Published in final edited form as:

J Electromyogr Kinesiol. 2006 February ; 16(1): 51–57.

Active Trunk Stiffness Increases with Co-contraction

Patrick J. Lee, Ellen L. Rogers, and Kevin P. Granata*

Musculoskeletal Biomechanics Laboratories, Department of Engineering Science and Mechanics, School of Biomedical Engineering and Science, Virginia Polytechnic Institute and State University, 219 Norris Hall (0219), Blacksburg, VA 24061, United States

Abstract

Trunk dynamics, including stiffness, mass and damping were quantified during trunk extension exertions with and without voluntary recruitment of antagonistic co-contraction. The objective of this study was to empirically evaluate the influence of co-activation on trunk stiffness. Muscle activity associated with voluntary co-contraction has been shown to increase joint stiffness in the ankle and elbow. Although biomechanical models assume co-active recruitment causes increase trunk stiffness it has never been empirically demonstrated. Small trunk displacements invoked by pseudorandom force disturbances during trunk extension exertions were recorded from 17 subjects at two co-contraction conditions (minimal and maximal voluntary co-contraction recruitment). EMG data were recorded from eight trunk muscles as a baseline measure of co-activation. Increased EMG activity confirms that muscle recruitment patterns were different between the two co-contraction conditions. Trunk stiffness was determined from analyses of impulse response functions (IRFs) of trunk dynamics wherein the kinematics were represented as a second-order behavior. Trunk stiffness increased 37.8% ($p < 0.004$) from minimal to maximal co-activation. Results support the assumption used in published models of spine biomechanics that recruitment of trunk muscle co-contraction increases trunk stiffness thereby supporting conclusions from those models that co-contraction may contribute to spinal stability.

Keywords

Co-activation; Spine; Stiffness; Stability

1. Introduction

Muscle stiffness increases with activation as a result of the increased number of activated cross-bridges [25]. Muscle activation has been shown to increase joint stiffness in the elbow [1,35], the ankle [19], and the trunk [28,29]. With co-contraction the activity of the agonist and antagonist muscles increase thereby causing increased joint stiffness [17]. Although biomechanical models of the spine have assumed that recruitment of antagonistic co-contraction causes increased trunk stiffness [8,13] in it has never been empirically demonstrated in the trunk [9].

Trunk stiffness is an important contributor to spinal stability. Stability describes the ability to maintain equilibrium despite the presence of kinematic and/or control disturbances. Although passive tissues contribute to trunk stiffness, the ligamentous spine without active muscular support is unstable [6], therefore trunk stiffness is primarily associated with active muscles of the torso musculature [7]. Research concludes that paraspinal muscle reflexes also contribute to apparent stiffness of a joint by responding to perturbation movements and associated muscle strain with proportional muscle activation [5,14]. The stiffness data reported in the current

*Corresponding author. Tel.: +1 540 231 5316; fax: +1 540 231 4547. E-mail address: Granata@VT.edu (K.P. Granata).

study is the combined behavior of the active intrinsic muscle stiffness and the reflex response. This is referred to as the “effective stiffness” [4]. Effective stiffness can be accurately measured using small amplitude pseudo-random force disturbances and resulting trunk movement [28].

Empirical measurements demonstrate that trunk stiffness increases with exertion effort and associated muscle activity [4,28]. Clearly recruitment of antagonistic co-contraction must cause an increase in trunk muscle activation of both the antagonist and agonist muscle groups [12]. Therefore, we hypothesize that active trunk stiffness must increase with co-activation of the torso musculature due to the associated increase in recruitment. The specific aim of the project was to evaluate trunk dynamics, specifically stiffness, using systems identification analyses from data recorded during trunk extension exertions with minimal and maximal voluntary co-contraction recruitment.

2. Methods

2.1. Experiment

Seventeen subjects with no previous history of LBP participated after signing informed consent approved by the institutional review board at Virginia Tech. The mean (standard deviation) height and mass of the subjects were 175.5 (12.0) cm and 74.3 (14.2) kg, respectively.

The experiment consisted of an assessment of trunk stiffness at two recruitment conditions (minimal and maximal voluntary co-contraction) while maintaining constant trunk extension exertions. Subjects were attached to a servomotor (Pacific Scientific, Rockford, IL) via a harness and cable system such that anteriorly directed horizontal loads were applied at the T10 level of the trunk (Fig. 1). The servomotor applied constant isotonic preloads, which the subject was instructed to resist by maintaining an upright posture. Isotonic loads included 15% and 30% of the subject's maximum voluntary exertion (MVE). MVE force was measured in isometric trunk extension prior to the experiment. Subjects were instructed to maximally recruit their trunk flexor muscles as antagonists during maximum co-contraction trials while maintaining an upright posture against the preload. During minimum co-contraction trials subjects were instructed to relax their trunk flexor muscles while maintaining an upright posture. Recruitment and preload conditions were presented in random order.

During the exertions pseudorandom binary (PRB) perturbations of ± 70 N were superimposed on the force preload and were measured with a force transducer (Omega, Stamford, CT) attached to the motor. The force perturbations produced small flexion and extension movements of the trunk that were recorded with electromagnetic position sensors (Ascension Technology Corp., Burlington, VT; position resolution = 0.8 mm). Two six degree-of-freedom position sensors were taped to the skin over the spinous process at S1 and T10 and sampled at 100 Hz. Two trials with duration of 20 s were performed in each condition. Subjects were instructed to maintain the desired co-contraction effort throughout the trial. Fatigue was minimized by experimental design of low exertion levels, i.e., 15% and 30% MVC and also by requiring at least 1 min rest between each trial.

EMG data were collected during each trial to document co-active recruitment. EMG signals were collected from bipolar surface electrodes (Delsys, Boston, MA) on the left and right rectus abdominus (RA), lumbar paraspinal (LP), internal oblique (IO), and external oblique (EO) as described in Granata [11]. All EMG data were band-pass filtered in hardware between 20 and 450 Hz and sampled at 1000 Hz. The EMG signals were rectified and filtered using a 15 Hz, low-pass, seventh-order Butterworth filter. EMG were normalized to their corresponding peak EMG values recorded during maximum isometric flexion, extension, and lateral twisting exertions. Reported EMG recruitment data represents the average isometric baseline value from the first 250 ms of recorded data i.e., during steady state preload and prior to perturbations.

All trunk extension exertions were sagittally symmetric. Preliminary results indicated that there was no significant difference between left and right muscles within each muscle group. Therefore reported muscle recruitment is the average of the left and right muscle of each of the four muscle groups.

2.2. Analysis

Prior to system identification, data were processed to ensure accurate calculation of trunk stiffness. Force and displacement signals were filtered using a 75 Hz, low-pass, seventh-order Butterworth filter (Matlab, Natick, MA). Both force and displacement signals were treated with the same filter to avoid phase shift discrepancy from data processing. The data signals were demeaned prior to analysis to properly estimate the correlation functions [2]. Published techniques illustrate that adding white noise to the input signal reduces the effect of system noise by decorrelating the extraneous noise and data sampling from the input signal [24]. The magnitude of white noise added was 30% of one standard deviation of the input trunk force.

To quantify trunk stiffness, a nonparametric impulse response function (IRF) was estimated using the method of Moorhouse and Granata [28]. The IRF represents the transfer function relating the pseudorandom force input disturbance, $F(t)$, to the T10 displacement output, X_{meas} . IRFs were determined from the matrix solution relating the force autocorrelation function, C_{FF} , and the force/displacement cross-correlation function, C_{FX} [20,28]. This can be written mathematically as

$$\text{IRF} = \frac{1}{\Delta t} C_{FF}^{-1} C_{FX}, \quad (2.2.1)$$

where Δt is the sample period. To measure the quality of the IRF estimate, it was convolved with the measured pseudorandom force sequence, $F(t)$, to produce an estimate of the trunk displacement, X_{Est} . The equivalence between the predicted and observed displacements (Fig. 2) was measured in terms of the percentage of displacement variance accounted for (VAF), defined as

$$\text{VAF} = 100 \left(1 - \frac{\sum (X(t)_{\text{Meas}} - X(t)_{\text{Est}})^2}{\sum X(t)_{\text{Est}}^2} \right), \quad (2.2.2)$$

where VAF equal to 100% indicates the IRF exactly predicts the measured trunk displacement signal from the input force perturbations.

Inspection of the IRF shapes revealed that dynamics appeared similar to a second-order, underdamped system exhibiting one oscillation (Fig. 2). Therefore the nonparametric IRFs were subsequently parameterized by determining the coefficients of the second-order system that provided the best least-squares fit to the IRF. The compliance model had the form of

$$H(s) = \frac{1}{ms^2 + bs + k}, \quad (2.2.3)$$

where m is the effective trunk mass, b is the effective trunk damping, and k is the effective trunk stiffness. The quality of the second-order fit compared with the nonparametric IRF was calculated as a percentage and referred to as model accuracy.

2.3. Statistical analysis

Repeated measures statistical analysis, ANOVA, was performed to determine the effect of co-contraction condition and preload on trunk stiffness and recorded muscle recruitment. Significance was determined at the level of $\alpha < 0.05$. Tukey honest-significant difference (HSD) post hoc analyses were used to compare differences among significant treatments.

3. Results

Antagonistic muscle recruitment was successfully achieved as evidenced by significantly increased flexor muscle activity (Fig. 3). During the maximum co-contraction conditions the rectus abdominis EMG activity was greater than the minimum co-contraction conditions by 12.5% ($p < 0.005$). Similarly, external obliques increased by 19.4% ($p < 0.02$) and internal oblique recruitment increased 7.5% ($p < 0.04$) during co-contraction recruitment conditions. Likewise, lumbar paraspinal muscle groups demonstrated a trend toward increased EMG activity during the maximum co-contraction conditions but failed to reach statistical significance ($p = 0.248$). As expected, baseline extensor muscle activity increased significantly with preload. During the 30% MVC trials the lumbar paraspinal activity was 27.5% ($p < 0.03$) greater than during the 15% MVC exertions. Similarly, internal obliques increased 11.9% ($p < 0.01$) with preload. Muscle recruitment in the flexor muscles did not statistically change with respect to preload.

IRFs accurately predicted the active trunk kinematic response as evidenced by mean VAF of 79.4 (13.0)%. It is clear by inspection (Fig. 2) that the second-order model provides an excellent description of the trunk dynamics within experimental conditions. This was validated numerically by determining the second-order model accuracy with a mean of 95.2 (3.6)%. Therefore, the analysis accurately represented measured trunk dynamics.

Trunk stiffness increased 37.8% ($p < 0.004$) with co-contraction recruitment (Table 1). Trunk stiffness increased 18.4% ($p < 0.002$) with preload effort. There was a significant ($p < 0.03$) interaction between co-contraction and preload conditions. Within the 15% MVE preload condition the trunk stiffness increased by 56.0% ($p < 0.0002$) with co-contraction recruitment. Co-contraction was associated with a 24.3% increase ($p < 0.001$) in trunk stiffness within the 30% MVE pre-load condition. Within the minimal co-contraction condition stiffness increased 35.1% with preload ($p < 0.01$). Conversely, preload did not affect stiffness within the maximal co-contraction conditions ($p = 0.33$). This was attributed to the fact that measured activity of the lumbar paraspinal muscle group significantly ($p < 0.05$) increased with preload during minimum co-contraction. Measured activity of the lumbar paraspinal muscle group was not significantly ($p = 0.128$) different with respect to preload during maximum co-contraction.

Mean effective mass was 47.8 (18.7) kg. Effective mass increased 22.3% with preload effort ($p < 0.03$) but was not statistically affected by co-contraction recruitment. Mean effective damping was 487.0 (132.8) N s/m and increased 21% with preload ($p < 0.0001$). Effective damping did not change with co-activation condition. The damping ratio, ζ , was calculated from the measured values of effective mass, m , effective damping, b , and stiffness coefficients, k , such that $\zeta = \frac{b}{2\sqrt{mk}}$. Mean value of damping ratio was 0.575 (0.154) and did not change with preload effort, but increased 19.2% with co-activation ($p < 0.007$).

4. Discussion

Voluntary recruitment of co-contraction in the torso musculature has been associated with an increase in muscle recruitment of both agonist and antagonist muscle in order to maintain a neutral posture [12,17]. Recognizing that increased muscle activity is associated with greater joint stiffness [22] the co-contraction recruitment about the ankle [17] and elbow [23] was also associated with an increase in joint stiffness. In the trunk, antagonistic co-contraction recruitment in the abdominal musculature generates trunk flexion moment that must be offset by increased activity in extensor muscles to maintain an upright posture. Therefore, results demonstrated increased activity in both trunk flexor and extensor muscle groups when recruiting voluntary co-contraction of the flexor muscles during trunk extension exertions. Trunk stiffness was quantified as a function of co-contraction effort. Results demonstrated that

trunk stiffness was greater when antagonistic co-contraction was recruited. Therefore, this study empirically validates the assumption used in published models of spine biomechanics that co-contraction influences trunk stiffness [9].

Increased extension exertion has been shown to increase trunk stiffness presumably due to increased activity and stiffness in the trunk extensor muscles [28]. This effect was confirmed in the current study. Increased pre-load efforts were associated with greater lumbar paraspinal and internal oblique muscle activity as well as significantly increased trunk stiffness. The extent to which the change in trunk stiffness can be attributed to the increased activity in the extensor muscles versus the increased activity in the flexor muscles should be addressed in future research. A significant interaction between recruitment condition and preload level indicated that the potential influence of co-contraction recruitment is modulated by trunk extension force. At high preload levels there is a limit to the available increase in extensor muscle activity. Recognizing that extensor muscle activity must increase commensurate with antagonistic recruitment to balance the trunk moment, this limits the upper bound of antagonistic co-contraction. Moreover, inhibitory neural pathways have been identified between agonist-antagonist muscle pairs in many joints [30,32] that may limit antagonistic co-contraction recruitment with increase extension effort. Results from the measured EMG support this mechanism. The increase in lumbar paraspinal activity associated with preload was less during the maximum co-contraction condition than during the minimum co-contraction condition. Hence the ability to voluntarily modulate trunk stiffness is influenced by preload effort. Recognizing that spinal stability is related to recruitment of active muscle stiffness [8,26] this interaction may indicate potential risks of spinal instability in heavy exertion tasks.

An increase in effective damping was to be expected [15,16,28]. Joint stiffness and joint damping tend to increase in such a way as to maintain a constant damping ratio [22]. Our results were consistent with the literature in this regard, i.e., damping ratio was independent of preload effort. The joint dynamics described above, such as the constant damping ratio, are technically valid only for exertions without co-contraction [22]. However, results show that damping ratio increased with co-contraction. Hunter and Kearney [22] acknowledge that co-active recruitment can change joint dynamics and observed this effect in the ankle [18]. The ability to voluntarily modulate trunk dynamics by means of stiffness and damping ratio provides robust neuromuscular control throughout a variety of dynamic loading conditions. Evidence suggests that patients with low-back pain have limited neuromuscular control of the torso dynamics [31] and may recruit abnormally high levels of co-contraction to increase trunk stiffness in order to maintain spinal stability [34].

Mean effective mass recorded from the measurement protocol was 64.3% of total body mass. Trunk mass should be independent of joint torque [10,19] and therefore independent of co-activation and preload. Results indicated that co-activation did not influence effective mass. However preload level significantly increased effective mass. Anthropometric data tables suggest that trunk mass is approximately 55% of total body mass [3]. Effective mass was 57% of body mass at the lower preload but was significantly greater at the higher pre-load. Results suggesting significant changes in effective mass with exertion effort are unexpected. Gardner-Morse and Stokes [33] observed similar effects and concluded that the mass coefficient may not fully describe the inertial effects of the flexible trunk. In a linear second-order dynamic model the effective mass parameter represents a mass being rotated about a constant center of rotation (CoR). Our analysis makes no assumption about the location of the CoR but it assumes the location is constant. We are unaware of any study to determine whether the instantaneous CoR of the lumbar spine changes with extension effort. Results suggest the CoR in the sagittal plane of the trunk may be a function of extension effort. If CoR moves caudally with increased trunk extension effort, then trunk dynamics will demonstrate greater effective driving point mass due to the decreased moment arm from the CoR to the center of mass. Further research

is necessary to understand this phenomenon. Fortunately the analyses were capable of solving separately for effective mass and stiffness, therefore our concerns regarding mass should not influence stiffness results.

Results must be considered in light of methodological limitations. The constant perturbation force used in the current study may cause different magnitudes of motion response between different subjects and preload conditions. The small angle kinematic response to the force perturbations is influenced by the mass and stiffness of the system. Hence, anthropometry and preload effort may influence the muscular reflex response and its associated contribution to effective stiffness. Although the system was treated as a linear system, it has been demonstrated that disturbance amplitude may influence results [22]. Typical force impulse values ranged from 8 to 38 N s. From the measured values of mass, damping and stiffness it is readily demonstrated that trunk displacement ranged from 9 to 42 mm. Smaller kinematic disturbances of consistent displacement amplitude may improve measurement accuracy. Therefore, it is suggested that instead of using force perturbations, future research should investigate stochastic position perturbations. Trunk stiffness results include the effects from both intrinsic muscle stiffness and reflex contributions to trunk dynamics. Improved analysis techniques should be applied to separate trunk stiffness into its intrinsic and reflexive components [21]. The analyses were also limited by assumptions regarding the trunk as a point mass rather than a distributed system with movement throughout the length of the spine. Parameter values including the magnitude of the stiffness coefficient may be influenced by placement location of the motion sensor represented by a scalar value of distance between the CoR and the sensor. Fortunately, this does not influence the statistical trends. Treating the trunk as a point mass rather than a distributed system was necessary as a result of the difficulty of noninvasively measuring rapid movement of individual spinal segments. Nonetheless, biomechanical models [8] suggest that similar behavior will be observed if one could do similar experiments for smaller segments of the spine, i.e., co-contraction contributes to increased stiffness during voluntary trunk extension exertions.

This study confirms that torso dynamics are influenced by recruitment co-contraction. Accurate estimates of joint dynamics parameters must therefore include effects of co-active recruitment. Biomechanical models of spinal stability suggest that antagonistic co-contraction can be used to augment stability by increasing the net bending stiffening of the spinal column [8,13,27]. Our empirical measurements support those conclusions by demonstrating that antagonistic co-contraction can be used to augment trunk stiffness. Existing stability models of the spine are typically limited to static stability analyses. Current results suggest damping ratio of trunk dynamics can be modulated voluntarily. Thus, future analyses should investigate the voluntary control of dynamic biomechanical stability of the lumbar spine.

Acknowledgement

This research was supported in part by a grant R01 AR49932 from CDC/National Institute for Occupational Safety and Health.

Biographies

Patrick J. Lee earned his M.S. in mechanical engineering in the Musculoskeletal Biomechanics Laboratory, Virginia Tech. His research focuses on understanding the stability and muscle recruitment of trunk flexion exertions associated with pushing-related manual materials handling tasks.

Ellen L. Rogers earned her M.S. in mechanical engineering in the Musculoskeletal Biomechanics Laboratory, Virginia Tech. Her research focuses on quantification of active torso stiffness and paraspinal reflex dynamics.

Kevin Granata earned an MS in physics from Purdue University and a Ph.D. in biomechanics from The Ohio State University. He served as Research and Engineering Director of the Motion Analysis Laboratory at the University of Virginia where he was an Associate Professor of Orthopaedic Surgery and Biomedical Engineering. He is currently Director of the Musculoskeletal Biomechanics Laboratory and Associate Professor in the Department of Engineering Science and Mechanics at Virginia Tech. His research focuses on neuromuscular dynamics and control of human movement with specific interests in musculoskeletal stability for injury prevention and rehabilitation.

References

- [1]. Agarwal GC, Gottlieb GL. Compliance of the human ankle joint. *J. Biomech. Eng* 1977;99:166–170.
- [2]. Bendat, JS.; Piersol, AG. *Random Data: Analysis and Measurement Procedures*. John Wiley and Sons; New York: 2000.

- [3]. Chaffin DB, Stump BS, Nussbaum MA, Baker G. Low-back stresses when learning to use a materials handling device. *Ergonomics* 1999;42:94–110. [PubMed: 9973874]
- [4]. Cholewicki J, Simons APD, Radebold A. Effects of external trunk loads on lumbar spine stability. *J. Biomech* 2000;33:1377–1385. [PubMed: 10940396]
- [5]. Crago PE, Houk JC, Hasan Z. Regulatory actions of human stretch reflex. *J. Neurophysiol* 1976;39:925–935. [PubMed: 978238]
- [6]. Crisco JJ, Panjabi MM. Euler stability of the human ligamentous lumbar spine: Part I Theory, Clin. *Biomech* 1992;7:19–26.
- [7]. Dolan P, Earley M, Adams MA. Bending and compressive stresses acting on the lumbar spine during lifting activities. *J. Biomech* 1994;27:1237–1248. [PubMed: 7962011]
- [8]. Gardner-Morse MG, Stokes IAF. The effects of abdominal muscle coactivation on lumbar spine stability. *Spine* 1998;23:86–92. [PubMed: 9460158]
- [9]. Gardner-Morse MG, Stokes IAF, Laible JP. Role of muscles in lumbar stability in maximum extension efforts. *J. Orthop. Res* 1995;13:802–808. [PubMed: 7472760]
- [10]. Gottlieb GL, Song Q, Almeida GL, Hong DA, Corcos D. Directional control of planar human arm movement. *J. Neurophysiol* 1997;78:2985–2998. [PubMed: 9405518]
- [11]. Granata KP, Marras WS. An EMG assisted model of biomechanical trunk loading during free-dynamic lifting. *J. Biomech* 1995;28:1309–1317. [PubMed: 8522544]
- [12]. Granata KP, Marras WS. The influence of trunk muscle coactivity upon dynamic spinal loads. *Spine* 1995;20:913–919. [PubMed: 7644956]
- [13]. Granata KP, Orishimo K. Response of trunk muscle coactivation to changes in spinal stability. *J. Biomech* 2001;34:1117–1123. [PubMed: 11506782]
- [14]. Granata KP, Slota GP, Bennett BE, Kang HG. Paraspinal muscle reflex dynamics. *J. Biomech* 2004;37:241–247. [PubMed: 14706327]
- [15]. Granata KP, Slota GP, Wilson SE. Influence of fatigue in neuromuscular control of spinal stability. *Hum. Factors* 2004;46:81–91. [PubMed: 15151156]
- [16]. Granata KP, Wilson SE. Trunk posture and spinal stability. *J. Biomech* 2001;16:650–659.
- [17]. Granata KP, Wilson SE, Massimini AK, Gabriel R. Active stiffness of the ankle in response to inertial and elastic loads. *J. Electromyogr. Kinesiol* 2004;14:599–609. [PubMed: 15301778]
- [18]. Hunter I, Kearney RE, Weiss PL. Stiffness-torque-EMG relations at various tibialis anterior and triceps surae co-contraction levels obtained by real-time feedback of elastic stiffness. *Soc. Neurosci. Abstr* 1984;10:334.
- [19]. Hunter IW, Kearney RE. Dynamics of human ankle stiffness: variation with mean ankle torque. *J. Biomech* 1982;15:747–752. [PubMed: 7153227]
- [20]. Hunter IW, Kearney RE. Two-sided linear filter identification. *Med. Biol. Eng. Comput* 1983;21:203–209. [PubMed: 6887997]
- [21]. Kearney RE, Stein RB. Identification of intrinsic and reflex contributions to human ankle stiffness dynamics. *IEEE Trans. Biomed. Eng* 1997;44:493–504. [PubMed: 9151483]
- [22]. Kearney RE, Hunter IW. System identification of human joint dynamics. *Crit. Rev. Biomed. Eng* 1990;18:55–87. [PubMed: 2204515]
- [23]. Lacquaniti F, Licata F, Soechting JF. The mechanical behavior of the human forearm in response to the transient perturbations. *Biol. Cybern* 1982;44:35–46. [PubMed: 7093368]
- [24]. Ljung, L.; Ljung, EJ. *System Identification: Theory for the User*. Prentice-Hall, Upper Saddle River, NJ: 1998.
- [25]. Ma SP, Zahalak GI. The mechanical response of the active human triceps brachii muscle to very rapid stretch and shortening. *J. Biomech* 1985;18:585–598. [PubMed: 4055813]
- [26]. McGill SM. Low back stability: from formal description to issues for performance and rehabilitation. *Exer. Sport Sci. Rev* 2001;29:26–31.
- [27]. Meakin JR, Hukins DW, Aspden RM. Euler buckling as a model for the curvature and flexion of the human lumbar spine. *Proc. Roy. Soc. Lond. B* 1996;263:1383–1387.
- [28]. Moorhouse KM, Granata KP. Trunk stiffness and dynamics during active extension exertions. *J. Biomech.* (in press)

- [29]. Moorhouse KM, Kaufman G, Granata KP. Parametric deconvolution for calculation of dynamic trunk stiffness. *Int. Soc. Electromyogr. Kinesiol.* 2004(Conference Proceeding)
- [30]. Myklebust BM, Gottlieb GL, Penn RD, Agarwal GC. Reciprocal excitation of antagonist muscles as a differentiating feature in spasticity. *Ann. Neurol* 1982;12:367–374. [PubMed: 7149662]
- [31]. Radebold A, Cholewicki J, Polzhofer GA, Green TP. Impaired postural control of the lumbar spine is associated with delayed muscle response times in patients with chronic idiopathic low back pain. *Spine* 2001;26:724–730. [PubMed: 11295888]
- [32]. Simon, SR. *Orthopaedic Basic Science*. American Academy of Orthopaedic Surgery; Rosemont, IL: 1994.
- [33]. Stokes IAF, Gardner-Morse MG, Henry SM, Badger GJ. Decrease in trunk muscular response to perturbation with preactivation of lumbar spinal musculature. *Spine* 2000;25:1957–1964. [PubMed: 10908940]
- [34]. Van Dieen JH, Cholewicki J, Radebold A. Trunk muscle recruitment patterns in patients with low back pain enhance the stability of the lumbar spine. *Spine* 2003;28:834–841. [PubMed: 12698129]
- [35]. Zhang LQ, Rymer WZ. Simultaneous and nonlinear identification of mechanical and reflex properties of human elbow joint muscles. *IEEE Trans. Biomed. Eng* 1997;44:1192–1209. [PubMed: 9401219]

Fig. 1. Experimental setup. Force perturbations in a pseudo-random fashion superimposed on isotonic loads were applied to subjects to elicit small trunk movement during two co-activation conditions (maximal and minimal voluntary recruitment of flexor muscles). Subjects were securely strapped into a rigid pelvic support structure to isolate movement of the trunk for all trials.

Fig. 2.

IRFs relating trunk displacement to the applied trunk force during maximal and minimal co-activation and the superimposed second-order least-mean-squares fit. This clearly illustrates that a linear second-order model can accurately describe the trunk dynamics. Conditions represent maximum co-activation (a) and minimum co-activation (b) and demonstrate differences in trunk stiffness.

Fig. 3.

Comparison of EMG activity with co-activation. Muscle activity in the rectus abdominus (RA), external oblique (EO), and internal oblique (IO) increased significantly ($p < 0.05$) during maximum co-activation as denoted with *. No significant EMG change was noted in the lumbar paraspinals (LP).

Table 1
Mean (standard deviation) of trunk stiffness during active voluntary exertions

	15% MVE	Voluntary exertion effort 30% MVE	Average
Minimum co-contraction	3.12 (1.12)	4.22 (1.66)	3.67 (1.50) ^a
Maximum co-contraction	4.87 (1.81)	5.24 (1.98)	5.05 (1.87)
Average	3.99 (1.72) ^b	4.73 (1.88) ^b	4.36 (1.83)

Stiffness values are expressed in N/mm.

^a Significant effect for exertion.

^b Significant effect for co-contraction.