

Neuromuscular Neutral Zones Associated With Viscoelastic Hysteresis During Cyclic Lumbar Flexion

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Study Design. The reflexive EMG from the L3–L4 to L5–L6 multifidus of the *in vivo* feline was recorded during application of single passive flexion–extension cycle of the lumbar spine.

Objective. To determine the effect of viscoelastic hysteresis associated with a single-cycle flexion–extension and of increasing cycle frequency on the initiation and cessation displacement and tension thresholds of reflexive EMG from the multifidus muscles.

Summary of Background Data. It is known that reflexive EMG can be recorded from some paraspinal muscles as a result of mechanical stimulation of lumbar ligaments and other viscoelastic structures. It is also known that mechanical neutral zones exist in the spine, that viscoelastic hysteresis is associated with a stretch–release cycle, and that the rate of stretch and release has a profound impact on viscoelastic tissue responses. It is unknown what are the neurologic neutral zones of the spine within which reflexive EMG does not exist, as well as the dependence of such neurologic neutral zones on viscoelastic hysteresis and increasing frequency of a flexion–extension cycle.

Methods. Single passive flexion–extension cycles of frequencies ranging from 0.1 to 1.0 Hz were applied to the lumbar spine of the feline while recording intramuscular EMG from the L3–L4 to L5–L6 multifidus. The displacement and tension thresholds associated with the initiation and cessation of EMG activity during the cycle were analyzed with respect to the cycles' viscoelastic hysteresis and frequency. The peak EMG discharge was tested for relationships with cycle frequency.

Results. The displacement and tension thresholds during the flexion phase of the cycle were significantly lower than the corresponding thresholds in the extension phase of the cycle. As the cycle frequency increased, EMG was triggered significantly earlier (lower displacement and tension thresholds) in the flexion phase and terminated earlier (higher displacement and tension thresholds) in the extension phase. The peak EMG was significantly larger as cycle frequency increased.

Conclusions. Reflexive muscle forces are triggered at lower displacement or tension during flexion but diminish early during extension, leaving the spine unprotected for a substantial part of the extension movement. The muscle forces are recruited earlier and with larger intensity as the velocity of the movement increases, lending

more protection to the spine. Faster extension movement, however, creates a larger window during which the spine is exposed to instability and injury because of lack of muscle forces. [Key words: lumbar, muscle, multifidus, EMG, reflex] **Spine 2001;26:E314–E324**

It is established that the stability of the spine is maintained by forces generated by the passive viscoelastic structures (ligaments, discs, and capsules)^{16,25} and, by a major part, with active forces generated from muscular contractions.^{9,14,15,23,25,39,42} Recent evidence points out that one important source of muscular forces aimed at maintaining lumbar stability originates from reflexive rather than voluntary contractions. The reflexive feedback control of muscular contraction consists of afferents in the viscoelastic structures of the ligaments, discs, and capsules,^{13,34} spinal interneurons, and selected trunk muscles. Mechanical or electrical stimuli applied to the afferents were shown to trigger EMG activity in the multifidus and longissimus muscles.^{11,12,36,38} An important issue in the study of the dynamic response of the sensorimotor feedback loop of the reflexive muscular activity of the spine is the sensitivity threshold to stretch or tension developed within the viscoelastic tissues containing the afferents. Solomonow et al³⁶ observed that reflexive EMG activity recorded from the multifidus was triggered only after an appreciable elongation or tension was developed in a single isolated supraspinous ligament. The observed thresholds were, however, significantly lower when several ligaments, discs, and capsules were loaded simultaneously. The importance of such observations is vested in the general question: "Is the reflexive muscular activity adaptive to various external and internal factors, or is it independent of the prevailing conditions?" Internal factors in the above case constituted inputs from afferents in a single ligament as opposed to multi-inputs from several structures. Other internal and external factors could be the effect of motion velocity, orientation relative to the gravity vector, inherent tissue load, external load, and trunk angles, *etc.*

It was shown by Panjabi³⁰ that the spine is relatively compliant for small perturbations about the neutral position. These small displacement ranges within which stability is not disturbed were designated as neutral zones (NZ). It is expected that reflexive muscular contractions will not be present within the NZ, as the viscoelastic tissues are minimally disturbed. Conversely, as the relative displacement of vertebrae exceeds the NZ, the resistance to displacement increases and the vis-

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coelastic tissues are substantially disturbed. One should expect to observe some reflexive muscle activity at this range because of stimulation of afferents in the viscoelastic tissues. Overall, it is reasonable to assume that neuromuscular neutral zones (NNZs) could also exist, such that reflexive muscle forces are absent or low within the mechanical neutral zones (MNZs) and rather active when the MNZ is exceeded. Panjabi³⁰ also observed that muscular activity tend to minimize the MNZ, hereby demonstrating an association between the two phenomena. Direct and simple dependence of the NNZ on the MNZ, however, should not be expected for several reasons.

As viscoelastic structures are known to exhibit creep or tension-relaxation when subjected to load or stretch, respectively, the effectiveness of the afferents within such tissues is expected to markedly change. Recently, it was shown that application of prolonged cyclic or static passive lumbar flexion to the *in vivo* feline resulted in viscoelastic tissue tension-relaxation and significant reduction of reflexive muscular activity in the multifidus.^{37,41}

The length-force behavior of viscoelastic tissues is also known to exhibit hysteresis in the course of a single stretch-release cycle. Such behavior is also expected to modify the performance of the reflexive muscular activity of the lumbar spine. The rate (or frequency) at which mechanical stimuli are applied to afferents²⁸ and to viscoelastic tissues^{31,40} is also known to modify their behavior and possibly the overall activation pattern of the reflexive muscular contraction.

The objective of this investigation is to determine the effect of viscoelastic hysteresis developed at different lumbar cyclic flexion-extension frequencies on the response pattern of the reflexive muscular activation thresholds (e.g., NNZ) while excluding creep or tension-relaxation. It is also the objective to assess the reflexive response threshold pattern of the multifidus muscles of one lumbar joint above and below the level of stimulus application. We hypothesize that viscoelastic hysteresis would modify the activation threshold of the multifidus muscles and that such changes will be further influenced by the frequency at which a passive flexion-extension cycle is applied.

■ Methods

Preparation. Eight adult cats (4.7 ± 0.44 kg), anesthetized with a single dose of chloralose (60 mg/kg), were used in a protocol approved by the Institutional Animal Care and Use Committee. The skin over the spine was dissected from the thoracic level to the sacral level to expose the intact dorsolumbar fascia. The preparation was placed in a rigid stainless steel frame that allowed the isolation of various lumbar levels *via* external fixation. The external fixation did not prevent micromotion or macromotion in the lumbar spine and was intended to avoid interaction with thoracic and sacral structures.

Instrumentation. Three pairs of fine stainless steel wire EMG electrodes, insulated except for a 1-mm exposed tip, were inserted 3–4 mm apart into the multifidus muscles of the L3–L4,

L4–L5, and L5–L6 (cat has seven lumbar vertebrae) on the right side 6–8 mm from the midline. Each electrode pair constituted the input to a differential EMG amplifier with a 110-db common mode rejection ratio, gain capabilities of up to 200,000, and a band pass filter in the range of 6–500 Hz. A ground electrode was inserted elsewhere in the preparation. The EMG response from each of the three channels was monitored on the oscilloscope, recorded and stored in a computer with a sampling rate of 1000 Hz.

An S-shaped stainless steel hook was inserted around the middle of the L4–L5 supraspinous ligament and connected to the actuator of the Bionix 858 Material Testing System (MTS Inc., St. Paul, MN), which was instrumented with a computer-controlled displacement system. The output of a force transducer placed in the actuator was sampled into the computer along with its displacement. The overall setup is shown schematically in Figure 1. In summary, EMG from three lumbar levels, displacement, and tension data were recorded and stored in the computer for later analysis.

Protocol. One external fixator was applied to the L1 dorsal process, and a second one to the L7 dorsal process. The stainless steel hook was attached to the actuator of the Bionix 858 System, which was set to deliver a single sinusoidal displacement cycle of 12–15 mm peak (according to the animal's size) with a pretension of 0.5 N. Vertical displacement of 12–15 mm was shown in our previous work⁴¹ to result in ~15% axial strain of the supraspinous ligament, which is about 75% of its maximal physiologic strain. A pretension of 0.5 N was selected as it standardizes the initial conditions across all preparations while having very little effect, if any, on the supraspinous ligament's initial elongation-tension.

Initially, five sinusoidal cycles at 0.25 Hz and 12–15-mm peak displacement were applied to allow the preparation to settle into a steady-state condition, to assess and prevent any motion artifacts or instabilities at the fixation points, and to assess the integrity of noise-free EMG recordings in all three channels. After the five cycles a 20-minute rest period was observed to allow recovery of any tension-relaxation that may have developed in the viscoelastic structures.^{8,26}

After the 20-minute recovery period, single-test cycles at frequencies of 0.1, 0.25, 0.5, 0.75, and 1.0 Hz were applied at random. In order to assess repeatability, each test cycle was repeated. A 20-minute rest period was allowed between each recording.

EMG Threshold Analysis. The first 500 msec of each EMG record, which was before the initiation of movement by the actuator, was used as a benchmark for baseline EMG signal level. The mean absolute value (MAV) of this baseline was calculated; the signal recorded from each level was full wave rectified and smoothed with a low-pass filter of a 200-msec time constant to yield the EMG MAV. After this period any point along the signal that exceeded three times the baseline MAV was denoted as the threshold of activity in that channel. The corresponding tension and displacement values during the stretch (increasing) phase of the cycle were recorded as threshold values for that particular trial, frequency, and lumbar level. Similarly, as the EMG dropped below the three times baseline MAV threshold in the release (decreasing) phase of the cycle, the corresponding tension and displacement were recorded as the threshold of end of activity. This automatic procedure was

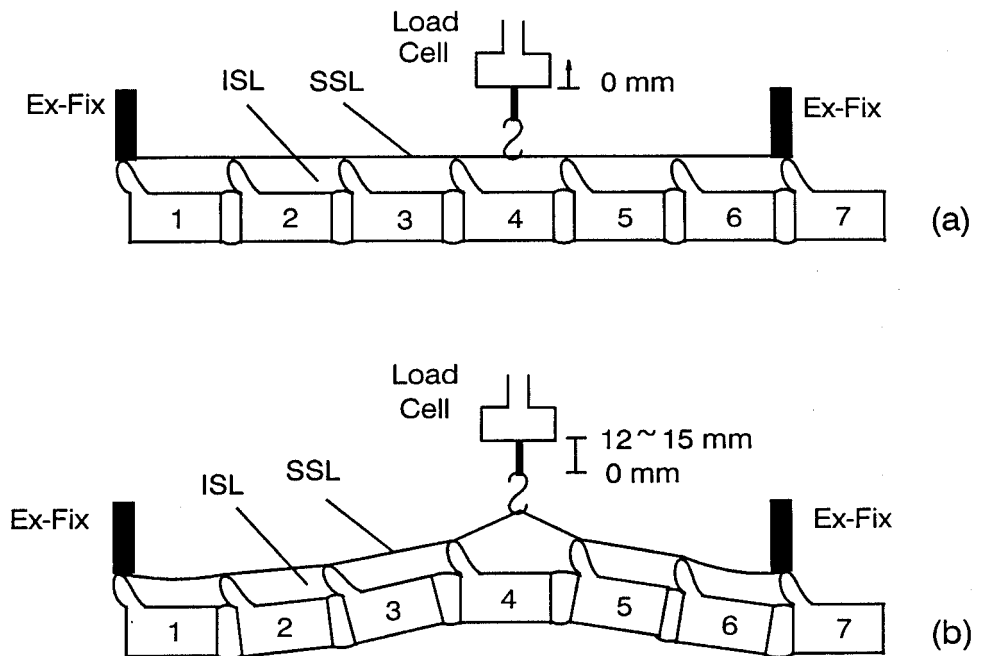


Figure 1. Schematic of the experimental setup: during rest (A) and during full displacement (B) applied via the L4–L5 supraspinous ligament.

manually supervised to insure that any unexpected signal artifacts were not detected as thresholds.

The displacement and tension thresholds of each of the three EMG channels of each of the eight preparations at the same cycle frequency were pooled and the mean (\pm SD) were calculated and plotted as a function of cycle frequency. The two measurements made for each condition were first averaged before pooling with the corresponding values of the other preparations.

EMG Peak Amplitude Analysis. The peak MAV at a cycle frequency of 0.1 Hz was used as a normalization value for the MAV recorded at the remaining cycle frequencies of the same preparation. The normalized values at each frequency of all the preparations (average of the two trials from each) were pooled, and the mean (\pm SD) was calculated and plotted as normalized peak MAV *versus* frequency plot to assess whether the contraction level increased or decreased because of cycle frequency.

Statistical Analysis. As stated above, the two responses for replicated trials for each cat were averaged to represent that cat's response for each condition. One-way analysis of variance with repeated measures was then applied to determine the existence of significant effect of frequency on the tension and displacement thresholds in the stretch and release phases as well as for the peak EMG mean absolute value. The significance of differences between the stretch and release thresholds was evaluated by paired *t* tests to determine possible effects of hysteresis.

Results

A typical recording from a 0.1-Hz single-cycle application to the lumbar spine *via* the L4–L5 supraspinous ligament is shown in Figure 2. The displacement stimulus input as a function of time is shown in the bottom trace, whereas the resulting recorded tension *versus* time is shown in the trace above. The top three traces show the recorded EMG from L3–L4 to L5–L6 with the associated

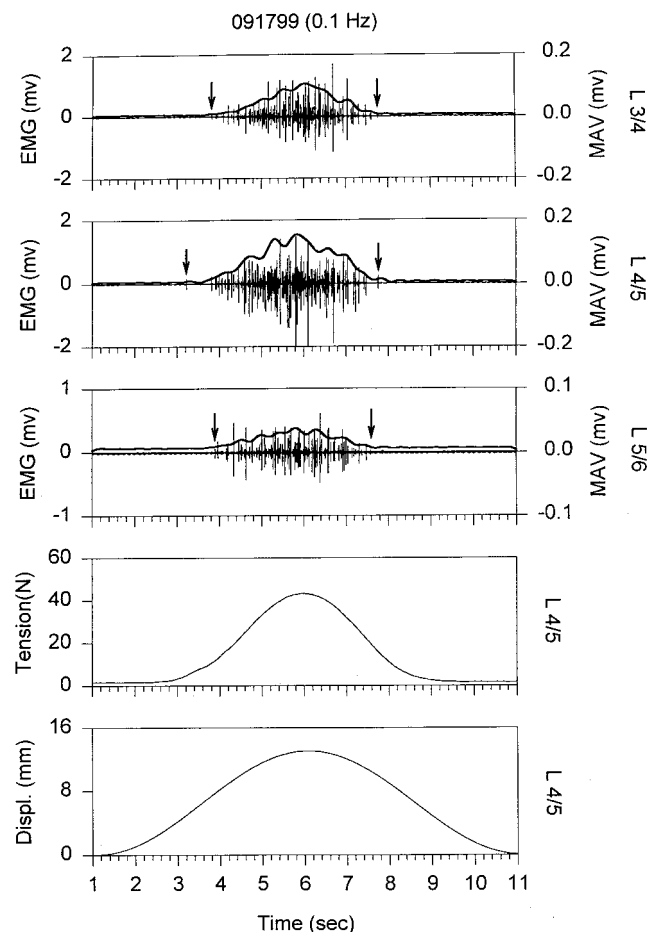


Figure 2. Typical recording of EMG from L3–L4, L4–L5, and L5–L6 (top three panels), tension, and the applied displacement (bottom panel) at a cyclic frequency of 0.1 Hz. The mean absolute value (MAV) of the EMG from each lumbar level is superimposed on the raw EMG traces. The arrows at the beginning and end of the EMG discharge indicate the thresholds detected in the stretch and release phase of this cycle.

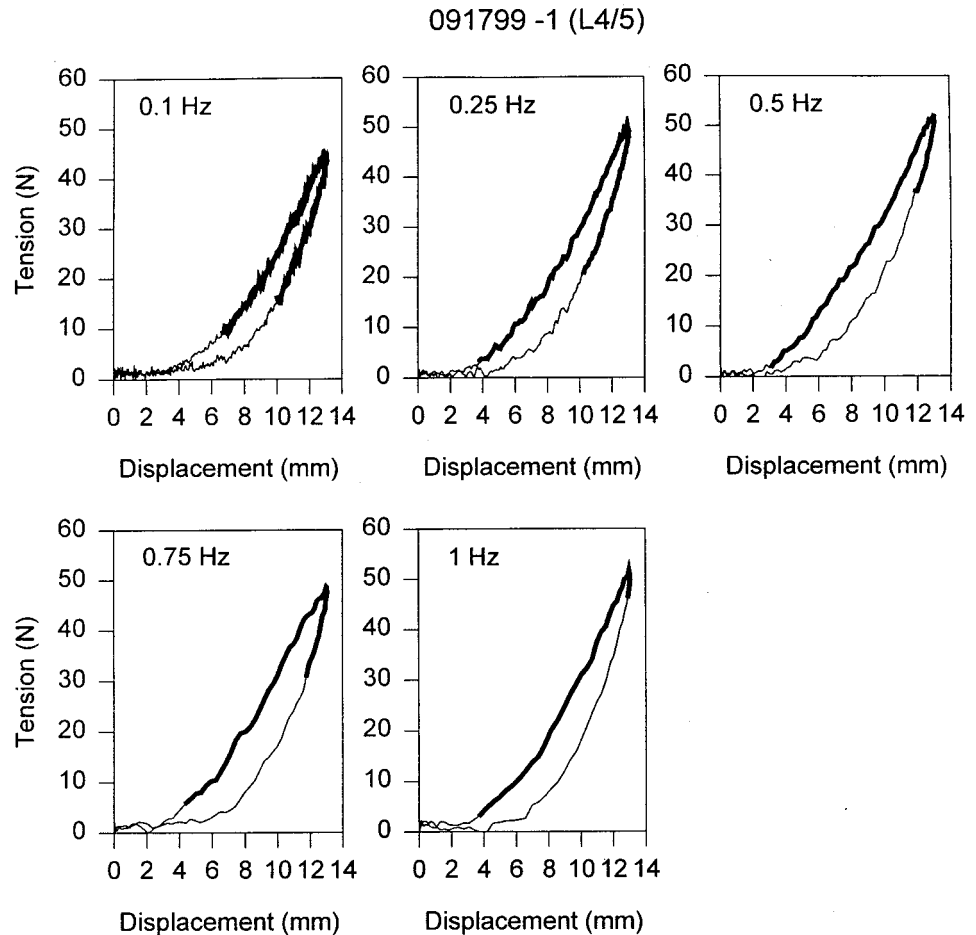


Figure 3. Typical hysteresis curves where the tension *versus* displacement of a single cycle at each of the frequencies employed is shown; the period where EMG was recorded from its initiation in the stretch phase to its termination in the release phase is designated in boldface on the curve.

MAV of the EMG superimposed. The arrows at the initial and final phase of the cycle denote the detected stretch and release thresholds associated with this cycle.

In Figure 3 the tension *versus* displacement hysteresis curves of L4–L5 of the same preparation is shown for each of the five cycle frequencies. The bold portion on each of the hysteresis curves denotes the portion of the cycle during which EMG was recorded in that level from the initiation threshold to the cessation threshold. From Figure 3 it is evident that EMG was first triggered at a displacement of 7 mm for a frequency of 0.1 Hz, and at displacements of 3.75, 3, 4.1, and 3.95 mm for frequencies of 0.25, 0.5, 0.75, and 1.0 Hz, respectively. The displacement thresholds at which reflexive EMG terminated were 10, 10.2, 12, 12, and 13 mm, for the frequencies of 0.1, 0.25, 0.5, 0.75, and 1.0 Hz, respectively. Similarly, the tension thresholds at which EMG was initiated in the stretch phase followed a decreasing pattern for increasing cycle frequency and an increasing tension threshold pattern for increasing cycle frequency for the release phase.

Figure 4 provides the mean (\pm SD) of the displacement thresholds (top row) *versus* cycle frequency for L3–L4, L4–L5, and L5–L6 calculated from the data pooled from the eight preparations. The bottom row provides the respective mean (\pm SD) of the tension thresholds for the three lumbar levels. The same data are summarized

numerically in Table 1. In general, the mean initial displacement threshold (during the stretch phase) at L3–L4 decreased from 4.82 to 4.08 mm (15.4% decrease), 3.31 mm (31.4% decrease), 3.20 mm (33.5% decrease), and 2.62 mm (–45.6% decrease) for the frequencies of 0.1, 0.25, 0.5, 0.75, and 1.0 Hz, respectively.

For L4–L5 the displacement thresholds in the stretch phase of the cycle decreased from 7.07 to 5.69 mm (19.6% decrease), 4.50 mm (36.3% decrease), 4.12 mm (41.7% decrease), and 4.10 mm (42% decrease) for frequencies of 0.1, 0.25, 0.5, 0.75, and 1.0 Hz, respectively. Similarly, at L5–L6 displacement thresholds decreased from 8.47 to 6.77 mm (20.1% decrease), 5.93 mm (29.9% decrease), 5.02 mm (40.8% decrease), and 5.13 mm (39.4% decrease) for cycle frequency increase from 0.1 to 1.0 Hz.

The displacement thresholds during the release phase of the cycle demonstrate a mild increase. In L3–L4 gradual changes of up to 23.7% were recorded as the frequency varied from 0.1 to 1.0 Hz. For L4–L5 an increase of up to 9.69% was noted as cycle frequency increased from 0.1 to 1.0 Hz. An increase of up to 6% was noted in L5–L6.

The changes in the tension thresholds for the stretch as well as release phases of each cycle are as follows: For L3–L4 the tension threshold in the stretch phase demon-

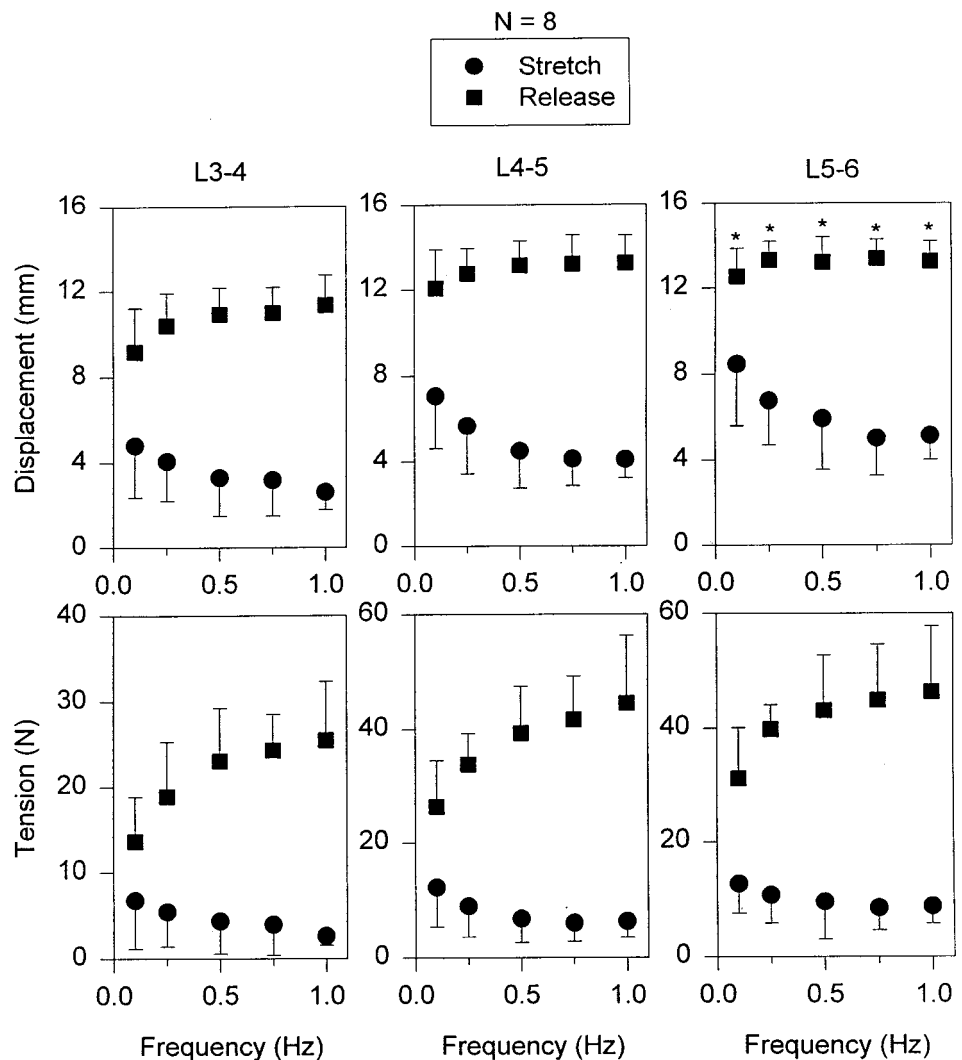


Figure 4. The mean (\pm SD) of the displacement (top row) and tension (bottom row) thresholds in the stretch and release phase are plotted as a function of cycle frequency for the three lumbar levels. Data are pooled from the eight⁸ preparations used in this study.

strated a gradual steady decrease with cycle frequency from 6.75 to 5.42 N (19.8% decrease), 4.37 N (35.3% decrease), 3.95 N (41.4% decrease), and 2.62 N (61.2% decrease) for frequencies of 0.1, 0.25, 0.5, 0.75, and 1.0 Hz, respectively. The tension threshold decreases in the stretch phase for increase in cycle frequency from 0.1 to 1.0 Hz in L4–L5 were from 12.3 to 8.99 N (26.7% decrease), 6.77 N (44.8% decrease), 6.0 N (51.1% decrease), and 6.31 N (48.6% decrease). For L5–L6 the decrease was from 12.7 to 10.7 N (–15.4%), 9.58 N (24.6%), 8.5 N (33.1%), and 8.77 N (31%).

Increase in the tension threshold during the release phase for the different cycle frequencies in L3–L4 was from 13.6 at 0.1 Hz to 18.9 N (38.8% increase), 23 N (69.4% increase), 24.3 N (78.8% increase), and 25.5 N (87.5% increase) as frequency increased to 0.25, 0.5, 0.75, and 1.0 Hz, respectively. At L4–L5 the tension threshold increased from 26.4 N at 0.1 Hz to 33.8 N (28.2% increase), 39.3 N (49% increase), 41.7 N (58% increase), and 44.5 N (68.7% increase) for frequencies of 0.25, 0.5, 0.75, and 1.0 Hz, respectively. At L5–L6 the tension threshold in the release phase of the cycle increased from 31.1 N at 0.1 Hz to 39.8 N (27.8%

increase), 43.1 N (38.4% increase), 44.9 N (44.3% increase), and 46.4 N (49% increase) as the cycle frequency increased to 0.25, 0.5, 0.75, and 1.0 Hz, respectively.

The peak normalized MAV data of each cycle frequency at each lumbar level are summarized in Table 2 and presented graphically in Figure 5. At L3–L4 the mean peak normalized MAV gradually increased from 1.0 to 1.6 (60% increase) as the cycle frequency increased from 0.1 to 1.0 Hz. In L4–L6 the mean peak normalized MAV increased from 1.0 to 2.07 (100% increase) for the cycle frequency range of 0.1 to 1.0 Hz. In L5–L6 it increased from 1.0 to 2.16 (116% increase) as the cycle frequency increased from 0.1 to 1.0 Hz. Overall, the increase in the EMG discharge for increasing cycle frequency indicates that stronger muscular contraction, and therefore more multifidus force, may have been associated with faster flexion of the lumbar spine.

The *t* tests comparing the thresholds of the stretch phase to the thresholds of the release phase revealed significant differences ($P < 0.01$) for both displacement and tension, with the stretch phase thresholds lower than the release phase thresholds in every case. The one-way anal-

Table 1. Mean and Standard Deviation of Displacement and Tension Thresholds During the Stretch and Release Phases of the Different Cycle Frequencies for the Three Lumbar Levels*

		L3/4				
Frequency (Hz)		0.1	0.25	0.5	0.75	1.0
Displacement threshold (mm)	Stretch	4.82 ± 2.45	4.08 ± 1.88 (−15.4%)	3.31 ± 1.83 (−31.4%)	3.20 ± 1.70 (−33.5%)	2.62 ± 0.83 (−45.6%)
	Release	9.18 ± 2.04	10.4 ± 1.51 (+13.3%)	10.9 ± 1.24 (+19.1%)	11.0 ± 1.19 (+19.9%)	11.4 ± 1.41 (+23.7%)
Tension threshold (N)	Stretch	6.75 ± 5.59	5.42 ± 4.00 (−19.8%)	4.37 ± 3.82 (−35.3%)	3.95 ± 3.56 (−41.4%)	2.62 ± 1.07 (−61.2%)
	Release	13.6 ± 5.25	18.9 ± 6.40 (+38.8%)	23.0 ± 6.18 (+69.4%)	24.3 ± 4.23 (+78.8%)	25.5 ± 6.86 (+87.5%)
		L4/5				
Frequency (Hz)		0.1	0.25	0.5	0.75	1.0
Displacement threshold (mm)	Stretch	7.07 ± 2.45	5.69 ± 2.26 (−19.6%)	4.50 ± 1.76 (−36.3%)	4.12 ± 1.27 (−41.7%)	4.10 ± 0.88 (−42.0%)
	Release	12.1 ± 1.82	12.8 ± 1.18 (+5.75%)	13.1 ± 1.15 (+8.88%)	13.2 ± 1.36 (+9.42%)	13.3 ± 1.31 (+9.69%)
Tension threshold (N)	Stretch	12.3 ± 6.90	8.99 ± 5.35 (−26.7%)	6.77 ± 4.10 (−44.8%)	6.00 ± 3.18 (−51.1%)	6.31 ± 2.79 (−48.6%)
	Release	26.4 ± 8.16	33.8 ± 5.43 (+28.2%)	39.3 ± 8.21 (+49.0%)	41.7 ± 7.61 (+58.0%)	44.5 ± 11.8 (+68.7%)
		L5/6				
Frequency (Hz)		0.1	0.25	0.5	0.75	1.0
Displacement threshold (mm)	Stretch	8.47 ± 2.87	6.77 ± 2.07 (−20.1%)	5.93 ± 2.39 (−29.9%)	5.01 ± 1.75 (−40.8%)	5.13 ± 1.13 (−39.4%)
	Release	12.5 ± 1.32	13.3 ± 0.90 (+6.06%)	13.2 ± 1.19 (+5.29%)	13.4 ± 0.91 (+6.68%)	13.2 ± 0.96 (+5.58%)
Tension threshold (N)	Stretch	12.7 ± 5.16	10.7 ± 4.95 (−15.4%)	9.58 ± 6.57 (−24.6%)	8.50 ± 3.84 (−33.1%)	8.77 ± 3.02 (−31.0%)
	Release	31.1 ± 8.95	39.8 ± 4.26 (+27.8%)	43.1 ± 9.69 (+38.4%)	44.9 ± 9.71 (+44.3%)	46.4 ± 11.4 (+49.0%)

* Percent increase/decrease of the thresholds relative to the thresholds at 0.1 Hz is given in the parentheses.

yses of variance indicate that there are significant effects of frequency on the stretch phase thresholds for both displacement and tension in the EMG of all the lumbar levels (in all cases, $P < 0.006$). In all cases the trends were for the threshold to decrease with increasing frequency. For the release phase thresholds, a significant effect of frequency was found in all tension thresholds, as well as in the L3–L4 and L4–L5 displacement thresholds. The only exception was the L5–L6 displacement thresholds in the release phase, which showed a trend but not a significant effect between the threshold and frequency ($P = 0.0929$). In all cases the threshold in the release phase increased with increasing frequency. Table 3 summarizes the statistical analysis.

For the EMG peak mean absolute values, all three lumbar levels exhibited significant effects of frequency ($P < 0.05$) with trends of increasing peak amplitude with increasing frequency as shown in Figure 5.

Discussion

The major findings of this investigation are that differences in the NNZ exist between the stretch and the re-

lease phases of a cyclic lumbar flexion and that such NNZs are further modified by the frequency of the flexion–extension cycle. The data obtained indicate that reflexive muscular forces are available progressively earlier in the stretch phase to stabilize the intervertebral joints at higher flexion frequencies. A progressively larger window of high risk, however, may develop in the release phase in fast cyclic motion, increasing the exposure to instability and injury. It was also found that faster flexion–extension cycles result in larger peak EMG discharge, indicating that more reflexive muscle force was present.

It should be recalled, right from the start, that the EMG recorded from the multifidus muscles was reflexively elicited (compared with voluntary activation). It should also be recalled that this reflexive system is a sensorimotor feedback of the spine. As in any feedback system, its dynamic response is a compounded behavior of its individual components shown in the simplified schematic of Figure 6. In this case the feedback system consists of at least four types of mechanoreceptors embedded in the viscoelastic structures, sensory nerves to the spinal cord, spinal interneuronal network, and the

Table 2. Mean and Standard Deviation of Normalized Peak Mean Absolute Value (MAV) of the EMG Recorded During the Different Cycle Frequencies at the Three Lumbar Levels

Frequency (Hz)		0.1	0.25	0.5	0.75	1.0
MAV	L3/4	1.000 ± 0.000	1.081 ± 0.217	1.271 ± 0.480	1.462 ± 0.811	1.605 ± 1.087
MAV	L4/5	1.000 ± 0.000	1.347 ± 0.466	1.750 ± 0.955	1.729 ± 0.884	2.070 ± 1.227
MAV	L5/6	1.000 ± 0.000	1.389 ± 0.488	1.736 ± 0.669	2.054 ± 1.062	2.160 ± 0.8365

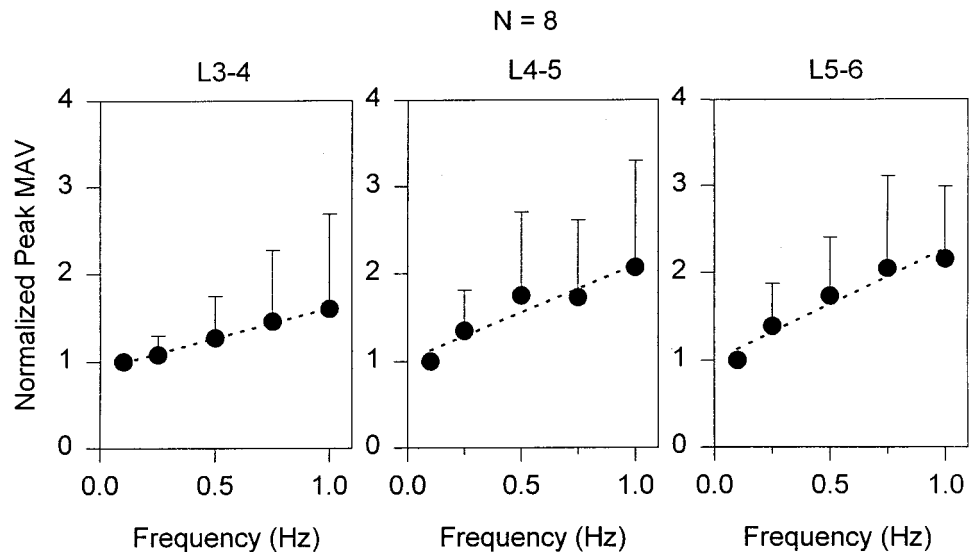


Figure 5. The mean (\pm SD) of the peak MAV of the EMG is shown as a function of frequency, demonstrating that progressively stronger muscle contraction was associated with increasing cycle frequency.

motor units of the multifidus muscles. There are also the ascending and descending nerve tracks within the spinal cord.⁶ The ascending tracks relay the sensory information to the cerebellum, sensory cortex, and brain stem nuclei. Descending tracks modulate interneurons and motor unit activity as a response. Overall, it is a rather complex multi-input system charged with the management of complex motor performance of routine daily and specialized occupational and sports activities while maintaining stability, preventing tissue damage, and compensating for a spectrum of internal and external disturbances (*e.g.*, changes in the gravity vector, velocity, and acceleration of a body mass and load). The interpretation of the data obtained in this study, therefore, should be done while considering the system perspectives given above and accounting for each of the components.

Mechanoreceptor Dynamics

The neurophysiologic literature is replete with data describing the relationships of a stimulus *versus* neural activity of different types of afferents. Afferents were shown to be sensitive to various stimuli, such as force, elongation, pressure, vibration, joint angle, and velocity.²⁸ Receptors require a sufficient stimulus intensity to trigger neural activity. This stimulus threshold may or may not be the same during an increase followed by a decrease in stimulus intensity.²⁸ This behavior is noted as one pos-

sible source of the large changes of the tension-displacement threshold of EMG activity in the stretch and release phases of the flexion cycle.

Furthermore, the available data also confirm that some mechanoreceptors respond to stimulus onset with high frequency discharge, frequency increase, and peak discharge frequency, which are functions of the rate of stimulus application and its final intensity.²⁸ This behavior can also explain, in part, the dependence of the EMG tension-displacement thresholds in the stretch and release phases on the cyclic frequency input.

Finally, it was anticipated that the specificity of mechanoreceptors to tension or displacement stimulus could be revealed when the EMG thresholds were superimposed on the displacement-tension hysteresis curve. This did not materialize (see below) as both the tension and displacement thresholds increased in the release phase of the cycle. This implies that some other mechanism was involved.

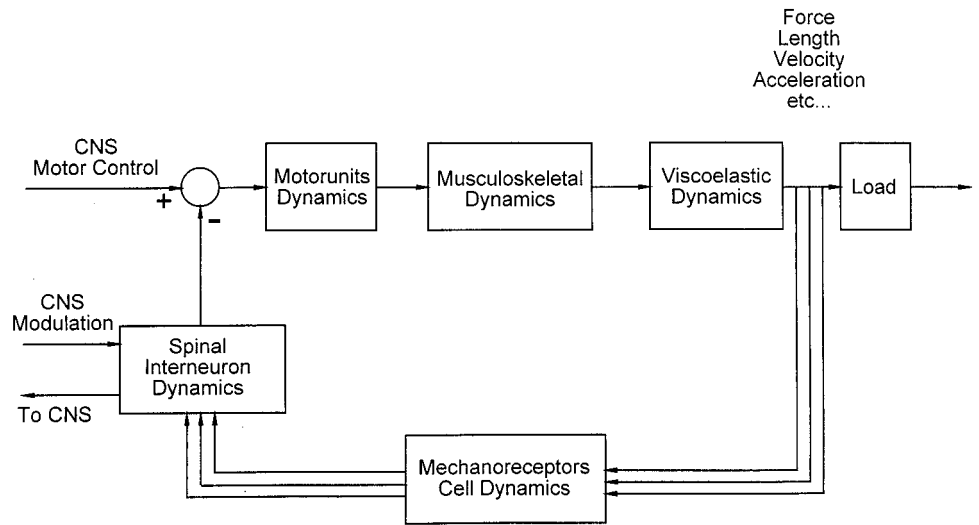
Viscoelastic Tissue Dynamics

The hysteresis in the tension *versus* displacement curve associated with a stretch-release cycle applied to the viscoelastic tissues of the ligaments, discs, and capsules of the lumbar spine were expected, as they represent the classic behavior of such materials. The afferent dynamics are expected to be strongly influenced by the viscoelastic

Table 3. Results of One-Way Analysis of Variance With Repeated Measures of Parameters with Frequency as Repeated Measure Variable

Lumbar Level	Stretch		Release	
	Tension	Displacement	Tension	Displacement
L3-4	f = 4.59 p = 0.006	f = 8.29 p = 0.0002	f = 6.79 p = 0.001	f = 3.92 p = 0.012
L4-5	f = 8.97 p = 0.0001	f = 14.51 p = 0.0001	f = 9.14 p = 0.0001	f = 3.75 p = 0.0146
L5-6	f = 5.68 p = 0.0023	f = 6.73 p = 0.0009	f = 8.41 p = 0.0001	f = 2.22 p = 0.929

Figure 6. A simplified control system of the spine showing the major anatomic structures associated with the forward and feedback loops, as well as spinal and supraspinal components. The three parallel lines in the feedback loop emphasize that at least three inputs (from ligaments, discs, and capsules) are stimulating afferents to activate spinal interneuronal networks.



properties, as the tension or elongation measured is developed by and within these structures. The hysteresis, however, was expected to shift the displacement threshold of the release phase to the right if the afferents were tension specific, *e.g.*, initiate and terminate activity at the same tension (Figure 7). Similarly, if the afferents were length specific, the tension threshold was expected to be shifted downward, in the release phase. Furthermore,

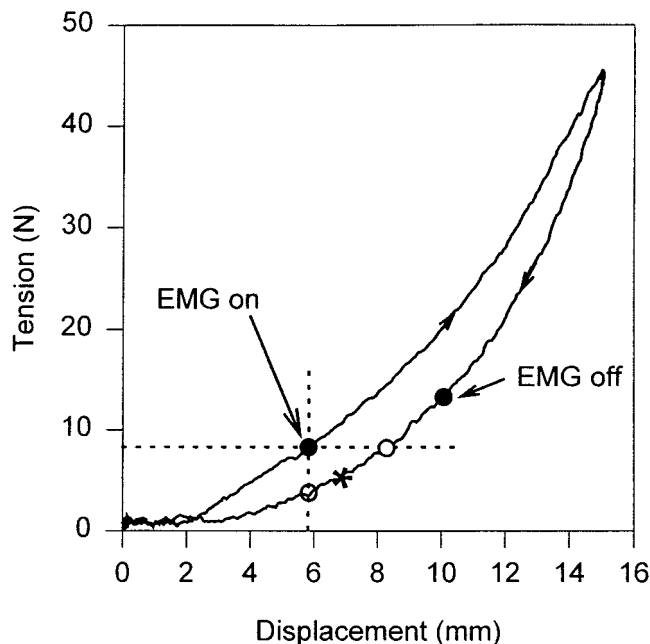


Figure 7. A schematic of a hysteresis curve of a single cycle is shown. Filled circles indicate the actual location of the initiation and termination of EMG (*e.g.*, stretch and release thresholds); open circles represent the anticipated release threshold if the afferents were tension- or displacement-sensitive and initiate or terminate neural activity at the exact tension or displacement threshold. Stars indicate the range in the release cycle where EMG activity would be anticipated to terminate if afferents were sensitive to both tension and displacement. The actual release threshold is well above the anticipated range, pointing out that other processes or structures determine the actual thresholds.

assuming specificity of some afferents to tension and others to length, their overall response will cause a threshold shift to the right and down in the release phase of the cycle. The thresholds shifted to the right and upwards, not supporting this hypothesis. The fact that both the tension and displacement thresholds increased in the release phase indicates that the hysteresis in the viscoelastic structures may not be the only factor in the difference of the NNZ of the stretch and release phases. The frequency at which the flexion-extension cycle was applied seems to be another factor that influences the NNZ variability.

Motor Unit Dynamics

A motor unit is composed of a motor neuron in the anterior horn of the spinal cord and a motor axon that innervates many muscle fibers. Several delays are associated with action potential conduction velocity in the axon, neuromuscular junction transmission, conduction velocity of action potential in the muscle fibers, and excitation-contraction coupling. However, of importance is what was recorded and what was assumed. In the paradigm of this experiment the EMG was assumed, as in many biomechanical studies, to represent force. It must be noted that the EMG is a fast electrical event that occurs before the force, a slower mechanical event, fully builds up. The pure time delay between the appearance of increasing EMG and increasing force is approximately 10–25 msec. The rate of buildup of force is also a function of the predominant fiber composition of the muscle under investigation. The paraspinal muscles are mostly mixed or slow twitch,¹⁷ and the buildup of force and its delay are slower than other fast twitch muscles. Applying the physiologic delays to the hysteresis curve provides a new insight. The tension-displacement thresholds of the stretch phase shown in Figure 8 should be shifted upward and to the right to account for the time delay between the EMG and the associated muscle force. Similarly, in the release phase the thresholds should be decreased to lower tension and displacement points to

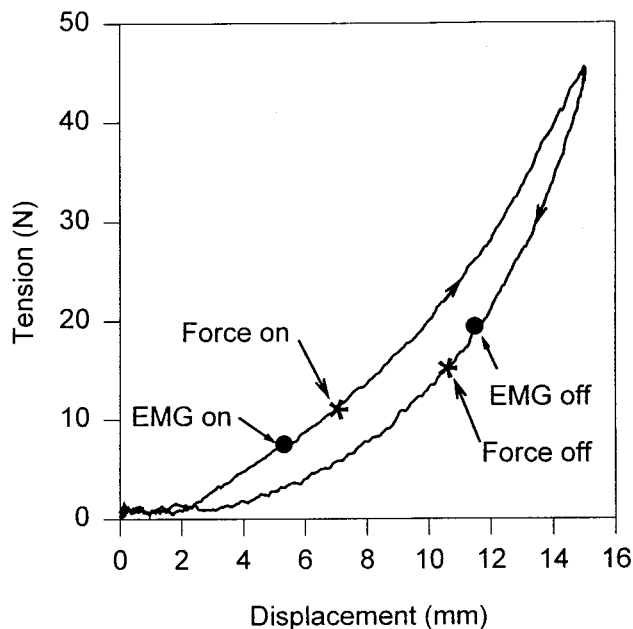


Figure 8. Filled circles superimposed on a schematic of tension versus displacement hysteresis curve indicate the location of the stretch and release thresholds as calculated from the EMG, whereas the stars indicate the location where force was actually developed or terminated, accounting for the physiologic delays between EMG and force.

represent the exact location where the reflexive muscle force actually diminished to zero.

Once this issue is taken into consideration, the differences between the thresholds in the stretch and release phases could be substantially smaller but not fully eliminated. The dependence of the thresholds on cycle frequency also indicates that other factors influence the NNZ differences.

Spinal Interneuronal Dynamics

Information relayed to the spinal cord from peripheral afferents is transmitted monosynaptically or polysynaptically before reflexive activity is initiated in a motor unit. Monosynaptic transmission directly from the afferent cell to the motor unit is well established for the muscle spindle. Most other afferents are reflexively linked to motor units *via* at least one and mostly many interneurons. A great deal of summation and integration of sensory information take place in spinal interneurons. Furthermore, the excitatory level of interneurons is also controlled and modulated by higher brain centers such that adaptation, accommodation, inhibition, anticipatory preparation, and reflex modulation for diverse types of movements are possible.^{6,28}

The adaptive properties of spinal neural networks in regulating and compensating muscular co-activity for various internal and external changes, such as movement velocity, contraction type, gravity vector orientation, skill acquisition, load levels, and even some pathologic conditions, were reported.^{1-5,10,35} The work of Hagood et al¹⁰ assessing the effect of movement velocity on co-

activity patterns is of particular interest. Increasing movement velocity was shown to result in changes in the timing and magnitude of EMG co-activity.

It is reasonable to assume, based on anatomic, physiologic, and biomechanical knowledge, that spinal neural networks are most likely one of the major sources of the changes or differences in the multifidus activity thresholds in the stretch and release phases of a flexion-extension cycle of different frequencies.

The impact of the velocity of spinal motion on increase in internal loads, change of muscle recruitment patterns, and time to peak activities and viscoelastic responses was discussed in the literature.^{19-22,29,31,40} The overwhelming conclusions of such studies point out that increasing velocity results in an increase in antagonist co-activity, larger internal loads, larger compressive forces, and increased stress in the viscoelastic tissues. It was also pointed out that higher movement velocities pose a risk factor for low back disorders. The data presented in this article confirm the results of the above studies and also point out that the physiology attempts to compensate for the increased risk imposed by increased velocities yet fails to fully cover the release phase of a flexion cycle. A larger window of exposure may be created in the release phase with increasing velocity of movement.

From another standpoint Pope et al³² showed that exposure to whole body vibrations at 4–5 Hz also had an impact on muscular responses. The EMG response to sudden loads was stronger but associated with larger latency, increasing the window of exposure to instability.

It should be also recalled that the data presented in this article were collected from the feline model, a quadruped. Although any comparison to data collected from humans might be viewed as speculative, it should be emphasized that neurologically the feline has long been used as a close replica to that of the human. Muscle, ligaments, and reflexive functions were shown to be similar in general responses. A major difference is that the feline is perpendicular to the gravity vector whereas the human is parallel to it. This difference, however, is well taken by the adjustment in the kinematics of a model. What is important, therefore, is that the *modus operandi* of the NNZ is probably similar with the appropriate calibration or scaling for size.

Overall, the activation and cessation thresholds of reflexive muscular activity are not identical. Viscoelastic hysteresis seems to be one factor that may be responsible for such variability, whereas the frequency at which the movement cycle is applied seems to be another. The dynamics of various afferents and their sensitivity to rate of stimulus application as well as sensitivity of spinal interneuronal network to the velocity of motion are the physiologic processes that modify the overall muscular activity in response to the conditions applied.

The new information provided in this report could be valuable in assessment of high-risk occupational activi-

ties when incorporated in dynamic spinal models.^{7,18,24,27,33} Such models need to incorporate the NNZ described in this article and the changes in the NNZ associated with a range of movement velocities. Assessment of windows of risk could be improved when predicting the lack of muscular activity in the flexion and extension phases relative to the motion velocity, applied load, disc pressure, and ligamentous strains. Properly calibrating the NNZs associated with muscular activity or co-activity together with the dynamics and kinematics of the various structures could significantly improve the insights available for detection of movements associated with exposure to instability and possible injury.

■ Conclusion

The general response seen in this investigation indicates that for passive cyclic flexion–extension of increasing frequency, EMG activity is triggered earlier in the stretch phase of the cycle. The early onset of reflexive muscular forces provides stability earlier in the flexion phase and lends an increased measure against instability and potential injury. During the extension (release) phase of the cycle, however, EMG activity terminates earlier and the threshold of activity is further elevated with increased cycle frequency. The extension movement is therefore performed for a substantial duration without stabilizing muscular forces. Fast extension may therefore be associated with increased exposure to instability and injury. The high-risk window during extension is present during slow extension but increases with faster movement. Faster cyclic flexion–extension is also associated with an increased level of reflexive muscular activity as the physiology attempts to stabilize the intervertebral joints, although at the expense of larger forces. Overall, the findings confirm that fast motion of the spine is associated with increased risk of instability and potential injury and with the development of larger forces within the spine. A simple conclusion suggests that occupational activities should be designed such that fast movements are minimized as much as possible. This is especially important in extension, where less protection exists even in slow motion.

■ Key Points

- For a flexion–extension cycle, reflexive EMG discharge was initiated at lower supraspinous ligament tension and displacement thresholds in the flexion phase relative to the same thresholds in the extension phase when the EMG terminated.
- Faster flexion–extension cycles result in lower EMG activation threshold in flexion and in higher EMG cessation thresholds in extension.
- Faster flexion–extension cycles also result in higher peak EMG activity.

- The velocity of the movement was shown to be a significant factor during the extension part of the motion.
- The viscoelastic hysteresis of ligaments, disc, and capsular tissues associated with such a flexion–extension cycle, as well as spinal interneuronal activity, are thought to be the other major factors causing the above behavior.

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