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Decrease in Trunk Muscular Response to Perturbation With Preactivation of Lumbar Spinal Musculature

Ian A.F. Stokes, PhD,* Mack Gardner-Morse, MS,* Sharon M. Henry, PT, PhD,† and Gary J. Badger, MS‡

Study Design. An experimental study of healthy subjects' trunk muscle responses to force perturbations at differing angles and steady state efforts.

Objectives. To determine whether increased preactivation of muscles was associated with decreased likelihood of muscular activation in response to a transient force perturbation.

Summary of Background Data. Trunk stability (ability to return to equilibrium position after a perturbation) requires the stiffness of appropriately activated muscles to prevent buckling and consequent "self-injury." Therefore, greater trunk muscle preactivation might decrease the likelihood of reflex muscle responses to small perturbations.

Methods. Each of 13 subjects stood in an apparatus with the pelvis immobilized. A harness around the thorax provided a preload and a force perturbation by a horizontal cable and a movable pulley attached to one of five anchorage points on a wall track surrounding the subject at angles of 0°, 45°, 90°, 135°, and 180° to the forward direction. Subjects first equilibrated with a preload effort of nominally 20% or 40% of their maximum extension effort. Then a single full sine-wave force perturbation pulse of nominal amplitude, 7.5% or 15% of maximum effort, duration 80 milliseconds or 300 milliseconds, was applied at a random time, with three repeated trials of each test condition. The applied force (*via* a load cell) and the electromyographic activity of six right and left pairs of trunk muscles were recorded. Muscle responses were detected by two methods. 1) Shewhart method: electromyographic signal greater than "baseline" values by more than three standard deviations, and 2) Mean Electromyographic Difference method: mean electromyographic signal in a time window 25 to 150 milliseconds after the force perturbation greater than that in a 25- to 150-millisecond window before the perturbation.

Results. Lower preload efforts were associated with more muscle responses (overall mean response detection rate = 33% at low preload and 25% at high preload). Using the Shewhart method, there were significant differences by effort ($P < 0.05$) for all abdominal muscles and for all left dorsal muscles except multifidus. Using the Mean Electromyographic Difference method, there were significant differences by effort ($P < 0.05$) for the same dorsal muscles, but only for one of the abdominal muscles.

Conclusions. Findings are consistent with the hypothesis that the spine can be stabilized by the stiffness of activated muscles, obviating the need for active muscle responses to perturbations. [Key words: spine stability, muscle stiffness, perturbation, human subjects, electromyography] **Spine 2000;1957–1964**

The vertebrae of the lumbar spine are like a series of inverted pendulums, producing a ligamentous spine that is inherently unstable. Stability in this context is defined as the ability of a system to return to its equilibrium position after a small perturbation. The spine must be stabilized by the stiffness of the muscles and motion segments to prevent buckling. Otherwise, a sudden excessive displacement could occur and result in tissue injury. It has been shown analytically that muscle stiffness, which increases with intensity of muscle activation, can prevent lumbar spine buckling that would occur otherwise in subjects under loaded conditions or when experiencing a perturbation.[2,5,7–11] Therefore, the patterns of human trunk muscle recruitment not only must provide static equilibrium and appropriate response to changes in loading and displacement perturbations, but also must provide sufficient stiffness to ensure stability of the vertebral column.[1,2,6,7,10,14,16,18,19,24,25,27,30] Coactivation of antagonistic muscles is a part of a strategy that can increase the muscular stiffness and hence stability, but at the cost of increased spinal loads.[11,12,20,21]

The restoration of equilibrium after a perturbation can be achieved by active adjustment of muscle tensions, but with inherent neuromuscular delays.[28,29] Alternatively, small perturbations might be accommodated without such active responses, provided there is sufficient muscular stiffness and damping in the trunk. The present study is concerned with the second mechanism. The biomechanics of such passive stabilization that requires no active central nervous system-mediated adjustment of the preset muscle activation or stiffness after a perturbation was demonstrated theoretically by Bergmark,[2] and has been further explored analytically.[5,7–11] This mechanism of trunk stabilization is difficult to study experimentally, and for practical and ethical reasons experimental investigations of stability in human subjects must focus on the strategies that are used to prevent spinal buckling or other instability events.

The present study was designed to investigate spinal stability by recording whether trunk muscles were recruited in response to a transient perturbation, with different magnitudes of preloading. Because muscle stiffness increases with activation, it was expected that the trunk would be stiffer under more heavily loaded conditions;

From the Departments of *Orthopaedics and Rehabilitation, †Physical Therapy, and ‡Medical Biostatistics, University of Vermont, Burlington, Vermont.
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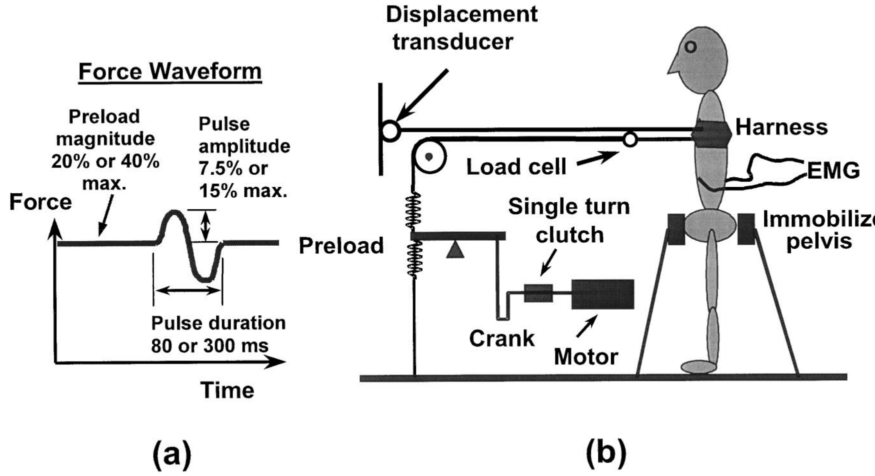

Figure 1. **a,** Parameters of the preload effort and of the force perturbation. **b,** Diagram of a subject standing in the apparatus, maintaining a steady-state preload effort to which a force perturbation can be added, here with an anterior acting force at 0° (*i.e.*, extension effort). EMG = electromyography.

therefore, an active response to a force perturbation would be less likely to occur for the same magnitude of perturbation. This study used a transient full-sine wave force perturbation to study the hypothesis that a stable trunk would not require muscular responses to the perturbation. A full sine-wave minimizes the momentum imparted to the subject, and there is no change in load magnitude after the perturbation. For these experiments, the perturbation was in the form of a single sine-wave force superimposed on a steady-state preload effort. This loading was chosen because the preload effort served to preactivate the trunk muscles, and the sinusoidal perturbation may not require an obligatory response, thus allowing study of the conditions under which no responses occurred.

The purpose of this investigation was to test the research hypothesis that an increase in preload effort results in decreased numbers of trunk muscle responses after a force perturbation, where a response is defined as increased muscle activation (measured by electromyographic (EMG) signal magnitude) above that before the perturbation. This was tested in 12 trunk muscles for five different loading directions because the muscles would be activated differently to oppose forces in different loading directions. Also investigated was whether the preload effort or perturbation characteristics altered the magnitude of muscle responses and whether the likelihood of a response depended on the perturbation amplitude and speed.

■ Methods

Thirteen healthy subjects (7 men, 6 women; age 21–33 years; height 1.77 ± 0.10 m; bodymass 74 ± 13 kg) were studied after they signed the informed consent form that had been approved by the Committee on Human Research. Each stood in an apparatus with the pelvis effectively immobilized by a support structure with pads pressing on the regions over the anterior superior iliac spines and the sacrum (Figure 1). A harness around the thorax was connected by a cable and pulley to a system for applying a horizontal steady-state preload, together with a superimposed force perturbation of variable (and controlled) amplitude and duration (Figure 1a). The cable was aligned approximately horizontal and level with the T12 vertebra. The pulley was attached to one of five anchorage points on a wall track surrounding the subject at angles of 0°, 45°, 90°, 135°, and 180° to the anterior direction. The mechanical system for generating the force perturbation (Figure 1b) consisted of a variable speed electric motor driving an eccentric-crank lever system by a single turn electromagnetic clutch. The clutch was activated by the experimenter pushing a button that caused the shaft connected to the clutch to execute a single turn. This produced a single full sine-wave displacement of the lever arm attached to the two springs in line with the cable connected to the harness around the subject. The amplitude of the sinusoidal displacement and the stiffness of the springs determined the amplitude of the force perturbation superimposed on the preload efforts.

Initially, the cable was anchored to the wall at 0° (extension effort), and subjects generated a timed ramped load test up to their maximum effort in 5 seconds with a further 5 seconds for gradual release of the load. Three trials were performed to help subjects learn how to achieve a maximum effort. The maximum achieved was used as the basis for determining the preload effort and perturbation amplitude in the perturbation experiments.

At each of the five test angles (the sequence of angles was randomly selected), subjects first generated a steady-state preload of nominal magnitudes 20% and 40% of the maximum force recorded in extension effort. An analog voltmeter with a target mark was used to help subjects maintain the desired steady-state preload effort by providing visual feedback of the force recorded by a load cell in the cable. The subjects were instructed to maintain a normal erect posture, symmetrically oriented with the apparatus during all tests. Then a single full sine-wave force perturbation pulse (nominal amplitude 7.5% or 15% of maximum effort, nominal duration 80 milliseconds or 300 milliseconds) was applied by the investigator without warning after a random time between 5 to 20 seconds after the subject reached the desired steady-state preload effort. The subjects had experienced the force perturbations in a practice session before the recorded trials. They were instructed "to try to maintain the needle position at the mark on the analog voltmeter." There were eight test conditions at each angle (two preload efforts, two pulse amplitudes, and two pulse durations), all of which were presented randomly. Three repeated trials were made in sequence of each test condition for a total of 24 trials

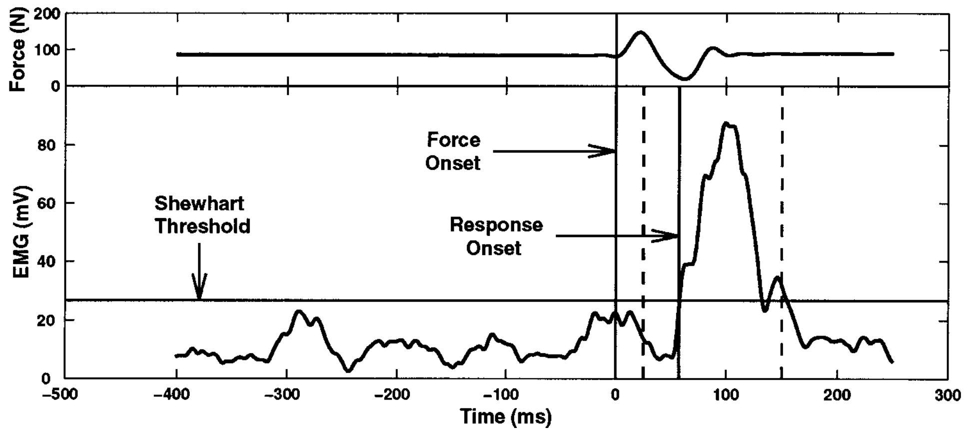

Figure 2. Example of the recorded force (top panel) and the EMG signal (lower panel, for left longissimus) in a perturbation experiment. The EMG signal is filtered and rectified. The Shewhart threshold (mean + 3 standard deviations of the signal in a preperturbation window) is indicated by a horizontal line. The dashed vertical lines indicate the 25- to 150-millisecond window for response detection. Here, a response with a latency of 57.6 milliseconds was detected.

at each of the five test angles and a grand total of 120 perturbation trials per subject. The total time of the testing session was approximately 4 hours, but the longest duration of a sustained effort was approximately 30 seconds. Subjects could rest between trials and while the load direction and perturbation parameters were altered.

To measure the displacement of the subject occurring at the harness, a linear displacement transducer with a resolution of 0.1 mm was mounted next to the wall anchor point and was connected to the harness by a light string.

The preload efforts averaged 119 ± 48 N when nominally 20% of maximum effort and 225 ± 91 N when nominally 40% of maximum effort. The force perturbation amplitude averaged 47 N when set to 7.5% of maximum effort and 87 N when set to 15% of maximum effort.

Electromyelography. Bipolar EMG electrodes recorded signals from six right and left pairs of muscles (rectus abdominis, internal and external obliques, longissimus, iliocostalis, and multifidus). Surface electrodes were used except for multifidus, whose activity was recorded by 50-μm nickel alloy indwelling wires connected to Motion Control Inc., Type 3030001 (Salt Lake City, UT USA) preamplifiers taped to the adjacent skin. Dorsal muscles and the oblique abdominal muscles were recorded by surface electrodes (Delsys Inc. Type DE-02.3, Boston, MA, USA). For rectus abdominis, neonatal Ag/AgCl monitoring electrodes with nominally 30-mm spacing were applied and connected to adjacently mounted Motion Control Inc. Type 3030001 amplifiers. The Delsys electrodes have 10-mm × 1-mm silver bar electrodes with 10-mm spacing; their single differential amplifiers have a gain of 1000, 20 Hz to 450 kHz bandwidth, 92 dB (typical) common mode rejection ratio, and 10^{12} ohms input impedance. The Motion Control single differential amplifiers have gain of 3000, 10 Hz to 24 kHz bandwidth, >100 dB common mode rejection ratio and 10^{11} ohms input impedance. A ground electrode was placed over the anterior aspect of the tibia. Electromyelographic and load cell signals were recorded digitally at 2048 or 1024 Hz.

Electrodes were placed as follows: rectus abdominis—30 mm lateral to the midline at the level of the umbilicus, aligned vertically; external oblique—halfway between the iliac crest and the 12th rib along the midaxillary line, aligned at an 80° angle to the horizontal; internal oblique 20 mm medial and superior to the anterior superior iliac spine, aligned at 45°; longissimus—30 mm lateral to the midpoint of the spinous process of L3, aligned vertically; iliocostalis—60 mm lateral to the midpoint of the spinous process of L3, aligned vertically. For multifidus, fine wire electrodes[22] were fabricated from 50-μm gauge nylon-insulated twisted wire pairs with a 2-mm long uninsulated section set back 5 mm on one wire and 9 mm on the other. A posteroanterior radiograph with subjects in the prone position was used to locate the spinous process of L4 with the help of lead balls as reference points set into a plastic sheet that was placed over the lumbar region. The electrode insertion point was 5 mm lateral to the edge of the L4 spinous process and 10 mm cranial to the inferior aspect of the L4 spinous process. The wires were threaded through a needle that was inserted at a 90° angle to the skin surface until the periosteum of the L4 lamina was contacted, then the needle cannula was withdrawn, leaving the two wires in place. This procedure was intended to place the bare sections of the wires in the individual muscle slip of multifidus that inserts at L3.[13,23]

Signal Processing. First, EMG signals were bandpass filtered by a 10- to 100-Hz Chebyshev type II filter with no lag and rectified. The filtration was intended to reduce any electrocardiograph (EKG) or motion artifact and high frequency noise contamination of the signals. A 25-millisecond moving average of the rectified EMG signal then was calculated. The onset of the force perturbation was first identified from the load cell recording by detecting the time at which there was a significant increase in the force-time slope (Figure 2). A 25- to 150-millisecond time window after the force perturbation was examined using two different methods to detect any short and medium latency muscle responses to the perturbation.

Suspect data resulting from technical problems such as loose electrodes or EKG artifacts were excluded from further statistical analyses. These records were eliminated by a dual process of visual inspection and identification of outliers. A multiple-turning point algorithm was used to identify probable EKG and motion artifacts, and these were identified in traces of EMG *versus* time by custom software. Each detected onset then was visually inspected, and the first onset not identified as an artifact was accepted. For outliers caused by electrical transients associated with the electromagnetic clutch, recordings with a Mean EMG Difference greater than five times the mean EMG in a preperturbation window were eliminated. This criterion was established by inspection of the response magnitude distribution. For the 13 subjects, with five loading directions, two preload magnitudes, two perturbation amplitudes, and two perturbation durations, with three repetitions of each test condition, there were a total of 1560 possible trials. Of these, 170 were not completed for technical reasons. Of the resulting 16,680 muscle recordings (12 muscles per trial), 1772 were eliminated because of artifacts in recordings. Of these, 576 were from left multifidus muscles and 572 were from right multifidus muscles.

Response Detection by Shewhart Method.[15] A muscle response to force perturbation was considered to occur if the processed EMG signal exceeded a threshold of three standard deviations (SDs) above a baseline mean. The "false-positive" rate of this method was estimated by applying the same detection criterion to a 1000- to 875-millisecond window preceding the perturbation. The baseline mean was computed using the median mean EMG signal magnitude in five sequential 100-millisecond windows in the 500 milliseconds preceding the force perturbation. The standard deviation of the 100-millisecond window corresponding to the median mean was used to compute the three standard deviation threshold. Using the median minimized the potential effect of outliers resulting from EMG contamination by EKG or other artifacts. When a response was detected, its latency was measured as the time from the start of the force perturbation to onset of the EMG response.

Response Detection by Mean EMG Difference Method. The difference between the mean EMG signal in a 25- to 150-millisecond window after the perturbation and a 275- to 150-millisecond window before the perturbation was computed for each EMG signal. A response was defined as an increase in the mean EMG signal after perturbation (*i.e.*, a positive mean EMG difference). The observed number of responses was examined relative to the 50% rate expected by chance if there were no true increase in muscle activity. This detection method was not influenced by the differences in the standard deviation of the baseline EMG signal between experimental conditions that potentially could produce detection bias with the Shewhart method. Latencies could not be calculated by this method.

Response Magnitude. The magnitude of the mean EMG difference was used to test for differences in response magnitude across experimental conditions. These analyses were performed separately for trials in which a response was detected by each of the two previously defined criteria.

Statistical Analyses. The response frequency (measured by the Shewhart or Mean EMG Difference method) was measured (for each muscle under each testing condition) as the average of the dichotomous response scores from the three replicate trials. Repeated measures analyses of variance were used to determine the significance associated with differences in average response frequency as a function of steady-state preload, angle, perturbation amplitude, perturbation speed, and their potential interactions. Separate analyses were performed for each muscle. Because the primary dependent measure was an estimated muscle response frequency from three repeated trials, these data were subjected to an arcsine square root transformation before analysis to satisfy the assumptions of the analyses of variance.[3]

Magnitudes of responses, measured by values of the Mean EMG Difference, were also compared between experimental conditions using repeated measures analyses of variance. These analyses were performed separately for trials in which a response was detected using each of the two detection criteria. All statistical analyses were performed using SAS statistical software (SAS Institute, Cary NC). Statistical significance was determined based on $\alpha = 0.05$.

Results

When averaged for all muscles, only 28.9% of recordings showed a detectable muscle response by the Shewhart method, with individual muscles response frequencies ranging from 16% (right multifidus muscle) to 42% (right and left rectus abdominis muscles). Low preload efforts were associated with higher muscle response frequencies (Figures 3 and 4). When averaged for all test conditions, the average number of trials producing a detected response was observed to be greater at low preload than at high preload for all 12 muscles (overall response frequency: 33% *vs.* 25% by Shewhart method, 0.77 *vs.* 0.73 for Mean EMG Difference method). Using the Shewhart method, there were significantly more detected responses at low preload effort ($P < 0.05$) for right and left internal and external obliques, right and left rectus abdominis, left longissimus, and left iliocostalis. These eight muscles produced estimated response frequencies averaging 37% for the low preload condition compared with 27% for high preload. Using the Mean EMG Difference method, there were significantly more responses at low preload effort ($P < 0.05$) for the same dorsal muscles, but for only one of the abdominal muscles (right external oblique).

Greater perturbation amplitude produced significantly more detected responses in all but one muscle (right multifidus). Responses were detected by the Shewhart method in 20% of trials (averaged across muscles) with low perturbation amplitude compared with 38% of trials with high amplitude perturbation.

The number of "false-positive" responses detected by the Shewhart method in the preperturbation window averaged 5.66% across all muscles and testing conditions. This is much lower than the number of responses detected in the postperturbation window (28.9%).

The magnitudes of responses as measured by the mean EMG difference were observed to be greater by 24% (averaged for all muscles) at high preloads for responses detected by the Shewhart method and by 20% for responses detected by the mean EMG difference method. These differences only reached statistical significance, however, for the right rectus abdominis. The magnitudes of the responses were significantly greater (by an overall average 1.3 times) for larger perturbations at both preload efforts. Also, there were significantly larger responses at force angles for which the muscles were opposite to the applied force direction. The largest magnitude responses occurred when the force acted at 0°, 45°, or at 90° for dorsal muscles (longissimus, iliocostalis, and multifidus), 135° or 180° for the oblique abdominal muscles, and at 180° for the rectus abdominis. The average response magnitude was between 1.5 times and 7.6 times greater at these angles than at the

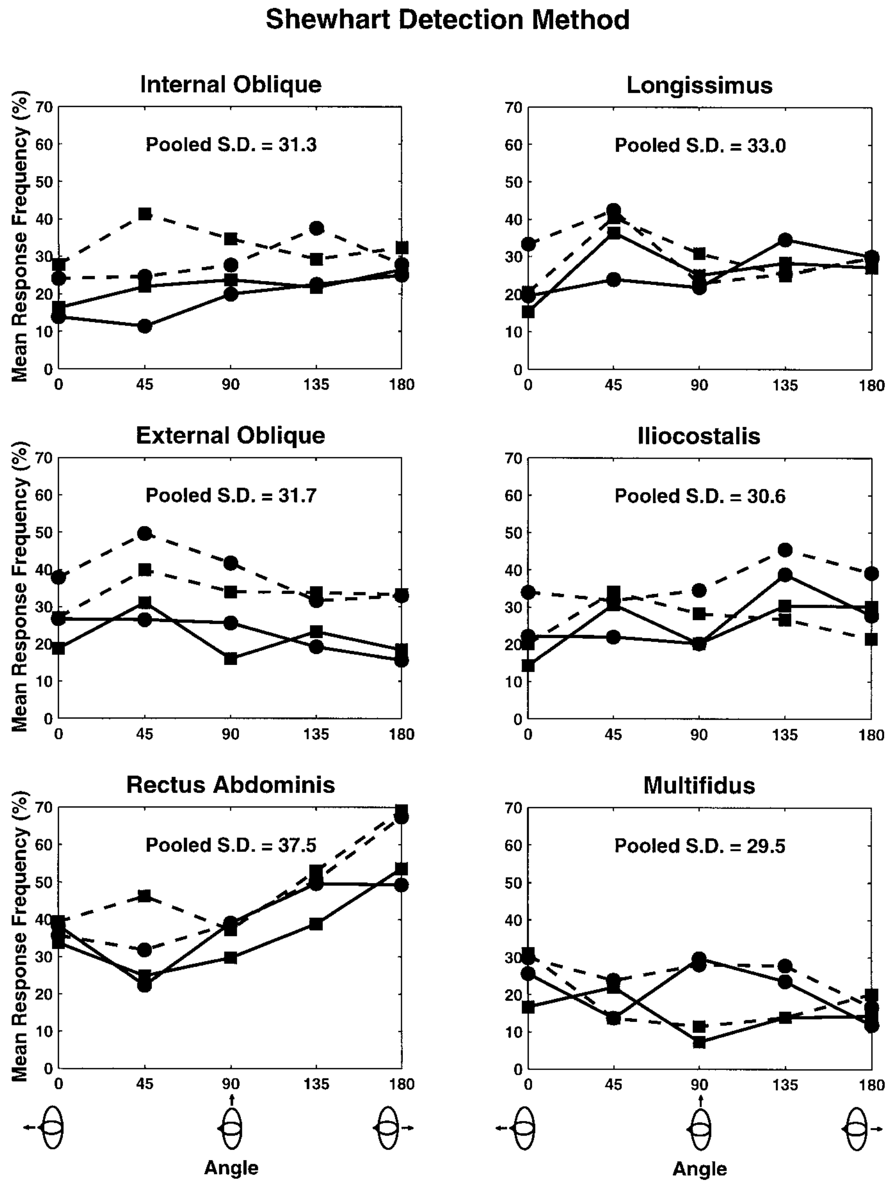

Figure 3. Detected responses to a force perturbation (Shewhart method) of trunk flexor and extensor muscles (% of total number of trials) as a function of angle of the effort. Left and right muscles are plotted separately for low (20%) and high (40%) preload efforts: ● = left muscles; ■ = right muscles, dashed line = low (20%) preload effort, solid line = high (40%) preload effort. The figurine, viewed from above, and the arrow indicate the direction of the preload and perturbation forces.

angles where the lowest magnitude responses were observed.

There was no evidence that the speed of stimulus had any influence on response magnitude. There were also no significant perturbation speed effects on the number of detected muscle responses except that there were more responses at higher speeds for right rectus abdominis (48% and 37% response frequency at high and low speed respectively) and right external oblique (32% and 23% for high and low speed, respectively).

The latency of detected responses after the onset of the force perturbation is summarized in the histogram in Figure 5, showing that most detected responses occurred within 100 milliseconds of the onset of the force perturbation. There were no significant differences in latency between experimental conditions.

The force perturbations produced an average displacement of 2.91 mm for low amplitude perturbations and 5.9 mm for high amplitude perturbations at the low (20% effort) preload. For 40% effort preloads, the perturbations caused lesser displacements (averages of 1.9 mm and 4.4 mm for low amplitude and high amplitude perturbations). The high speed perturbations caused approximately half as much trunk displacement as the low speed perturbations.

■ Discussion

The results of this study show that the amount of preactivation of the trunk musculature (determined by the preload effort) influenced the likelihood of muscular response to a force perturbation. As hypothesized, the likelihood of a muscular response was less when the preload effort was higher. A change in muscular activation was detected in a minority of recordings when the preloaded trunk was perturbed by a force pulse. This is consistent with the hypothesis that the spine may be stabilized by the stiffness of preactivated muscles, avoiding the need for an active response to perturbations. For

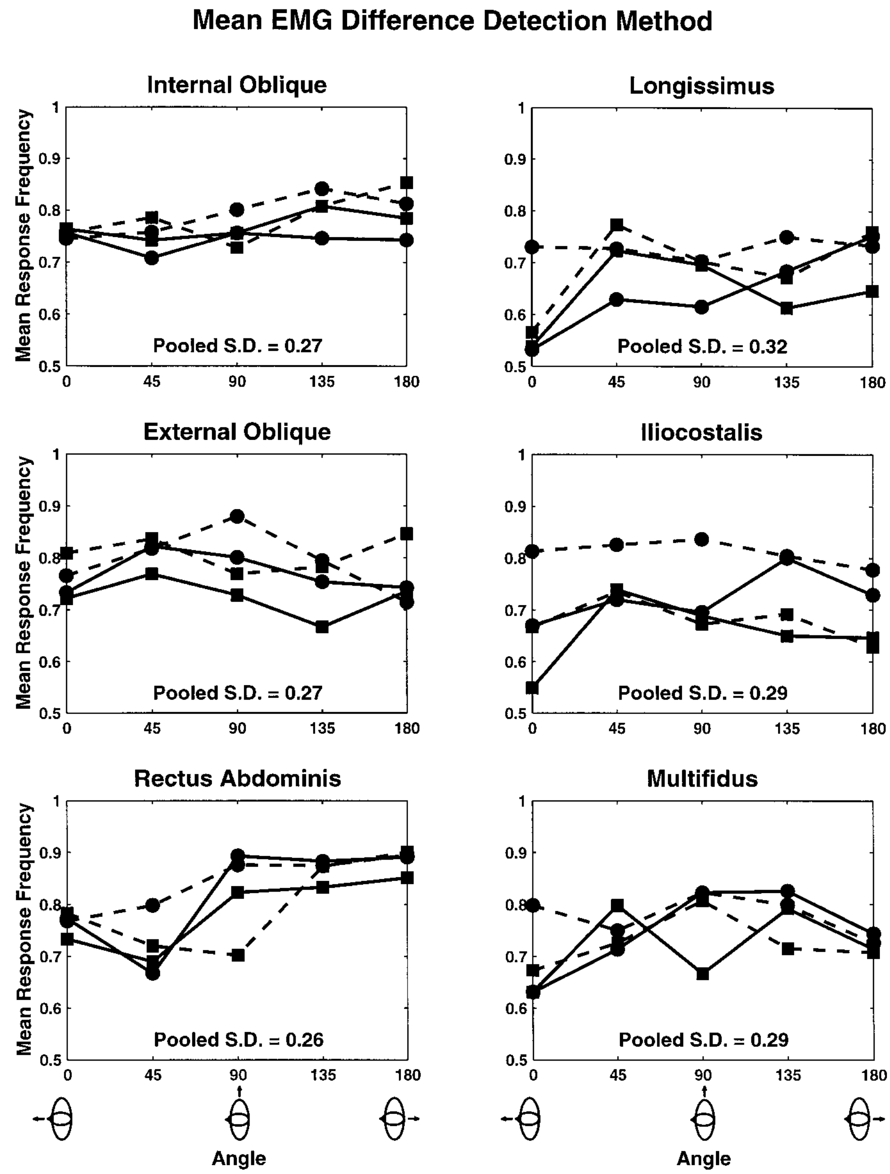

Figure 4. Detected responses to a force perturbation (proportion of total number of trials) by the Mean EMG Difference Method of trunk flexor and extensor muscles as a function of angle of the effort. Left and right muscles are plotted separately for low (20%) and high (40%) preload efforts: ● = left muscles, ■ = right muscles. Dashed line = low (20%) preload effort; solid line = high (40%) preload effort. The figurine, viewed from above, and the arrow indicate the direction of the preload and perturbation forces. Note: the expected frequency of response detection by chance is 0.5 if no responses occurred.

some muscles at certain angles (*e.g.*, the left iliocostalis in extension efforts), the number of detected responses at low preload was approximately twice that at high preload. In most other cases, however, the difference by preload was less marked.

Reliable response detection was difficult because detected responses were small relative to the fluctuation of the EMG signal from preactivated muscles. To compromise between false-positive and false-negative detections, the Shewhart threshold for detection was set as the mean plus three standard deviations of the steady-state EMG signal. The median of the heights of detected EMG responses was only 3.96 standard deviations above the preperturbation mean, and 80% had peaks less than six standard deviations. The number of responses detected by the Mean EMG Difference method would be 0.5 (by chance) if there were no muscular responses; therefore, the proportions of responses detected by these two methods are not directly comparable. Despite concern that the Shewhart detection algorithm might be influenced by the level of preactivation of the muscle and would thereby be biased by the preload effort, the consistency of response frequencies and preload effects between the two detection methods (Shewhart and Mean EMG Difference) indicates that this was not a problem.

Increased amplitude of force perturbation increased the number of responses, as expected, but the number of responses was relatively insensitive to both the speed of perturbation and the angle of the force. The observed muscle response latency in the range 25 to 150 milliseconds is indicative of a monosynaptic reflex (probably stretch reflex) or "medium" latency response. The responses that occurred at the lower end of this latency range were not unexpected, because the short conduction distances for reflexes in paraspinal muscles allow rapid responses to direct stimulation of paraspinal structures.[17,26] The observed muscle responses that occurred in some cases are consistent with triggering by muscle

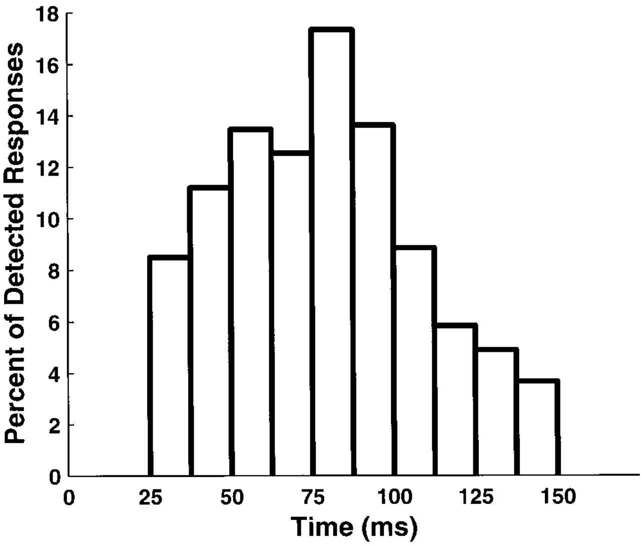

Figure 5. Histograms of detected muscle responses at different latencies after the onset of force perturbation in the 25- to 150-millisecond detection window.

spindles, Golgi tendon organ, joint mechanoreceptors, or cutaneous sensory afferents, or a combination of these, based on the magnitude of forces applied through the harness (on the order of 50 N) and the thorax displacements induced (on the order of 2 to 6 mm). Because the exact triggering pathways are unknown, it is difficult to speculate on why responses were inhibited with higher preload. The preactivation of the muscles and increased activity of the gamma system could increase the sensitivity of the muscle spindles, thus increasing the likelihood of a response. This speculation goes against the hypothesis of the present study and is contrary to what was found. The increased stiffness at higher preload reduced the displacements occurring, however, perhaps lessening the role of spindles in triggering a response. These neurophysiologic mechanisms would predict a reciprocal response of muscles based on whether they were stretched or shortened by the perturbation. Because there was no clear reciprocal pattern evident, the probability of response could be determined by a threshold phenomenon that changes depending on the state of the entire system, including local spinal inhibition and supraspinal (perhaps back lifting mediated) inhibition influenced by central set. The absence of large differences in number of detected responses by force direction (Figures 4 and 5) is suggestive of a common central drive of responses to these perturbations, such that muscle recruitment is directionally nonspecific under these conditions.

Previous studies of trunk muscle responses to force perturbations have used a sudden increase or decrease in load by a dropped weight or a sudden release.[4,6,19,24,30] These produce both step and impulse loading and require an obligatory muscle response of trunk muscle responses. The paradigm in the present study used a transient perturbation to study the hypothesis that a stable trunk would not require muscular responses. In studies in which a response was obligatory, Lavender and Marras[19,24] found that if their subjects were expecting the sudden loading from a dropped weight, there were anticipatory activations of trunk muscles, especially of abdominal muscles. They noted that the resulting muscular forces could cause large forces on the spine, but presumably they also increase the muscle stiffness and hence increase trunk stability. Carlson et al[4] and Cresswell et al[6] used a similar paradigm with weights suddenly added or removed from a harness over the shoulders. They observed that the abdominal muscles were usually the first to activate. Wilder et al[30] reported the latency of responses to sudden perturbation (from a dropped weight) in the range of 65–385 milliseconds, with evidence that latency was increased with muscle fatigue. In subjects with low back pain, latency was longer than in others, but the latency of the trunk muscle EMG responses was reduced after an exercise program.

The results of these experiments provide insights into how the central nervous system controls muscle preactivation in expectation of a perturbation and when responses to external force perturbations are triggered. The likelihood of muscle responses was greater at low preload effort and for greater perturbation amplitude, as expected. The angle of the force direction and the perturbation speed had no strong effect on the responses. The fact that most of these perturbations did not elicit a detectable muscle response supports the contention that the spine is stabilized by the stiffness of preactivated muscles, obviating the need for active (central nervous system-mediated) responses.

■ Key Points

- Trunk stability was defined as its ability to return to an equilibrium position after a perturbation.
- Muscles can actively assist in restoring the equilibrium position by alteration of their activation, and the stiffness of activated muscles stabilizes the loaded spine.
- It was found that the likelihood of muscle active response to a perturbation would reduce with preactivation of the muscles, because muscles then would be stiffer.

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Address reprints requests to

Ian A.F. Stokes
Department of Orthopaedics and Rehabilitation
University of Vermont
Burlington, VT 05405