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In Vivo Transient Vibration Assessment of the Normal Human Thoracolumbar Spine

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ABSTRACT

Objective: The objective of this study was to quantify the mobility characteristics (dynamic stiffness and mechanical impedance) of the normal human thoracolumbar spine with a transient vibration analysis technique.

Design: This study is a prospective clinical investigation to obtain normative biomechanical data from the human male and female spine *in vivo*.

Setting: Musculoskeletal research laboratory, university setting.

Subjects: Twenty asymptomatic subjects (age range, 20-60 years) with no recent history of musculoskeletal complaints.

Main Outcome Measures: Mechanical impedance, effective stiffness, and resonant frequency analyses were used to quantify the dynamic stiffness of the thoracolumbar spine in this subject population. Data were obtained from posteroanterior mechanical thrusts delivered with an activator adjusting instrument equipped with a load cell and accelerometer by means of a portable computer.

Results: In response to the activator adjusting instrument thrusts, the thoracolumbar spine typically exhibited an impedance minimum at frequencies ranging between 30 and 50 Hz. The maximum posteroanterior impedance and corresponding maximum effective stiffness of the thoracolumbar spine and sacrum was roughly 2 to 8 times greater than the magnitude of the impedance minimum. Statistically significant differences in mobility between male and female subjects were noted, particularly for frequencies corresponding to the maximum mobility (40 Hz) and minimum mobility (10-20 Hz, 70-80 Hz). For most subjects (both male and female), the lumbar region exhibited a higher impedance and stiffness (less mobility) when compared with the thoracic region.

Conclusions: The posteroanterior mechanical behavior of the human thoracolumbar spine was found to be sensitive to mechanical stimulus frequency and showed significant region-specific and gender differences. In the frequency range of 30 to 50 Hz, the lumbar spine of this subject population is the least stiff and therefore has the greatest mobility. From a biomechanical point-of-view, the results of this study indicate that dynamic spinal manipulative therapy procedures will produce more spinal motion for a given force, particularly when the posteroanterior manipulative thrust is delivered in frequency ranges at or near the resonant frequency. In this regard, spinal manipulative therapy procedures designed to target the resonant frequency of the spine require less force application. Both magnitude and frequency content of manual and mechanical thrusting manipulations may be critical elements for therapeutic outcome. (J Manipulative Physiol Ther 2000;23:521-30)

Key Indexing Terms: Biomechanics; Manipulation-Chiropractic; Mechanical Impedance; Spine; Stiffness

INTRODUCTION

The spinal column combines an intricate architectural arrangement of bone, muscle, and soft tissue to form a structure of mechanical and physiologic significance. Not only does the spinal column serve to protect the spinal cord, but it also transmits, attenuates, and distributes the static (invariable time) and dynamic (variable time) forces associated with daily activities.

Although the spinal column provides the structures for load transmission and attenuation, the pathways for load transmission and attenuation may be greatly altered during voluntary (postural changes) and involuntary (fatigue) activities, which produces unstable and pathologic changes to the kinematic behavior of the spinal column. Segmental instability and pathologic features of the spine are believed to produce abnormal patterns of motion and forces that may play a significant role in the cause of low back pain (LBP).[1] The ability to quantify *in vivo* spine segment motion in response to forces is thus considered to be of clinical significance in terms of both diagnosis and treatment of spinal disorders, including back pain.

Knowledge of spine segment motion patterns and forces is of fundamental interest to understanding the postural, time-dependent, and dynamic response of the spine, the role

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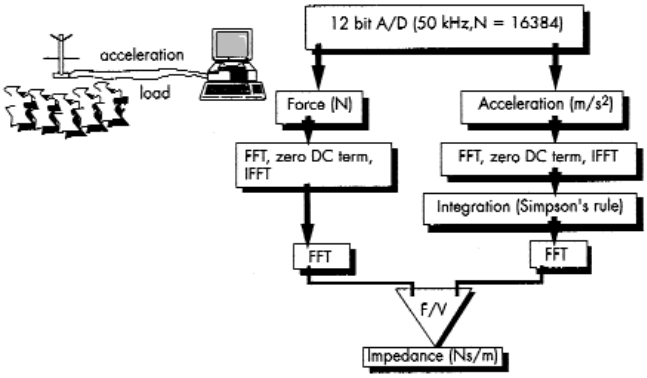

Fig 1. *Schematic diagram of the transient vibration assessment protocol used to determine the driving point impedance and effective stiffness spectra of the human spine.* IFFT, *Inverse Fast Fourier Transformation.*

of spinal implants in mechanical load sharing, and the response of the spine to externally applied forces (such as chiropractic spinal manipulation). Although chiropractic treatments are generally considered to be therapeutic, little is understood about the source of the positive treatment effects that are associated with manipulation and mobilization therapy.[2-5] Possible therapeutic mechanisms that have been postulated include realignment of vertebral bodies, mobilization of spinal joints, relaxation of back musculature through reflex pathways, production of a respiratory burst, cavitation that causes a temporary increase in joint space, release of inflammatory agents within the body, and coactivation of mechanically sensitive somatic afferents.[6-10]

To understand the biomechanical consequences of spinal manipulative therapy (SMT) more fully, therefore, a number of chiropractic researchers are focusing on quantifying the applied forces that are associated with spinal manipulation and the mechanical response of the spine to these forces.[2,5,11,12] In commonly used clinical procedures such as posteroanterior spinal manipulation, spinal stiffness or the load/displacement response is often of interest. Although only preliminary studies have appeared that have investigated the relationship between pain and posteroanterior stiffness[13,14] and the effect of treatment on spinal stiffness, a common recommendation is that a clinician take note of the stiffness[15] felt during mobilization palpation procedures.[8,16-18] In principle, an unstable segment should exhibit increased displacement or decreased stiffness, and a stiffened segment should exhibit decreased displacement, compared with adjacent segments.[19] The displacement of the vertebra and the resistance to the displacement during mobilization therefore is potentially useful in spinal diagnosis and for the establishment of effective treatment protocols. Although manual tissue compliance assessments are commonly used during mobilization or motion palpation procedures to formulate clinical diagnoses and treatment regimens, most compliance assessments are based on qualitative assessment strategies. A number of studies have shown that clinical judgments that are based on qualitative compliance assessments are unreliable or inaccurate.[20-23]

Direct estimates of the dynamic *in vivo* force versus displacement or stiffness characteristics of the spine are exceedingly difficult because of problems associated with obtaining precise measurements of spine segment and intersegment displacements during SMT.[24] Efforts to quantify the *in vivo* stiffness behavior of spinal segments during manipulation and low-frequency mobilization have included the use of surface displacement transducers,[8,10,25] force reaction methods,[24] and mathematic models.[26,27] The results of these studies indicate that the normal spine exhibits a posteroanterior displacement of about 2 to 10 mm per 100 N of applied load, corresponding to a posteroanterior stiffness of about 10 to 50 kN/m. Because the musculoskeletal structures in the spine are viscoelastic in nature, however, one would predict that the mechanical (motion) and physiologic (muscle activation, coactivation of mechanically sensitive somatic afferents) response to manipulation, and mobilization would depend on loading rate and frequency. With regards to the latter, Herzog et al[2,28-30] have shown that the neuromuscular response of the paraspinal muscles is greatest when thrusts are applied dynamically rather than statically. These authors further noted that neuromuscular responses associated with SMT have not been found to result from joint cavitation as previously hypothesized.[28-30]

If the spine is truly a viscoelastic structure, then increased mobility (motion response) will occur when manipulation or mobilization therapy is applied at certain loading rates and/or frequencies. There is a considerable amount of literature pertaining to the cyclic and transient vibration behavior of the spine during axial loading,[31,32] but very few experimental measurements of the frequency dependent mechanical response of the spine to posteroanterior manipulations or mobilizations exist in the literature. In recent years, dynamic mechanical devices have been developed and used to more objectively and more precisely quantify spine posteroanterior forces and displacements.[11,16,33-35] Using a spinal physiotherapy simulator, Lee and Svensson[36] observed that the low-frequency posteroanterior stiffness of the lumbar-sacral spine was about 25% to 50% greater for cyclic mobilization at 0.5 to 1.0 Hz in comparison with mobilization at quasi-static frequencies (0.05 Hz). Using an activator-adjusting instrument (AAI; Activator Methods International, Ltd, Phoenix, Ariz) that is equipped with an impedance head,[11] Nathan and Keller[37] determined the dynamic posteroanterior stiffness response of the human thoracolumbar spine in 3 human subjects. Their transient vibration analysis indicated that the stiffness of the spine varied 2-8-fold. Aside from these few studies, however, there are no comprehensive data concerning the frequency-dependent posteroanterior mechanical response of human spine.

Given the aforementioned frequency dependency that is associated with the physiologic and mechanical response of the spine, a more precise understanding of the normal dynamic posteroanterior mechanical behavior of the spine is an important first step to the understanding of the therapeutic benefits derived from SMT. The objective of this study was to quantify the dynamic posteroanterior stiffness char-

Table 1. *Summary of the normal subject demographic data*

Group	Subjects (n)	Mean age (y)	Mean weight (kg)	Mean height (cm)
Younger female	5	23.8	57	166
Younger male	5	26.8	81	178
Older female	5	56.4	66	167
Older male	5	57.7	79	181

acteristics of the normal human lumbar spine with the use of a transient vibration analysis technique. We hypothesized that the normal spine would show frequency-specific variations in mechanical stiffness.

METHODS

Twenty asymptomatic subjects were recruited to participate in this study. Subjects were excluded if they reported having disabling musculoskeletal pain in the past year or had consulted a health care professional previously for spine-related complaints, including back, neck, or related extremity pain. Subject recruitment and testing followed the human subjects committee institutional review board guidelines of a public university. Subjects ranged in age from 20 to 60 years and were uniformly stratified in terms of gender (10 male, 10 female) and age (Table 1).

An AAI was used to deliver posteroanterior thrusts to the spinous processes (L1-L5) of each subject. Posteroanterior thrusts were applied directly over the spinous processes, with the subjects lying prone on a padded table. The AAI I device delivers a very short duration (<5 msec) force-time profile to the spine that is characterized by a very high loading rate (approximately 250 kN/sec), haversine-like force-time pulse. This force-time profile results in a peak force of about 140 N and is associated with a relatively uniform load amplitude versus frequency spectrum.[11] A uniform load-frequency spectrum is a desirable feature for transient vibration analyses because this will impart an equal amount of load-signal intensity over the frequency range of interest.[11]

To quantify the input force and motion response of the spine, the AAI was equipped with an impedance head, comprised of a load cell and an accelerometer. The load cell and accelerometer measure the input force and acceleration response characteristics, respectively, of the spine at a common point. The impedance head consisted of a 2200 N quartz force ring (PCB model 201A03; PCB Piezotronics, Buffalo, NY) and a miniature 5000-g quartz accelerometer (PCB model 305A04; PCB Piezotronics) attached to an aluminum housing that is fixed to the end of the stylus. The force ring is light weight (14 g) and has a load resolution of 0.04 N, a resonant frequency of 70 kHz, and a low-frequency response of 0.0003 Hz. The accelerometer is light weight (4.5 g) and has an acceleration resolution of 1 g, a resonant frequency of 60 kHz, and a low-frequency response of 0.25 Hz. Consequently, the useful frequency range of the load and acceleration transducers is above (0.25 Hz) and below resonance (60 kHz). The force ring and accelerometer are powered by conventional charge amplifiers. The total weight of the impedance head assembly is less than 50 g.

Five repeated transient thrusts were delivered in the posteroanterior direction to 4 thoracic spinous processes (T7, T9, T11, T12) and 5 lumbar spinous processes (L1-L5). Thrusts were always applied at the end of the expiration of breath. The force and acceleration signals were sampled at 50 kHz (20 μs/sample), with a 12-bit analog-to-digital converter connected to a computer (Macintosh; Apple Computers, Inc, Cupertino, Calif). The 8192 samples were collected over a 163.8-msec time interval. The effects of any direct current offset in the force and acceleration signals were removed by applying the Fast Fourier Transformation (FFT) to each signal, setting the DC term to zero, and applying the inverse FFT. The acceleration-time signal was integrated with the use of Simpson's rule to derive the thrust response velocity. A force window (50 msec, 2500-samples wide) was then applied to the zeroed force-time signal. An FFT was then applied to the force and velocity signals, and the posteroanterior impedance was calculated at each discrete frequency, as the ratio of FFT(force)/FFT(velocity). Because the velocity response is measured at the point of force application, the derived frequency response function (impedance amplitude versus frequency) is termed *driving-point* impedance. A custom LabView (National Instruments, Austin, Tex) program was used to process and analyze the force/velocity versus frequency data. This analytic procedure is summarized in Fig 1 and is described in more detail in a previously published validation study.[11]

Ensemble-averaged impedance and phase plots were derived from the 5 transient vibration-response measurements that were obtained at each spinal level. Note that the data conversion procedure that was used resulted in a frequency resolution of approximately 6 Hz (50,000 Hz/8192 samples). Three mechanical impedance parameters were identified from each impedance versus frequency curve: (1) first peak (peak1), (2) first minimum (min1 or max mobility), and (3) second peak (peak2); together with the frequency (f) at which each occurred (Fig 2). The impedance frequency peaks (peak1, peak2) and valley (min1) were easily identified from the impedance versus frequency curves and were always associated with a rapid change in phase angle. On the basis of the impedance versus frequency-response curves, the posteroanterior effective dynamic stiffness was calculated as the impedance (Z, Ns/m) × the circular frequency ($\omega = 2\pi f$). The effective dynamic stiffness has units of N/m because the circular frequency has units of radians/second, where a radian is a dimensionless quantity.

Descriptive statistics were performed on all of the frequency-response data and a 1-way analysis of variance (ANOVA) was used to determine across-group differences in posteroanterior mechanical impedance, effective dynamic stiffness, and frequency. A paired-observations t-test (POTT) was used to assess within subject region-specific differences.

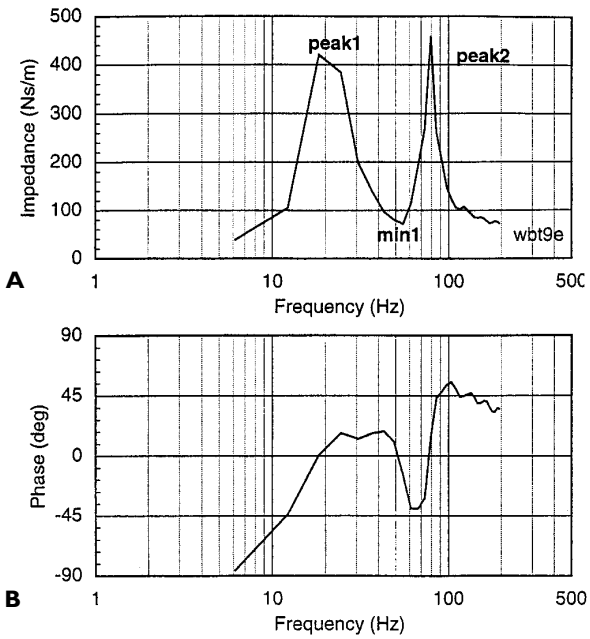

Fig 2. ***A****, Typical driving point impedance (Z, Ns/m) versus frequency (Hz) response curve (log-log scale) obtained for posteroanterior thrusts applied to the spine. Three parameters were identified from each impedance curve: (1) peak1, (2) min1, and (3) peak2. The parameter min1 represents the frequency for which the spine has the greatest mobility (least stiffness). Peak1 and peak2 are representative of frequencies for which the spine has the least mobility (greatest stiffness). Note that there is a large variation (approximately 8-fold) in impedance over the frequency range examined.* ***B****, Corresponding phase versus frequency response curve. Rapid changes in phase were closely associated with the impedance maximum and minimum and were used to verify that the impedance curve parameters (peak1, min1, and peak2) were correctly identified by the LabView program that was used to analyze the impedance response curves. Note that both impedance and phase are plotted by semilogarithmic scales.*

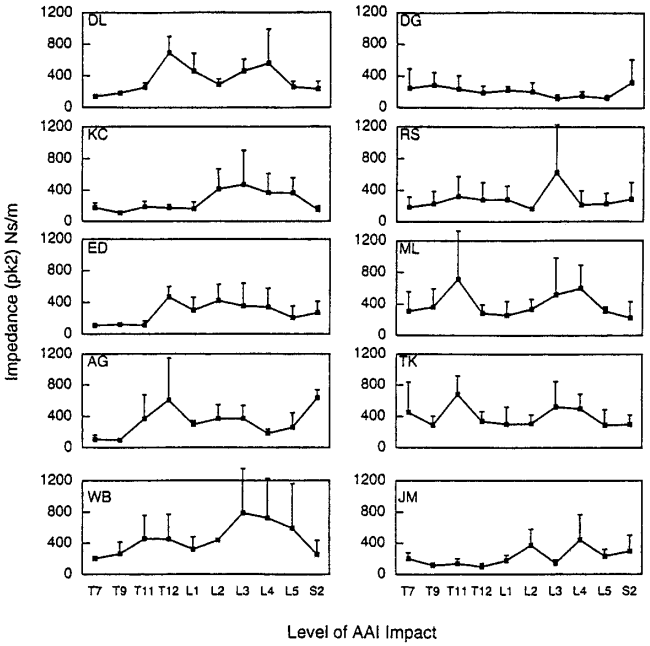

Fig 3. *Region-specific variations in the impedance (peak2, Fig 2) obtained for 5 of the female subjects and for 5 of the male subjects.*

Table 2. *Summary of resonant frequency, mechanical impedance (force/velocity), and effective stiffness (impedance × circular frequency) test results obtained on 20 normal subjects*

Group	Resonant frequency (Hz)	Impedance (Ns/m)	Effective stiffness (kN/m)
Younger female (n = 5)	39 ± 4	75 ± 20	18 ± 5
Younger male (n = 5)	41 ± 5	68 ± 12	18 ± 4
Older female (n = 5)	39 ± 7	83 ± 35	21 ± 11
Older male (n = 5)	38 ± 5	87 ± 29	21 ± 9

Min1 (see Fig 2) mean values ± SD of the combined lumbar levels.

RESULTS

In response to the AAI thrusts, the thoracolumbar spine typically exhibited an impedance minimum at frequencies that ranged between 30 and 50 Hz (min1, Fig 2). In this frequency range, the lumbar spine is the least stiff and therefore has the greatest potential mobility. Henceforth, mobility will be used to refer to impedance and stiffness collectively. The maximum posteroanterior impedance and corresponding maximum effective stiffness of the lumbar spine and sacrum occurred at frequencies between 10 and 20 Hz (peak1, Fig 2) and between 70 and 80 Hz (peak2, Fig 2). The maximum posteroanterior impedance was approximately 2 to 8 times greater than the magnitude of the impedance minimum (50-110 Ns/m). This indicates that the spine is 2 to 8 times stiffer for frequencies that correspond to the impedance maximum in comparison with the impedance minimum. Min1 mobility was significantly lower (ANOVA, $P < .001$) than peak1 and peak2 mobility values.

Fig 3 illustrates the maximum (peak2) impedance obtained for 5 of the female and 5 of the male subjects. The maximum posteroanterior impedance of the thoracolumbar spine ranged from 200 to 800 Ns/m for levels T7-S2. Most subjects (both male and female) exhibited a bimodal mechanical impedance distribution; namely, higher posteroanterior impedance was noted at T11/T12 and L3/L4 when compared with other levels. For the 5 repeated thrusts at each level, the percent coefficient of variation (100 × SD) in the mechanical impedance data ranged from 20% to 25%. Results obtained for the 4 thoracic levels and for the 5 lumbar levels were combined for the group comparisons (male vs female; thoracic vs lumbar).

In both the thoracic and the lumbar regions, the male subjects tended to be stiffer (higher impedance for a given frequency) than the female subjects at the first, lowest frequency peak (peak1, Fig 4) but were less stiff at the second, higher frequency peak (peak2, Fig 4). Statistically significant differences (ANOVA, $P < .05$) in mobility between male and female subjects were noted, particularly for frequencies that corresponded to the maximum mobility (min1 = approximately 40 Hz) and minimum

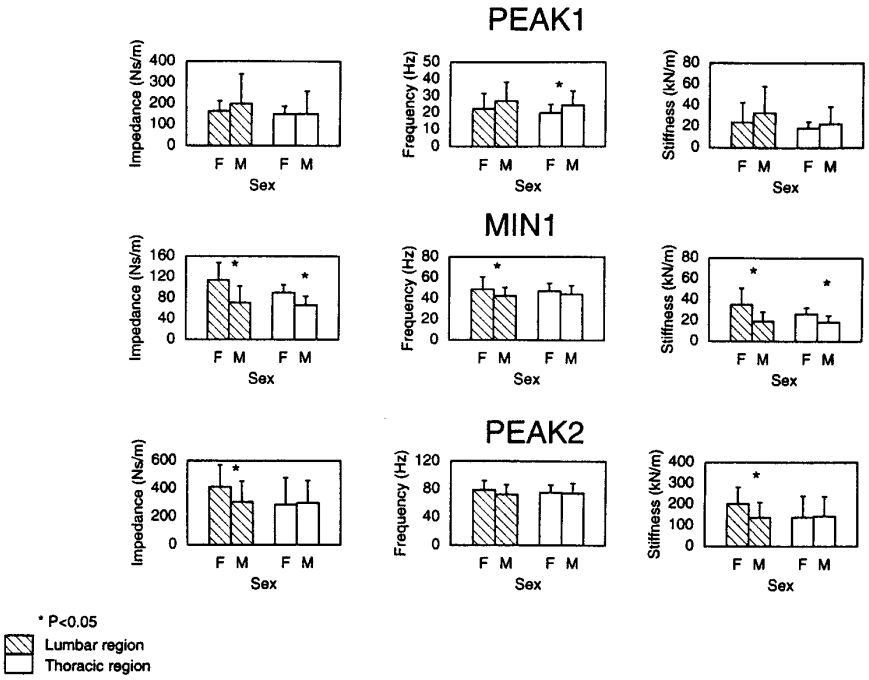

Fig 4. *Differences in spinal impedance (**left column**), frequency (**middle column**) and stiffness (**right column**) for the combined thoracic and lumbar levels of the male (M) and female (F) subjects. Rows correspond to peak1, min1, and peak2 regions of the impedance curves (see Fig 2 for peak1, min1 and peak2 definitions). Statistical significance (ANOVA, P < .05) between male and female groups is indicated by the asterisks (*).*

mobility (peak2 = approximately 75 Hz). Female subjects exhibited greater min1 stiffness (range, 21-42 kN/m) and consequently had lower mobility at all levels in comparison with the male subjects (range, 16-23 kN/m). Female subjects also had significantly higher peak2 stiffness and impedance values in comparison with the male subjects.

As a group (both male and female subjects), the lumbar region exhibited a significantly (POTT, $P < .05$) higher min1 and peak2 impedance and stiffness when compared with the thoracic region. No significant differences in peak1 impedance and stiffness were found between the thoracic and lumbar levels, but the thoracic region was significantly (POTT, $P < .05$) more mobile than the lumbar region at the lowest frequency examined (6 Hz). Older male and female subjects tended to have decreased spinal mobility in comparison with their younger counterparts, but the differences were not statistically significant. Table 2 summarizes the lumbar spine min1 impedance and effective stiffness for the subjects grouped by both age and sex.

DISCUSSION

As with previous biomechanical studies of the vibrational behavior of biologic tissues,[11,38] the basic premise underlying this work is the notion that the intrinsic mechanical behavior of the human spine can be determined by the quantification of the frequency-dependent motion response of various portions of the spine to a known force input. In this regard, identification of the frequency of maximum mobility (natural or resonant frequency) of the spine is considered particularly important, because lower forces can be used to excite a mechanical structure when applied at its natural frequency.[11,38] In the case of the normal human spine that is subjected to posteroanterior mechanical thrusts, we found that the motion response was greatest when excited at approximately 40 Hz.

In this study, a transient vibration analysis of posteroanterior mechanical thrusts delivered to the human spine with the AAI were used to identify several important frequency-dependent parameters of the normal thoracolumbar spine. Most notably, we found that the normal thoracolumbar spine is significantly less stiff or more mobile at frequencies at or near the natural frequency or resonance frequency (approximately 40 Hz). Others have also reported that the posteroanterior stiffness of the thoracolumbosacral spine was dependent on frequency,[16,36] but these studies have described dynamic mechanical responses over a very limited frequency range (typically less than 2 Hz). The current study examined the frequency-dependent stiffness response of the spine over a relatively broad frequency range (results were presented for 6 to 200 Hz). Our results indicate that there is up to an 8-fold variation in the stiffness of the normal male and female spine over this frequency range, for which there is a sharp stiffness valley at resonance. From a biomechanical point-of-view, this indicates that dynamic SMT procedures will produce more motion for a given force input when a significant amount of energy is delivered in frequency ranges at or near the resonant frequency.

For most subjects (both male and female) the lumbar region of the spine exhibited a significantly higher dynamic impedance and stiffness (min1 and peak2) when compared

with the thoracic region. Others report that the thoracic spine is stiffer than the lumbar spine and attribute the increased thoracic stiffness to the rib cage and sternum.[16] Such studies, however, have been restricted to the examination of the low-frequency or quasistatic stiffness characteristics of the spine, whereas the current study considered the dynamic stiffness response at much higher frequencies (≥6 Hz). Other anatomic characteristics may also influence the posteroanterior mechanical response of the spine. In this study, we observed that a number of subjects showed increased stiffness (increased peak2 impedance) at the T11/T12 and at the L3/L4 levels of the spine. In addition to the aforementioned influence of the rib cage and sternum, another possible explanation for the increased stiffness at T11/T12 may be the fact that the spinous processes of T11 and T12 are oriented more posterioanteriorly than the superior thoracic spinous processes. Hence, posteroanterior thrusts delivered at T11 and T12 were more directly aligned with the long axis of the spinous processes than thrusts applied at more cephalad levels of the thoracic spine. Posteroanterior thrusts at T7 and T9 would tend to produce more sagittal plane rotation in comparison with posteroanterior thrusts at T11 and T12. With regards to the increased peak2 stiffness observed at L3/L4, a possible explanation may be the fact that the apex of the lordotic curve in most individuals is at L3/L4. Clearly, the line of drive or force vector of the thrusts, the consistency of the line of drive of thrusts between patients, the degree of lordosis, and other patient-specific anatomic features will have an influence on the transient vibration-analysis measurements.

When we examined the frequency dependency of the thoracolumbar spine, we found that the relative stiffness of different regions of the spine varied with the stimulus frequency. Indeed, although the peak2 (highest frequency peak) and min1 (resonance valley) stiffness values were greater in the lumbar region in comparison with the thoracic region, there were no differences in peak1 stiffness (lower frequency peak) for the thoracic and lumbar regions of the spine. At the lowest frequencies examined (6 Hz), the thoracic region exhibited a significantly higher stiffness in comparison with the lumbar region. Our findings indicate therefore that the dynamic stiffness characteristics of the spine and different anatomic regions of the spine have a much more complex dependency on vibration frequency than previously recognized.

The dynamic posteroanterior frequency-dependent stiffness behavior of the human spine reflects the fact that the spine is a viscoelastic structure, albeit generally more elastic than viscous. Different structures (ligaments, cartilage, bone, tendons, muscle) will exhibit varying degrees of frequency-dependent viscoelastic behavior. Consequently, the overall structural/vibration response of the spine is modulated by both the architecture or the structural organization of component tissues and the load sharing provided by adjacent structures (eg, rib cage, sternum, pelvis). When such factors are combined with other considerations (such as spinal curvature), the net effect is a complex structure-frequency–dependent mechanical behavior. It is not surprising therefore that the viscoelastic mechanical characteristics of the spine will also be influenced by the direction of the applied load. Transient vibration analysis studies conducted on human subjects who were subjected to axial loads indicate that the resonant frequency of the spine is approximately 4 to 6 Hz,[31] which is appreciably lower than the resonant frequency observed for posteroanterior thrusts that were applied to the spine of human subjects lying prone. In general, stiffer structures have higher resonant frequencies (natural frequency) in comparison with less stiff structures. Since the spine is considerably stiffer axially, differences between axial and transverse (posteroanterior) vibration responses may reflect the fact that the loads were applied transversely or in the posteroanterior direction. On the other hand, the motion amplification effect that occurs at resonance appears to be similar for both axial and transverse loading modes. Namely, at resonance there is a 6- to 10-dB (2- to 3-fold) amplification of the mechanical stimulus along the axial spine,[31] which is consistent with the 2- to 8-fold motion amplification observed for mechanical stimulation directed along the posteroanterior direction as reported in the present study.

In addition to loading direction influences on posteroanterior spine stiffness (dynamic or static), the posteroanterior stiffness of the spine is dependent on the applied load magnitude. This occurs because the human spine and other biologic structures exhibit a "bi-phasic" load-deformation behavior. A bi-phasic load-deformation curve is characterized by a nonlinear "toe" region with low stiffness at low loads (large displacements for a small change in load), followed by a linear or "elastic" region characterized by higher stiffness (small displacements for a large change in load). The toe region is usually termed the *physiologic range* in which the tissue normally functions, whereas the elastic region is representative of the reserve strength of the tissue.[39] From a chiropractic point-of-view, the physiologic range is analogous to the active range of motion. The nonlinear, biphasic load–deformation nature of the spine indicates that the posteroanterior stiffness of the spine will change considerably, depending on the load magnitude of the thrust. The mechanical instrument used in this study (AAI) produced peak forces to approximately 140 N, which is less than 10% of the failure load of the human spine subjected to posteroanterior 3-point bending.[40] In this context, the AAI thrust forces applied in this study lie in the physiologic range, and the transient vibration analysis results are only applicable to loading within this range. We are currently conducting transient vibration experiments using an AAI device that is equipped with a preload frame that produces peak forces to approximately 190 N. This load range is probably closer to the elastic region of the load-deformation curve, wherein the spine will be considerably more stiff. Using this device, we expect to find higher dynamic stiffness values for a given frequency.

Soft tissues overlying the spinous processes and muscle and ligamentous structures will also have an influence on the results of the transient vibration analysis. At the present time the precise influence of these tissues are unknown. In gener-

al, skin and soft tissue overlying the spinous process will dampen the vibration response. Namely, the relative difference between the peak1/peak2 and min1 posteroanterior stiffness values (so-called motion amplification response) will be reduced. In the case of subjects who have prominent bony spinous processes, the influence of the overlying skin and fat on motion amplification are probably minimal, particularly because a preload is applied, which compresses the skin and fat tissues during the AAI thrusts. None of the subjects who were examined were excessively overweight or obese. Thus, a relatively solid contact between the AAI rubber tip and the spinous process was obtained in all subjects. In the case of subjects who have a considerable amount of overlying skin and fat, however, the damping effect on motion amplification could be appreciable. Changes in active stabilizing structures (such as muscles [tone]) and passive stabilizing structures (such as ligaments [tension]) could alter the posteroanterior stiffness of the spine appreciably, which in turn may shift the natural frequency of the spine to higher or lower frequencies, depending on the nature of the changes. A "stiffened" spine that results from increased muscle tone or ligament tension would exhibit a shift in the natural frequency to higher frequencies and vice versa. In this study, the transient vibration tests were performed on expiration with the subjects lying in a prone-neutral posture to control the effects of posture, respiration, muscle tone, and ligament tension on the transient vibration analysis results.

In this study, the impedance analysis frequency resolution was 6 Hz; thus our posteroanterior dynamic stiffness assessment was limited to vibration frequencies of 6 Hz and higher. Additional work is needed to more fully characterize the frequency response of the spine to posteroanterior loading. In particular, we would like to extend our transient vibration analysis to include lower frequency components. Because of the extremely short duration force-time characteristics of the AAI (force pulse, <10 msec), accurate representation of the force input and acceleration response of the spine requires data collection at rates of 10,000 samples/second (100 μs/sample) or higher. The frequency resolution of the impedance assessment presented in this study, however, can be improved to 3 Hz by increasing the number of data points collected by a factor of 2. A 1-Hz resolution or better can be realized by collecting 50,000 data points or more, but this also demands significantly more disk storage and data processing requirements.

We also focused our assessment on posteroanterior forces directed to the spinous processes. Posteroanterior thrusts are routinely applied over the transverse processes of the spine, so assessment of the dynamic stiffness response associated with this type of loading should also be of fundamental interest to the chiropractic clinician. Work is currently ongoing to characterize the posteroanterior response associated with the application of dynamic SMT forces over both the spinous and transverse processes of normal and symptomatic subjects.

Biomechanics researchers have long been seeking methods to assess the mechanical behavior of biologic systems in an effort to understand and quantify age-related processes (such as osteoporosis). For example, to directly study the age-related mechanical behavior of long bones in the appendicular and axial skeleton, some authors have reported *in vitro* and *in vivo* vibration assessment methods to determine resonant frequencies,[31,41-51] vibration modes,[46] and mechanical impedance.[11,37,47,48] Results obtained from analyses of modal and resonant frequency, however, require a priori knowledge of the tissue mass and are best suited for *in vitro* and *in situ* studies. On the other hand, results based on the determination of frequency response functions (such as mechanical impedance) provide a direct estimate of *in vivo* tissue mechanical parameters.[38,49] When frequency response function is performed based on assessments of the dynamic mechanical behavior of biologic structures, including the spine, discrete frequencies (sinusoidal oscillations at fixed frequencies) can be and are routinely used. Experiments conducted in this manner, however, are much more time and labor intensive because the mechanical response must be measured at each frequency. One of the key advantages of the transient vibration analysis in this study is the fact that the entire frequency-spectrum of interest can be obtained rapidly and efficiently with the use of a single input stimulus or thrust.

We believe that the dynamic stiffness assessment may have a number of potentially significant clinical ramifications. Frequency response parameters (such as mechanical impedance and effective stiffness) together with the identification of the resonant frequencies associated with the vibration could be used to provide important information concerning the mechanical status of the normal and pathologic spine. For example, when the spine is dynamically loaded along the posteroanterior direction, a lower mechanical impedance value at a given frequency implies that the intervertebral joints are easier to excite and are capable of greater mobility and storage of larger amounts of energy. On the basis of the results of transient vibration analyses, clinicians could prescribe dynamic SMT that maximizes the mechanical response and at the same time minimizes the magnitude of the force delivery to the spine. Indeed, by first probing the spine with a short-duration force thrust and recording the vibration response, it may be possible to selectively adjust or tune the frequency of the mobilization therapy used for a given patient or spinal level. In terms of mechanical factors involved in mobilizing the spine, the results of our study indicate that the normal human lumbar spine has a maximum potential for posteroanterior motion when mechanically stimulated at a frequency of approximately 40 Hz.

A detailed description of the dynamic mechanical status of the spine may also be useful for the identification of the location and precise quantification of the severity of spinal abnormalities. Preliminary results obtained for subjects with known spinal pathologic features[37] and patients who were treated for back pain[13] are encouraging in this regard. Additional research is needed to establish a normative database that includes standardized dynamic biomechanical data for various subject demographic and anthropomorphic factors,

including age, body weight, sex, and other attributes such as handedness and adiposity. Once the normal posteroanterior mechanical frequency dependency has been more fully characterized, changes in the frequency response spectrum may ultimately be used to identify specific underlying tissue disease.

An important issue that has long been of central importance to chiropractic is the source of the positive treatment effects associated with manipulation and mobilization therapy. Manipulation and mobilization therapies produce force-time histories that can range from relatively static to extremely dynamic. It appears that the frequency content associated with mechanical stimulation during SMT has a direct bearing on the subsequent physiologic response of the body. Indeed, there is an increasing body of scientific evidence that indicates that the healing and remodeling response musculoskeletal system, in general, may respond most efficiently to specific frequency components of mechanical signals.[50,51] Remodeling refers to the ability of biologic soft and hard tissues to alter their composition and structure in response to mechanical and systemic stimulation.

Numerous investigators have determined that modulation of the somatosensory system is also mediated in part by the frequency content of the signal input.[52-58] Although little work has been performed to investigate the relationships between frequency input and clinical relevance with the use of SMT, studies have shown that dynamic mechanical forces have been found to produce neuromuscular reflex responses, as opposed to SMT forces delivered in a more static nature. Slower rate SMT forces (1-3 sec) have not been found to produce neuromuscular reflex responses in the back musculature, regardless of whether joint cavitations were elicited.[2,28,29] Presumably, the production of associated reflex responses depends directly on the rate of change in force and deformation during the treatment rather than on the force or stretch magnitude itself.[9] Noteworthy in this regard is the fact that the mechanical instrument (AAI) used in our study is a spinal manipulative device that was specifically designed to impart a consistent high loading rate, short duration, multifrequency thrust but is not ordinarily associated with an audible joint cavitation.

An interesting hypothesis therefore is the notion that both magnitude and frequency content of manual and mechanical thrusting manipulations are critical elements in the determination of therapeutic effects. Consequently, the posteroanterior vibration of the spine may be not only valuable as a diagnostic tool but also may have intrinsic therapeutic potential. In another context, the vibration of the spinal column at certain frequencies may also be used as a noninvasive diagnostic measure of pain provocation through stimulating frequency–specific nociceptive neuron populations.[59-62] For successful and routine clinical use of spine biomechanical analyses, a given method should be noninvasive, reproducible, inexpensive, simple to operate, and painless in application. We found that the dynamic posteroanterior stiffness response was consistent (less than 25% variation for repeated thrusts) and was able to discriminate region-specific and gender-specific differences. The transient vibration measurement and analysis procedure described here appears to be a feasible means with which to satisfy clinical assessment requirements.

CONCLUSION

In summary, the posteroanterior mechanical behavior of the human thoracolumbar spine was found to be sensitive to mechanical stimulus frequency and showed significant region-specific and gender-specific differences. In a relatively small sample of young and elder subjects, however, we found no significant age-related differences in mobility when young male and female subjects were compared with their older counterparts. The latter finding most likely reflects the small numbers of subjects who were examined. Further research is needed to characterize age-related changes in spinal mobility in both asymptomatic and symptomatic patient populations. Such studies should also incorporate measures of spinal disorders that include visual analog pain scales and outcome assessment instruments to analyze the clinical utility of dynamic posteroanterior stiffness assessments.

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CORRECTION

In the July/August issue, bar sections of the legends for Figs 3-6 in "Innervation of the Human Costovertebral Joint: Implications for Clinical Back Pain Syndromes" (2000;23:395-403) should have read as follows:

Fig 3. Left, *bar = 0.10 mm*; Right, *bar = 50 μm*
Fig 4. Left, *bar = 0.1 mm*; Right, *bar = 50 μm*
Fig 5. Right, *bar = 20 μm*
Fig 6. Right, *bar = 20 μm*

We apologize for the error and regret any confusion that it may have caused.