceleration resulted in a decrease in the amplitudes of the electromyographic signals of those muscles significantly

muscle groups (Fig. 3A and B). The increases in the myoelectric signal amplitude of the erector spinae muscles were approximately equal among the various torque conditions over all velocity conditions (Fig. 3A). However, the activities of the other trunk muscles increased disproportionately among torque levels. For example, the activity of the latissimus dorsi muscle increased more between the 54 and 108 Nm conditions than it did between the 0 and 54 Nm conditions (Fig. 3B). This pattern indicates a relative increase in coactivity as torque increases, and this coactivation increases further with increasing trunk velocity.

The variability in muscle activity also increased more for the coactivating muscles (such as the latis-

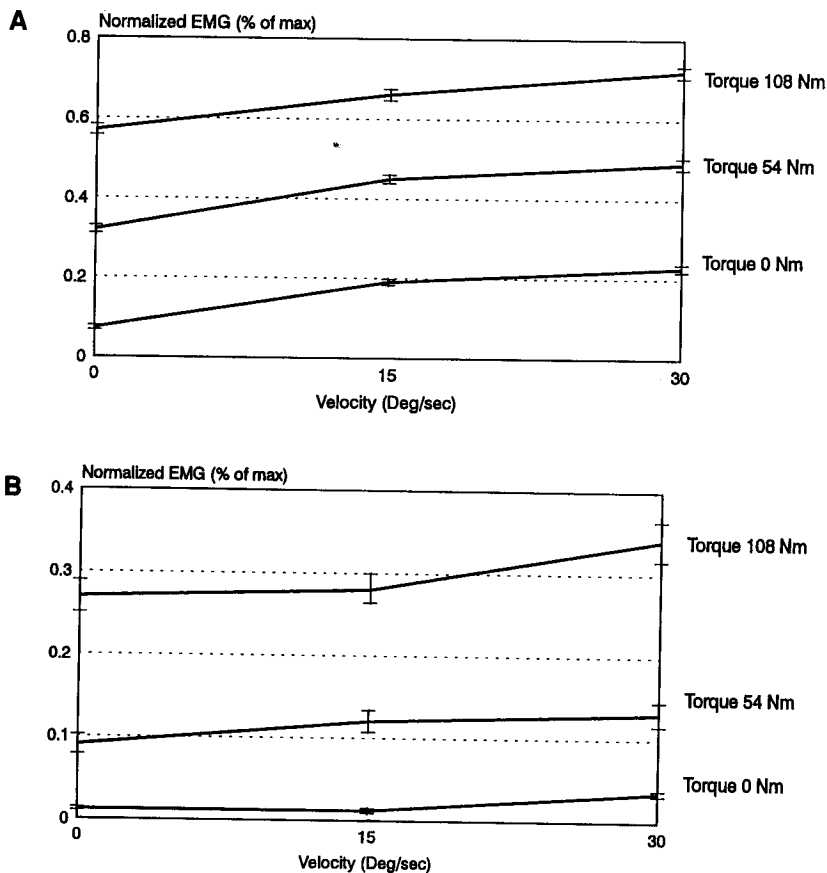


FIG. 3. Change in muscle activity as a function of velocity and trunk torque for the (A) erector spinae muscle group and (B) an example of other trunk muscle groups (latissimus dorsi). Note the difference in the incremental increase between the 0 and 54 Nm conditions compared with that between the 54 and 108 Nm conditions. Vertical bars indicate SD. EMG = electromyography.

simus dorsi muscles) than for the erector spinae muscles as velocity and torque increased (Fig. 3A and B). This variability may directly affect spinal loading patterns (17).

DISCUSSION

The study showed that low-level trunk acceleration under loaded conditions caused a reduction in the activity of trunk muscles. However, this reduction did not occur equally among all muscles as acceleration increased. The activity of the erector spinae muscles was reduced to a greater extent than the activity of other trunk muscles. The activity of the other muscles was not reduced to the same degree until greater torques were generated by the trunk. Many of these muscles are oriented in oblique directions; hence, this pattern can result in a relative change in coactivity of the muscles, which could change the shear loading on the spine (12,13,16). Furthermore, the spine-stiffening action of the erector spinae muscles is reduced due to less activity

of these muscles, and this also increases the development of shear force. A significant amount of evidence has indicated that shear forces can be particularly damaging to disc fibers (4-7,18,21). Thus, the effects of low-level acceleration on spinal loading appears to introduce trade-offs between the nature of loadings on the spine. This may help us to understand how injuries to the spine occur.

Intra-abdominal pressure was not affected by the accelerations examined in the current study; this confirms that intra-abdominal pressure is influenced by only trunk torque and trunk position. These findings were similar to those of Marras and Mirka in their study of intra-abdominal pressure and velocity effects (14). Thus, intra-abdominal pressure appears to have little biomechanical relationship to motion and may be a byproduct of changes in trunk volume during changes in posture and level of exertion.

The current study brings us one step closer to an understanding of how to interpret trunk muscle activity during free dynamic motion and demonstrates that the trunk's musculoskeletal control system is

complex and not easily predictable. It is necessary to understand such motions to determine how occupational behavior affects loading on the spine and how patients with low-back disorders compensate for injuries through muscular control.

It also should be emphasized that, even though this study is the only one to examine trunk muscle activity under controlled acceleration conditions, it still does not represent typical realistic lifting conditions. The motions employed by the subjects were not truly free dynamic, as are those involved in lifting. Due to the limitations of the dynamometer, we could explore only a very low range of accelerations. Thus, these results cannot be extrapolated to a situation beyond the $40^\circ/\text{s}^2$ condition. Such accelerations correspond to the very lowest accelerations observed in industrial work (15). Future studies must find ways to control experimental motion conditions over a far greater range of accelerations.

Another factor that may affect the results of this experiment is the inability of the subjects to control trunk torque exactly throughout the experimental exertion. In this experiment, a 10% torque exertion tolerance range was permitted. The variance in torque applied by the subject over the course of the experiment may contribute to some of the observed effects. However, the influence of this fluctuation in torque is expected to be minimal, as the torque tolerance was defined in terms of the target torque level, not the maximum torque level. In addition, similar effects of acceleration were observed under all torque conditions, even 0 load. Thus, we believe that the trends reported here were true effects of acceleration.

The trends noted here were not the same as those found in the previous study of dynamometer-controlled acceleration (11). In the earlier investigation, the mass of the trunk was accelerated at levels that were subjectively classified by the subject as low, medium, and high. This produced a wide range of accelerations that were more realistic, and the activity of the muscles farthest from the spine increased by the greatest amount. The finding that the activity of the erector spinae muscles was less involved with increases in trunk acceleration is consistent with the present findings. However, the accelerations in the current study are equal to or less than the subjective low level of acceleration in the earlier study. Thus, these two investigations collectively show that the activities of the muscles during trunk acceleration are nonlinear. They decrease at low levels of accel-

eration and then may increase in an approximate linear fashion (11).

Finally, this information might be useful as a clinical assessment for patients who have low-back disorders. The motion patterns of such patients often are much slower, presumably due to guarding behavior (9). Research in our laboratory has shown that rates of acceleration in patients with low-back pain are comparable with the accelerations tested here. Therefore, it may be possible to compare the muscle activities in such patients with those in the present subjects. This may help to explain how the trunk's musculoskeletal control system changes during injury and to pinpoint the areas of greatest sensitivity in an impaired spine. Such an experiment would be feasible, especially since the study indicated that there is no need to include high levels of torque, and would provide an indicator of the status of the trunk's natural muscular control system that would not be masked by production of torque. As acceleration is mechanically a higher derivative of position, such a test would provide a measure that could not be cognitively controlled by the patient.

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