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# Trunk stiffness increases with steady-state effort

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**Abstract**

Trunk stiffness was measured in healthy human subjects as a function of steady-state preload efforts in different horizontal loading directions. Since muscle stiffness increases with increased muscle activation associated with increasing effort, it is believed that coactivation of muscles helps to stiffen and stabilize the trunk. This paper tested whether increased steady-state preload effort increases trunk stiffness. Fourteen young healthy subjects each stood in an apparatus with the pelvis immobilized. They were loaded horizontally at directions of 0, 45, 90, 135 and 180° to the forward direction via a thoracic harness. Subjects first equilibrated with a steady-state load of 20 or 40% of their maximum extension effort. Then a sine-wave force perturbation of nominal amplitude of 7.5 or 15% of maximum effort and nominal period of 250 ms was applied. Both the applied force and subsequent motion were recorded. Effective trunk mass and trunk-driving point stiffness were estimated by fitting the experimental data to a second-order differential equation of the trunk dynamic behavior. The mean effective trunk mass was 14.1 kg (s.d. = 4.7). The trunk-driving point stiffness increased on average 36.8% (from 14.5 to 19.8 N/mm) with an increase in the nominal steady-state preload effort from 20 to 40% ($F_{1,13} = 204.96$, $p < 0.001$). There was a smaller, but significant variation in trunk stiffness with loading direction. The measured increase in trunk stiffness probably results from increased muscle stiffness with increased muscle activation at higher steady-state efforts. © 2001 Elsevier Science Ltd. All rights reserved.

*Keywords:* Stiffness; Muscles; Lumbar spine; Biomechanics; Stability

## 1. Introduction

Trunk stiffness is important not only in the elastic behavior of the upper body, but also in contributing to trunk stability. Stability is defined as the ability of a system to return to equilibrium after a small perturbation. The ligamentous spine is unstable at compressive loads of only 88 N (Crisco et al., 1992) while in vivo the compressive force acting on the spine can exceed 2600 N (Nachemson, 1966). Among the forces that return the trunk to an equilibrium position after a small displacement are those generated by elastic forces due to spine stiffness, augmented by activated muscle stiffness. It has been shown analytically that spine stiffness alone is insufficient, and that activated muscle stiffness is necessary for trunk stability (Bergmark, 1989; Cholewicki et al., 1997; Crisco and Panjabi, 1991; Gardner-Morse et al., 1995; Gardner-Morse and Stokes, 1998). Muscle stiffness increases with muscle activation as a result of the increased number of activated cross-bridges (Crisco and Panjabi, 1991; Ma and Zahalak, 1991). Therefore, theoretical considerations suggest that muscle stiffness contributes to trunk stability and that this mechanism could be controlled through modulation of muscle activation.

Theoretically, spine stiffness can be increased while maintaining equilibrium by increasing the coactivation of antagonistic muscles (Cholewicki et al., 1997; Gardner-Morse and Stokes, 1998) as has been shown experimentally in other joints (Baratta et al., 1988; Hunter and Kearney, 1982; Zhang et al., 1998). Disadvantages of coactivation are increases in tissue loading and metabolic energy consumption. While additional stiffness helps to stabilize the trunk, coactivation also paradoxically increases the compressive load that tends to destabilize the spine (Gardner-Morse and Stokes, 1998; Granata and Marras, 1995; Thelen et al., 1995). These qualitative concepts require quantitative experimental evidence to support them.

Trunk stiffness can be measured from the dynamic response of the trunk to a force or displacement

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perturbation. The dynamic behavior depends on the inertial, damping, and stiffness properties of the trunk and these variables can be extracted from the measured response to perturbations. The dynamic behavior of other joints have been studied either by measuring the displacements produced at the joint under controlled force inputs, or by measuring the forces produced at the joint under controlled displacement inputs (Kearney and Hunter, 1990). The inputs may be sinusoidal, pseudo-random, impulse, or a step function. For small displacements of a joint from a set position, a linear second-order differential equation adequately represents the dynamic behavior under a wide range of muscle activation (Kearney and Hunter, 1990).

Previous investigations of the trunk have reported changes in measured trunk stiffness with breath holding (increased intra-abdominal pressure) and belt wearing (Cholewicki et al., 1998; McGill et al., 1994), and decrease in the amount of trunk motion resulting from a perturbation with a increased flexion preload (Krajcarski et al., 1999). These reports suggest that the degree of muscle activation has an effect on trunk stiffness, but the exact relationship between the loading state, muscle activation and trunk stiffness is poorly understood. The purpose of this study was to measure the trunk driving point stiffness as a function of steady-state preload effort in different loading directions in the horizontal plane. These experiments were designed to test the hypothesis that increasing steady-state preload efforts increases trunk stiffness at all loading directions in the horizontal plane.

## 2. Methods

Fourteen young healthy human subjects were tested after they had signed a consent form approved by the institutional human research committee. There were 8 males and 6 females. The mean age was 25.7 years (range 20.7–33.2, s.d. = 3.9); mean height was 1.76 m (range 1.59–1.90, s.d. = 0.10); and mean body mass was 73.8 kg (range 52.6–102.1, s.d. = 12.5).

Each subject stood in an apparatus that effectively immobilized the pelvis and they wore a harness around the thorax, attached via 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 (Fig. 1a). The supporting apparatus consisted of a braced metal frame with three adjustable pads pressing on the right and left ASIS and the sacral region. The harness was a custom modified nylon wind-surfing belt with a steel cable in flexible plastic tubes around the outside of the belt and a pair of nylon shoulder straps for controlling the height of the harness on the subjects. The cable was aligned approximately with the T12 level and the pulley was attached to one of five anchorage points on a wall track surrounding the subject at directions of 0, 45, 90, 135 and 180° from the anterior direction along the subject's right side (Fig. 1b).

The mechanical system for generating the force perturbation (Fig. 1a) consisted of a variable speed electric motor driving an adjustable eccentric-crank lever system via a single turn electromagnetic clutch. The clutch was activated by the experimenter pushing a button, and this 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 spring in-line with the cable connected to the harness around the subject. The amplitude of the sinusoidal displacement was adjusted for each subject and together with the measured stiffness of the spring determined the amplitude of the force perturbation.

Initially, the cable was anchored to the wall at 0° (extension effort) and subjects generated a timed ramped load test up to their maximum isometric effort in 5 s with

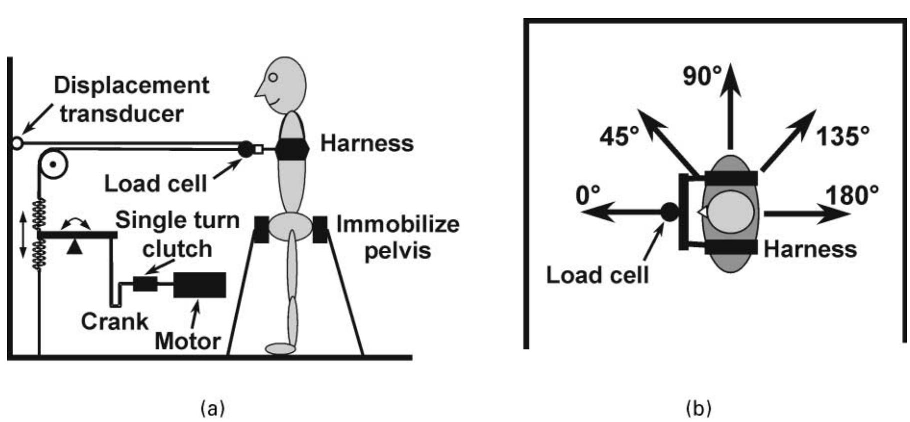


Fig. 1. (a) Diagram of a subject standing in the apparatus, maintaining a steady-state preload effort through a nylon harness to which a force perturbation was added. The subject is supported in a stiff frame with pads pressing on the pelvis. Here the force acts anteriorly at 0° (i.e. extension effort). (b) The force loading direction was at 0, 45, 90, 135 or 180° to the anterior direction around the subject's right side.

a further 5 s for gradual release of the load. This was repeated three times to give subjects the opportunity to learn how to achieve a maximum effort. A load cell in the cable measured the force generated. The maximum effort achieved was used as the basis for determining the steady-state effort and perturbation amplitude in the perturbation experiments.

At each of the five loading directions (the sequence of directions was randomly selected), subjects first equilibrated with a steady-state effort. Subjects were instructed to maintain a normal erect posture symmetrically oriented with the apparatus during all tests. The signal from the load cell was displayed to them on an analog voltmeter with a target mark. This helped subjects maintain the desired steady-state preload effort by providing visual feedback of the force. They were instructed "to try to maintain the needle position at the mark on the analog voltmeter".

The nominal magnitudes for the steady-state efforts were 20 and 40% of the maximum force recorded in the maximum extension effort tests. The 20 and 40% of maximum effort were selected to limit the number of force levels to just two in order to keep the experiment to a reasonable time and to minimize subject fatigue. The lower steady-state effort had to be greater than the perturbation amplitude to keep the cable in tension. After subjects equilibrated to the preload a single full sine-wave force perturbation (nominal amplitude 7.5 or 15% of maximum effort, nominal period 250 ms) (Fig. 2a) was applied by the investigator after a random time between 5 and 20 s. The 15% perturbation amplitude was considered to be consistent with subject safety (Carlson et al., 1981; Lavender et al., 1993; Marras et al., 1987). The 7.5% perturbation amplitude was used to determine whether measured stiffness varied with perturbation amplitude. The subjects had experienced the force perturbations in a practice session prior to the recorded trials. While a perturbation was expected by subjects, there were no cues for the exact time when it would occur. At each loading direction there were four test conditions (two steady-state efforts, two perturbation amplitudes) all of which were randomly presented. Three repeated trials were made in sequence of each test condition for a total of 12 trials at each of the five loading directions and a grand total of 60 perturbation trials per subject. The total time of the testing session was about four hours. The longest duration of a sustained effort was about 30 s for the ramped effort trials, while the typical submaximum effort perturbation trial was about 10 s. Subjects could rest between trials and while the load direction and perturbation parameters were altered.

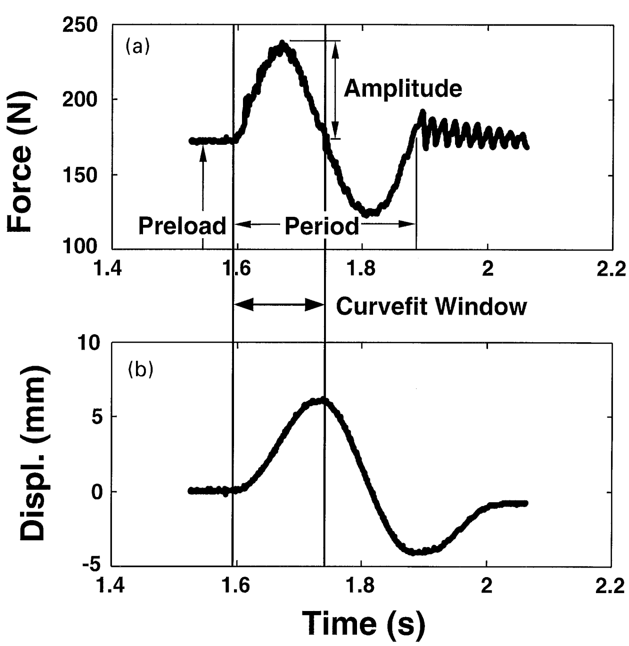


Fig. 2. (a) Typical input force trace and (b) displacement response. The time window used for the curvefits (first-half of the force sine wave) is also indicated.

The steady-state efforts averaged 122 N (s.d. = 44) when nominally 20% of maximum effort, and 234 N (s.d. = 83) when nominally 40% of maximum effort. The force perturbation amplitude averaged 28 N (s.d. = 12) when set to 7.5% of maximum effort, and 62 N (s.d. = 24) when set to 15% of maximum effort. The period of the force perturbation averaged 253 ms (s.d. = 23).

A spring-loaded displacement transducer attached to the load cell measured the resulting displacement with a resolution of 0.1 mm. The data were recorded at 1024 or 2048 Hz and low-pass filtered at 256 Hz with a no lag fourth-order Butterworth digital filter to reduce high-frequency electrical noise and high-frequency artifacts due to some ringing in the high stiffness components of the apparatus. Typical force and displacement time histories are shown in Fig. 2.

The displacement responses to the impulse load were analyzed using a second order differential equation of the trunk dynamic behavior. The differential equation describing the dynamic equilibrium of the trunk is

$$m\frac{\mathrm{d}^2\delta}{\mathrm{d}t^2} + \psi\frac{\mathrm{d}\delta}{\mathrm{d}t} + k\delta = A\sin(\omega t), \quad (1)$$

where $\delta$ is the displacement of the point of load application, $t$ is time, $m$ is the effective mass of the trunk, $\psi$ is a damping coefficient, $k$ is the effective trunk stiffness and $A$ is the amplitude of the force perturbation, $\omega$ is the circular frequency of the perturbation.

The value of the trunk driving-point stiffness $k$ and effective mass $m$ were evaluated by two nonlinear curvefit procedures. First, the first half period of the input force impulse was fit to a sine-wave function to determine the amplitude $A$, the circular frequency $\omega$, and time of force onset $t_0$ (Fig. 3a). Second, the trunk

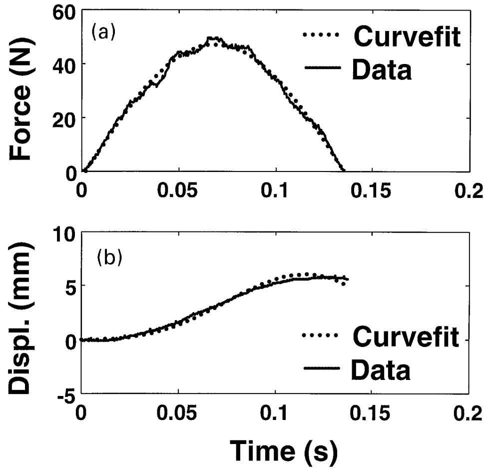


Fig. 3. Typical curvefits of (a) force data fitted to the first-half period of a sine wave, and (b) displacement data fitted to the second-order differential equation (see Eq. (1)).

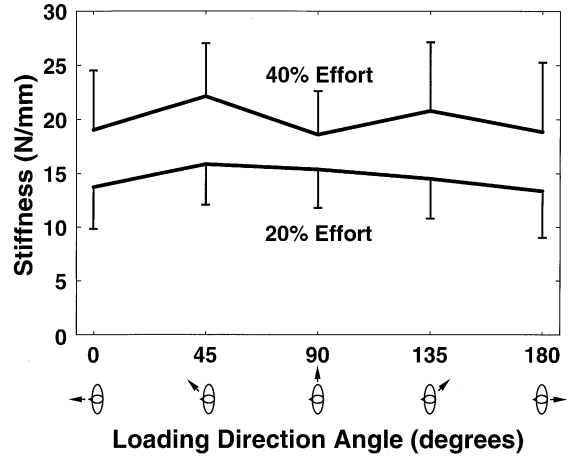


Fig. 4. Mean (and s.d.) of the experimentally determined trunk driving point stiffness for horizontal steady-state efforts at loading directions from 0 to 180° as indicated by the figurines. The trunk stiffness increased an average 36.8% with an increase in the nominal steady-state effort from 20 to 40% ($F_{1,13} = 204.96$, $p < 0.001$). The stiffness at the higher steady-state effort was significantly higher at all loading directions ($p < 0.05$).

driving-point stiffness $k$ and effective mass $m$ were found by fitting the trunk displacement response over the same time as the first curvefit using Eq. (1) (Fig. 3b). Since the head and arms are flexibly attached to the trunk and the trunk is not rigid, the effective mass is not the same as the total mass of the upper body. Both damping and gravity effects were assumed to be negligible compared to the other forces.

Each pair of force and displacement data was examined visually prior to curvefitting. Of the 840 possible trials (14 subjects, five loading directions, two perturbation amplitudes, two steady-state efforts, three repetitions) 49 were missed because of technical difficulties, 63 were excluded because the visual check revealed a problem in the recorded data (one channel missing, etc.), and two trials were eliminated because the curvefits explained less than 50% of the variance in the data. Thus, 726 valid trials (86% of the possible total) were available for analysis. Eight of 14 subjects had complete data. On average the curvefits explained over 91% of the variation in the data.

Three-factor repeated measures analysis of variance was used to test for differences in the two outcome measures (trunk driving point stiffness and effective mass) across experimental conditions using SAS (SAS Institute, Cary, NC). The independent variables were the steady-state efforts (two levels), loading directions (five angles) and perturbation sine wave amplitudes (two levels). Multiple comparisons of the outcomes (driving point stiffness or effective mass) between loading directions and of driving point stiffness between steady-state efforts at each loading direction were performed using Bonferroni $t$-tests. The significance level for all statistical analyses was set at $p = 0.05$.

## 3. Results

The measured trunk stiffness varied with steady-state effort ($F_{1,13} = 204.96$, $p < 0.001$) and with loading direction ($F_{4,47} = 2.81$, $p = 0.036$) (Fig. 4). There was an average 36.8% increase (from 14.5 to 19.8 N/mm) in the values of trunk stiffness with increased steady-state effort (pooled across loading directions and perturbation amplitudes). The stiffness at the 45° loading direction was significantly higher than the stiffness at 0, 90 and 180°, but not significantly different from the stiffness at 135°. At higher force perturbation amplitudes the measured stiffness was slightly lower (16.2 N/mm, s.d. = 4.9 vs. 18.0 N/mm, s.d. = 5.9) ($F_{1,13} = 28.45$, $p < 0.001$). The steady-state effort and loading direction interaction was also significant ($F_{4,45} = 3.52$, $p = 0.014$). However, the 40% steady-state effort stiffness was significantly higher than the 20% steady-state effort stiffness at each of the loading directions. The perturbation amplitude and loading direction interaction was also significant ($F_{4,45} = 2.91$, $p = 0.032$) because the perturbation amplitude effect was not evident at the 90° loading direction.

The mean effective trunk mass was 14.1 kg (s.d. = 4.7). The effective mass varied with steady-state preload ($F_{1,13} = 50.70$, $p < 0.001$) and loading direction ($F_{4,47} = 11.72$, $p < 0.001$) (Table 1). The effective mass increased by 26.6% with steady-state effort (from 12.4 kg, s.d. = 4.0 to 15.7 kg, s.d. = 4.8). The effective masses at the 45 and 90° loading directions were significantly higher than the effective mass at 0°. The effective masses at the 45, 90 and 135° loading directions

Table 1
Measured effective trunk mass (kg) (standard deviations in parentheses) by steady-state effort and loading directions. Values are averaged across subjects and perturbation amplitudes

| Steady-state effort (% maximum) | Loading direction (degrees) | | | | |
|---|---|---|---|---|---|
| | 0 | 45 | 90 | 135 | 180 |
| 20 | 11.6 (3.2) | 13.2 (2.9) | 14.8 (4.9) | 11.9 (3.4) | 10.4 (3.4) |
| 40 | 14.5 (4.1) | 16.9 (4.6) | 17.2 (3.9) | 15.8 (5.1) | 13.9 (5.2) |

were significantly higher than the effective mass at 180°. The effective mass also varied with the amplitude of force perturbation ($F_{1,13} = 70.20$, $p < 0.001$). Effective mass decreased by 18.7% at the higher amplitude of force perturbation (from 15.5 kg, s.d. = 5.2 to 12.6 kg, s.d. = 3.5).

Peak trunk displacements averaged 3.8 mm (s.d. = 1.5) for low-amplitude perturbations and 8.1 mm (s.d. = 2.1) for high-amplitude perturbations at 20% effort and averaged 2.4 mm (s.d. = 1.4) for low amplitude and 5.7 mm (s.d. = 2.1) for high-amplitude perturbations at 40% effort.

## 4. Discussion

Since muscle stiffness depends on muscle activation, we hypothesized that increasing steady-state trunk efforts would increase trunk stiffness. This experiment demonstrates that increasing muscle activation by increasing the steady-state effort from a nominal 20% to 40% of the maximum extension effort produced a 36.8% increase in measured trunk stiffness (Fig. 4). While the trunk stiffness varies with the loading direction, the trunk stiffness increased significantly with steady-state effort at all loading directions.

These findings are consistent with those reported for the ankle (Hunter and Kearney, 1982; Weiss et al., 1988) and knee (Zhang et al., 1998) where the joint stiffnesses also increased with increasing effort and decreased with increasing perturbation amplitude (Kearney and Hunter, 1982). Cholewicki et al. (1998) reported similar percentage increases in stiffness of the trunk with abdominal belt wearing and with voluntary increases in intra-abdominal pressure. The variation in trunk stiffness with horizontal loading direction is also consistent with the reported differences in driving point mechanical impedance and apparent mass of seated humans exposed to horizontal vibrations (Holmlund and Lundström, 1998; Mansfield and Lundström, 1999). These studies found variations in the resonances with loading directions between the forward–backward direction and 90° to the forward–backward direction. However, these studies of seated subjects can not be compared with our findings because the subjects were not performing steady-state preload efforts.

Experimentally, there were simplifications and approximations in the stiffness measurement method. The measured stiffness includes the flexibility of the chest soft tissue, the thoracic harness, and the test frame supporting the hips. Thus, the reported values probably underestimate the true values.

The representation of the trunk by a single rigid mass and spring (Eq. (1)) is a simplification. The effective mass measured in these experiments was less than the total upper body mass and varied significantly with the steady-state preload effort, loading direction and the force perturbation amplitude. This was probably caused by changes in the non rigid mass-coupling of the upper body segments. Mansfield and Lundström (1999) found two peaks in the apparent mass magnitudes which varied with the vibration magnitude and loading direction. Thus, for a more complete representation of the trunk dynamics, it should be considered as several masses connected by flexible structures. The stiffness of these connecting structures apparently varies with the degree of muscle activation and with the type of perturbation.

The increased trunk stiffness with steady-state preload effort is probably not due to reflex muscular responses. Stokes et al. (2000) using the same protocol observed muscle responses in fewer than 30% of trials. The effects of any muscle responses was also minimized by the short period of the force perturbation (mean = 253 ms) and only data from the first-half of the perturbation were used for curvefitting. In this time (~126 ms), most muscle reflex responses would not produce significant forces because of inherent delays. There are two sources of delay in these reflexes: latency of EMG onset (Cresswell et al., 1994; Wilder et al., 1996) and the delayed development of force relative to EMG onset (electromechanical delay) (Thelen et al., 1994; Vos et al., 1991).

While this short measurement time minimized the effects of reflex muscle responses, it produced insufficient data to measure the trunk damping. Thus, some of the measured variation in trunk mass may have resulted from damping, and this may explain some of the variation in effective mass with the experimental parameters. Neglecting nonlinearities and damping effects may introduce small errors into the resulting estimates.

The increase in driving point stiffness was not proportional to the increase in effort (stiffness did not double with a nominal doubling of the external effort). There are several possible explanations for this including a nonlinear relationship between muscle stiffness and effort, and the influence of the significant passive stiffness of the spine and surrounding tissues. McGill et al. (1994) report measurements of the passive rotational stiffness of the trunk in the range 0.13–0.32 Nm/deg with the lowest values close to the neutral posture. The linear driving point trunk stiffness measured in the present study averaged 14.5 N/mm at 20% of effort, which can be converted to an equivalent rotational trunk stiffnesses by assuming that the loading point (harness) was between 100 and 200 mm from the center of rotation of the trunk. This gives rotational stiffness in the range 2.5–10 Nm/deg. This is approximately 20–30 times larger than the passive trunk stiffness reported by McGill et al. (1994), indicating that the stiffness due to muscle activation is the major component of the values reported here.

An increased effort requires additional muscle activation (Lavender et al., 1992). These increased muscle forces increase muscle stiffness which increase trunk stiffness which contributes to trunk stability, but they also paradoxically increase axial loading of the spine and thus tend to destabilize the trunk (Bergmark, 1989). Therefore, the increase in stiffness with effort would not necessarily imply an increase in trunk stability. An analytical study by Gardner-Morse and Stokes (1998) of extension efforts with and without abdominal muscle coactivation shows that while coactivation produces small increases in spinal compression, the overall result is an increase in spinal stability. Although it is difficult experimentally to measure trunk stability in vivo, experimental studies of trunk stiffness provide supporting data for analytical models that predict stability based on variables including muscle stiffness.

The results of this study confirm the importance of increased steady-state effort and associated muscle activation in stiffening the trunk for effectively isometric loading in the horizontal plane. A more complete understanding of the roles of muscles in stabilizing the trunk requires investigations of a wider range of loading states and exploration of the differences that might be present in individuals susceptible to low back disorders.

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