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FIRST PRIZE

BIOMECHANICAL AND NEUROPHYSIOLOGICAL RESPONSES TO SPINAL MANIPULATION IN PATIENTS WITH LUMBAR RADICULOPATHY

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ABSTRACT

Objective: The purpose of this study was to quantify in vivo vertebral motions and neurophysiological responses during spinal manipulation.

Methods: Nine patients undergoing lumbar decompression surgery participated in this study. Spinal manipulative thrusts (SMTs) (~5 ms; 30 N [Sham], 88 N, 117 N, and 150 N [max]) were administered to lumbar spine facet joints (FJs) and spinous processes (SPs) adjacent to an intraosseous pin with an attached triaxial accelerometer and bipolar electrodes cradled around the S1 spinal nerve roots. Peak baseline amplitude compound action potential (CAP) response and peak-peak amplitude axial (AX), posterior-anterior (PA), and medial-lateral (ML) acceleration time and displacement time responses were computed for each SMT. Within-subject statistical analyses of the effects of contact point and force magnitude on vertebral displacements and CAP responses were performed.

Results: SMTs ($\geq$ 88 N) resulted in significantly greater peak-to-peak ML, PA, and AX vertebral displacements compared with sham thrusts ($P < .002$). SMTs delivered to the FJs resulted in approximately 3-fold greater ML motions compared with SPs ($P < .001$). SMTs over the SPs resulted in significantly greater AX displacements compared with SMTs applied to the FJs ($P < .05$). Seventy-five percent of SMTs resulted in positive CAP responses with a mean latency of 12.0 ms. Collectively, the magnitude of the CAP responses was significantly greater for max setting SMTs compared with sham ($P < .01$).

Conclusions: Impulsive SMTs in human subjects were found to stimulate spinal nerve root responses that were temporally related to the onset of vertebral motion. Further work, including examination of the frequency and force duration dependency of SMT, is necessary to elucidate the clinical relevance of enhanced or absent CAP responses in patients. (J Manipulative Physiol Ther 2004;27:1-15)

Key Indexing Terms: *Chiropractic Manipulation; Vertebral Motion; Neurophysiology*

INTRODUCTION

Because spinal manipulation (SM) is a mechanical intervention, it is inherently logical to assume that its mechanisms of therapeutic benefit may lie in the mechanical properties of the applied force (mechanical mechanisms), the body's response to such force (mechanical or physiologic mechanisms), or a combination of these and other factors. Basic science research, including biomechanical and neurophysiological investigations of the body's response to SM, therefore, should assist researchers, educators, and clinicians to understand the mechanisms of

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SM, to more fully develop SM techniques, to better train clinicians, and ultimately attempt to minimize risks while achieving better results with patients.

From a biomechanical perspective, human cadaver and in vivo studies have characterized the forces and force-time histories associated with various spinal manipulation techniques.[1-9] These studies provide important information concerning the forces and loading history transmitted to patients. The posterior-anterior (PA) stiffness or PA load-displacement response of the prone-lying subject during SM has also been investigated using static or low-frequency indentation types of techniques, including mobilization and other physiotherapy simulation devices.[10-15] These studies indicate that the thoracolumbar spine has a quasi-static PA structural stiffness of approximately 15 to 30 N/mm at loads up to about 100 N. While stiffness measurements quantify the force-displacement response of the area under test (vertebrae, disks, and associated soft tissues), such measurements cannot easily distinguish the contribution and/or displacement of individual vertebral components.[16] To precisely quantify relative and absolute movements of individual vertebrae or motion segments in response to dynamic forces, it is necessary to measure displacements, velocities, or accelerations using transducers fixed to intraosseous pins rigidly attached to the spine. Due to the invasiveness of such procedures, however, these techniques are generally limited to studies of human cadavers[17,18] or animals.[19,20] Indeed, research of this nature in living humans is very rare.[21]

In 1994, Nathan and Keller[22] quantified the sagittal plane, intersegmental motion response and stiffness of the thoracolumbar spine of human subjects during mechanical-force, manually-assisted (MFMA) short lever spinal manipulative thrusts (SMTs). In their study, forces were delivered to the spinous processes of the thoracolumbar spine using a chiropractic adjusting instrument equipped with a load cell and accelerometer. The motion response of adjacent lumbar vertebrae was quantified using an intervertebral motion device (IMD)[23] attached directly to intraosseous pins fixed to lumbar spinous processes. They found that the peak-to-peak amplitude of intervertebral or intersegmental motions were up to 6-fold greater when the short duration ($<$ 5 milliseconds) SMTs were delivered closer to the IMD measurement site. In response to the same force amplitude, differences in intervertebral acceleration time and displacement time histories were also noted among the 3 subjects examined in this study (1 normal subject and 2 subjects consulting for surgery). The study by Nathan and Keller[22] was limited to a single force amplitude PA thrust applied over the spinous processes, and only the relative movements of 2 adjacent vertebrae (intersegmental motion) could be determined. To our knowledge, there are no data in the literature that characterize the segmental and intersegmental motion responses of the spine to varying force amplitudes and contact points in living subjects.

From a neurophysiological perspective, the presence of mechanosensitive and nociceptive afferent fibers in spinal tissues (disk, facet, ligaments, and muscles)[24-28] and the subsequent neurophysiological research demonstrating the role of such afferent stimulation in pain production[29-31] and coordinated neuromuscular stabilization of the spine[32-37] provide a theoretical framework to investigate the mechanisms of chiropractic adjustments or spinal manipulation. The mechanical and physiologic influences of spinal manipulation on the targeted spinal tissues that have recently begun to be quantified experimentally represent an important first step in validating chiropractic theories. However, this work has been limited to animal models, noninvasive procedures, or minimally invasive procedures. For example, Pickar and McLain[38] measured afferent unit discharge to facet manipulation, and Pickar and Wheeler[39] measured muscle spindle and golgi-tendon organ responses to spinal manipulative-like loads in the feline. Basic animal research has now demonstrated the existence of neural discharge during spinal manipulative-like loads, but the results are not easily extrapolated in humans.

Intraoperative monitoring techniques have proven beneficial for monitoring neurophysiological events during spinal surgery, and such techniques have been used to study responses of spinal manipulation. Colloca et al[40] recently completed a pilot study investigating spinal nerve root action potential responses during intraoperative lumbosacral spinal manipulation. Spinal nerve root responses were found to be related to segmental contact point, and applied force vector and similarities were observed between internal and external thrusts. This study was limited to a single patient; nerve root measurements were unilateral; and the temporal relationships of the SMTs and nerve root responses could not be studied.

The purpose of the current study was to perform a comprehensive biomechanical and neurophysiological analysis of SMT in a series of 9 symptomatic patients. We hypothesized that neurophysiological and biomechanical responses would be related to the magnitude and location of the SMT, with differential responses dependent on patient symptomatology.

Methods

Nine patients (6 male, 3 female, 32-75 years of age, mean age = 53.4 years) undergoing lumbar laminarthrectomy to decompress the central spinal canal and neuroforamina, as clinically indicated, participated in this study. Two experimental protocols were performed, the first prior to spinal surgery and the second following the spinal decompression surgical procedure. Each patient provided informed consent for the surgical procedure and research protocol in accordance with the ethical standards of the hospital's ethical committee on human experimentation. Patients were selected for spinal surgery based on their history, clinical

Table 1. *Patient and clinical demographics*

Patient	Age (y)	Sex	Side of lower extremity symptoms	Diagnosis	Clinical presentation	Level(s) of decompression
1	72	M	Left	Sciatica and spinal stenosis (Congenital and acquired)	Low back and left leg pain	L2-3; L4-5; L5-S1
2	75	F	Left	Sciatica and spinal stenosis (acquired)	Low back pain, stiffness, left leg pain, and bilateral groin pain	L4-5; L5-S1
3	48	F	Left	Sciatica, disk protrusion, and Spinal stenosis (Congenital)	Left S1 dermatomal leg radiculopathy	L4-5; L5-S1
4	62	M	Bilateral	Spinal stenosis (acquired)	Low back and bilateral leg pain (worse on the right), urinary urgency, and neurogenic claudication	L2-3; L4-5; L5-S1
5	39	M	Left	Disk protrusion	Left leg pain	L4-5; L5-S1
6	41	M	Left	Spinal stenosis (acquired)	L4 dermatomal left foot pain	L3-4; L4-5
7	46	F	Left	Disk protrusion	Left leg pain	L2-3
8	32	M	Right	Disk protrusion	Right leg pain	L3-4
9	66	M	Bilateral	Spinal stenosis (acquired)	Bilateral leg pain with claudication	L3-4; L4-5; L5-S1

findings, and confirmed diagnostic imaging documentation of either spinal stenosis, osteoarthritis, and/or disk protrusion. All patients were unresponsive to conservative care for at least 6 months prior to surgery. Patient demographics, diagnoses, clinical presentations, and levels of spinal surgical decompression appear in Table 1.

Patients were brought to the operating room and general endotracheal anesthesia was induced. Patients were placed prone on a surgical frame and their lower backs were prepped and draped in a normal aseptic fashion. Padded supports were placed at the level of the iliac crests and sternum, with a slight flexion of the hips and knees to assure that the subjects were lying in a lordotic position simulating the normal erect posture. Preoperative medication included lorazepam. For induction, propofol, Sufenta, and Thivacron or Esmeron (rucuroniumbromide) were administered. For maintenance, a mixture of nitrous oxide (N_20), oxygen (0_2), and Sevorane was administered. Cefamandol was used for antibiotic prophylaxis. Initial anesthetics did not include any long-lasting ($>$ 15 minutes) paralyzing agents.

Using fluoroscopic guidance, finely threaded, 1.8-mm diameter intraosseous stainless steel pins were rigidly fixed to the L1, L3, or L4 lumbar spinous processes (Fig 1). A dynamic (0.3 Hz to 10 kHz), low-noise (0.0003*g* root-mean-square [RMS] resolution), AC-coupled piezoelectric, integral sensor, triaxial accelerometer (Crossbow Model CXL100F3, Crossbow Technology, Inc, San Jose, Calif) was attached to the intraosseous pin (Fig 1). The x-axis, y-axis, and z-axis of the accelerometer were oriented with respect to the medial-lateral (ML), posterior-anterior (PA), and cranial-caudal or axial (AX) axes of the vertebrae. The natural frequency of the pin and transducer, determined intraoperatively by "plucking" the pins in the ML and AX axes, was greater than 80 Hz. All equipment (electrodes, accelerometers, bone pins, and adjusting instruments) was gas sterilized prior to surgery.

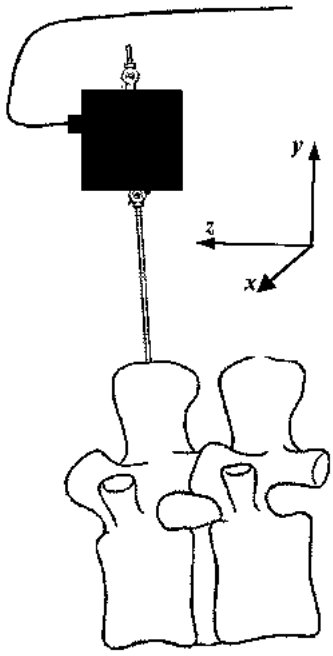

Fig 1. *Schematic illustration of the pin-accelerometer preparation. The Cartesian coordinate system shows the medial-lateral (x), posterior-anterior (y), and axial (z) motion axes.*

Mechanical force, manually-assisted spinal manipulative thrusts were delivered to the musculature overlying the facet joints (FJs) and to the spinous processes (SPs) using an Activator II Adjusting Instrument (AAI) (Activator Methods International, Ltd, Phoenix, Ariz). Four different AAI force excursion settings (0, 1, 2, and 3) were examined with thrusts delivered at the end of expiration during the patient's breathing cycle. In the first protocol, PA anterior-inferior vectored thrusts (approximately 20° with respect to vertical) were applied to the skin overlying the left facet joint (LFJ)

Table 2. *Summary of AAI thrust locations and excursion force settings for protocol 1 and protocol 2*

Contact Point	AAI Excursion Setting 0	1	2	3
	2 (3)†	2 (0)	2 (0)	8 (5)
Above Pin				LFJ RFJ SP
At Pin	LFJ (LFJ) RFJ (RFJ)	LFJ RFJ	LFJ RFJ	LFJ (LFJ x2) RFJ (RFJ x2)
Below Pin	(SP)			LFJ RFJ SP (SP)
Force (N)*	30.2 (5.8)	88.0 (9.0)	116.7 (8.5)	149.5 (48.5)

Protocol 2 in parentheses.
AAI, Activator II Adjusting Instrument; *LFJ*, left facet joint; *RFJ*, right facet joint; *SP*, spinous process.
*Total number of thrusts for protocol 1 and protocol 2 (in parentheses) force settings (each patient, 9 patients total).
†Mean (SD) for patients 003, 006, and 008.

and right facet joint (RFJ) at the level of the pin at each of the force settings (8 thrusts in each patient). SMTs were also applied at the max setting (setting 3) to the skin overlying the FJs (left and right) and to the spinous process above and below the level of the pin (6 thrusts in each patient). Thus, each patient received 14 SMTs (refer to Table 2).

Segmental contact points for the SPs were determined using fluoroscopic guidance and palpation. In the case of thrusts applied over the FJs, contact points were consistently established by contacting 10 to 15 mm lateral to the SPs. Henceforth, settings 0 and 3 will be referred to as the "sham" and "max" settings, respectively. SMTs were performed by an advanced proficiency rated clinician (CJC) who was careful to perform the thrusts in a manner consistent with delivery of MFMA SMT in routine clinical practice. Approximately 20 N of preload was applied prior to the application of each SMT including 0 setting sham SMTs. Details of the AAI and its clinical usage are found elsewhere.[6,41,42]

Each AAI included an electronic trigger to initiate data collection using a Biopac MP150 data acquisition system (Biopac Systems, Inc, Goleta, Calif). Vertebral accelerations (ML, PA, AX) and AAI force-acceleration responses (patients 3, 6, and 8) were recorded at a sampling frequency of 8192 Hz using a Biopac MP150 12-bit data acquisition system and Acknowledge software (Biopac Systems, Inc, Goleta, Calif).

Following the first experimental protocol, spinal decompression surgery was performed as clinically indicated (Fig 2). Incisions were made over L3-S2 in the midline and brought through the subcutaneous tissue. The fascia was incised and the musculature was carefully dissected on the left side of the spinous process, which was osteotomized at the base. Self-retaining retractors were set in place, thus exposing the full posterior arches and ligamenta flava, and manual suction was performed within the incised area. A laminarthrectomy was performed to decompress the central spinal canal and neuroforamina, as clinically indicated, and the integrity of the neural arches, facet joints, and most muscle attachments was preserved. This surgical procedure affords excellent visualization and a wide area available while minimizing destruction to tissues not directly involved in the pathologic process, including the paraspinal musculature, interspinous/supraspinous ligament complex, and facets.[43] The integrity of the facet joints is also preserved by this procedure. Inspection of the epidural space indicated that the L4-5 and L5-S1 intervertebral disks were not ruptured in any of the patients.

On completion of the decompression surgery, the L5 and S1 nerve root sleeves were clearly identified and free of all compression, and the second experimental protocol was initiated. Two bipolar, hooked, platinum electrodes (PolarProbe, Nicolet, Inc, Madison, Wis) were subsequently cradled around the left and right S1 spinal nerve roots at the level of the dorsal root ganglia to record neurophysiological responses (compound action potential [CAP]). The bipolar electrodes had 10-mm spacing and 64-mm tip length and were shielded and insulated such that the most distal (hooked) end was exposed for recording (refer to Fig 2). CAP electrodes were connected to biopotential amplifiers (ERS100B, Biopac Systems, Inc, Goleta, Calif) using a 3-m extension cable and plug (MEC100, Biopac Systems Inc, Goleta, Calif). The amplifier gain setting was 5000× to 10,000× and the amplifier filter settings were 5000 Hz low pass and 10 Hz high pass. To test the working order of the electrodes, the skin over the calf was stroked to stimulate the

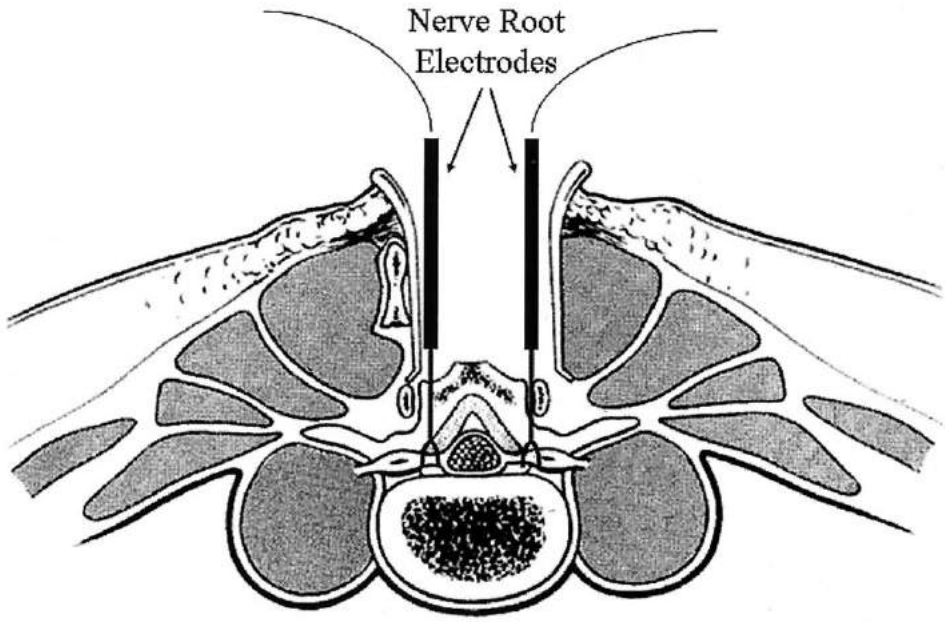

Fig 2. *Schematic illustration of the surgical exposure and experimental placement of the bipolar platinum nerve root electrodes around the spinal nerve roots.*

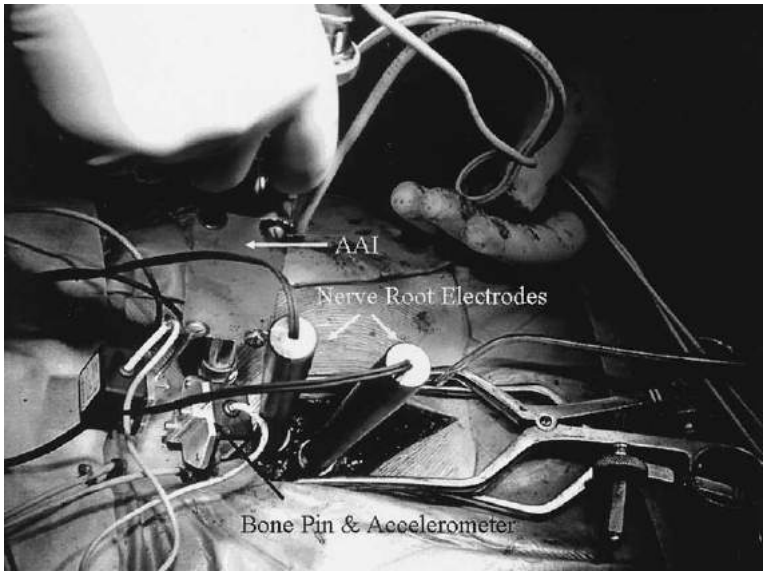

Fig 3. *Experimental setup showing the application of a spinal manipulative thrust delivered with a mechanical force, manually-assisted adjusting instrument (AAI) adjacent to the bone pin and accelerometer mount and spinal nerve root electrodes.*

S1 dermatome, and CAP electrode activity was noted. When nerve activity was not observed, the electrodes were readjusted by the surgeon and a repeated test was performed until satisfactory CAP activity was observed. On occasion, only sparse activity was observed during the S1 dermatome stimulation, which we believed to be neurological damage consistent with the clinical presentation of the particular patient.

In the second experimental protocol, a total of 8 SMTs were delivered to the skin overlying SPs and to the skin and musculature overlying the FJs of each patient (refer to Table 2). Specifically, PA anterior-inferior (1 max, 1 sham) and PA anterior-superior (1 max, 1 sham) vectored thrusts (approximately 20° with respect to vertical, caudal, or cranial, respectively) were each applied to the skin overlying the left and right FJs at the level of the pin. Two SMTs (1 max, 1 sham) were applied to the spinous process below the pin with a PA anterior-inferior vector. Biomechanical (AX pin accelerations only) and neurophysiological responses (left S1 and right S1 CAPs) were simultaneously recorded at a sampling frequency of 4096 Hz. The nerve root electrode

placement, pin accelerometer placement, AAI, and surgical preparation site are illustrated in Figure 3.

Displacement time responses were obtained from the acceleration time histories using trapezoidal numerical integration.[22] Postprocessing of the acceleration time histories was performed using Matlab software (The MathWorks, Natick, Mass) and included determination of peak-to-peak magnitudes of the vertebral acceleration, velocity, and displacement time histories. Based on acceleration measurements performed by displacing the pins a known amount, the trapezoidal numerical integration procedure was found to predict peak displacements within 5% to 10%. CAP signals were filtered using a 0-phase forward and reverse digital bandstop filter (45-55 Hz) followed by a 0-phase forward and reverse digital low pass filter (500 Hz). The filter was designed to reduce electrical noise associated with the operating theatre and did not alter the amplitude and temporal characteristics of the biopotential signals. Positive CAP responses were defined as a peak-peak (p-p) amplitude response greater than 2.5 times the peak-peak baseline (resting) signal.[44,45] A peak detector was used to find the peak in the axial acceleration time history. A 10-ms window immediately prior to the acceleration peak and a 100-ms window immediately following the peak was then analyzed to obtain baseline minimum, maximum, peak-peak, and mean values for each thrust.

The time interval or temporal relationship between initiation of the SMT and initiation of the CAP responses was calculated for each of 3 patients examined using the force-accelerometer instrumented AAI. The temporal relationship for the remaining 6 patients was estimated by adding the mean time interval (2.2 ms) from the onset of the SMT acceleration to the resulting pin acceleration to the peak-to-peak time interval of the pin axial acceleration to the peak CAP responses. For statistical purposes, only peak-to-peak acceleration and displacement responses are considered in this report. Descriptive statistics and within-patient statistical (paired observations *t* test) comparisons of the effects of contact point and force magnitude on peak-to-peak vertebral displacements and peak-to-baseline CAP responses were performed.

Results

For patients examined using the force-accelerometer instrumented AAI (patients 3, 6, and 8), the average setting 0, 1, 2, and 3 peak SMT forces were 30 N, 88 N, 117 N, and 150 N, respectively (Table 2). The approximately 5-ms duration MFMA SMTs produced vertebral oscillations (displacements and accelerations) spanning a time period of 100 ms to 150 ms (Fig 4). Thrusts over the FJs resulted in greater peak-peak ML and PA accelerations in comparison with peak-peak AX accelerations. Thrusts over the SPs resulted in greater peak-peak PA accelerations in comparison with peak-peak ML and AX accelerations. Average

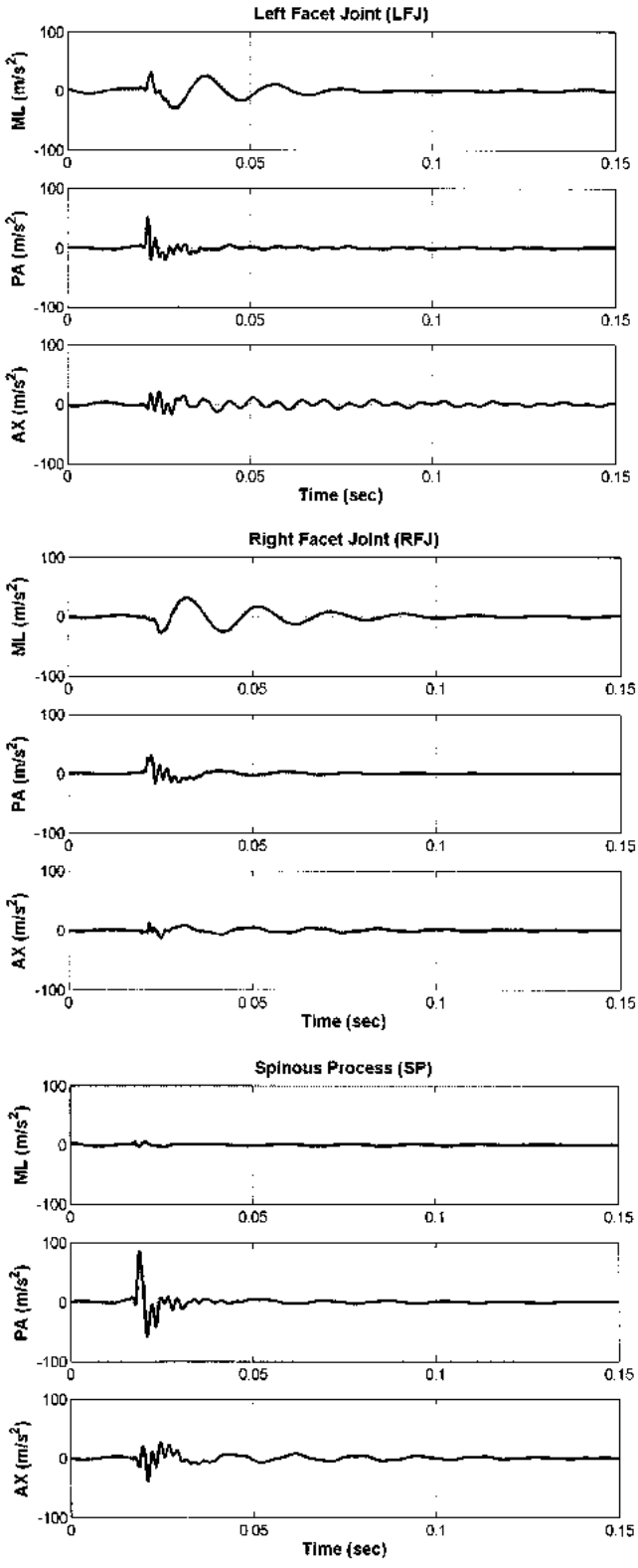

Fig 4. *Typical medial-lateral (ML), posterior-anterior (PA), and axial (AX) acceleration time responses for maximum SMTs on the left facet joint, right facet joint, and spinous process of the L2 vertebral body (patient 006).*

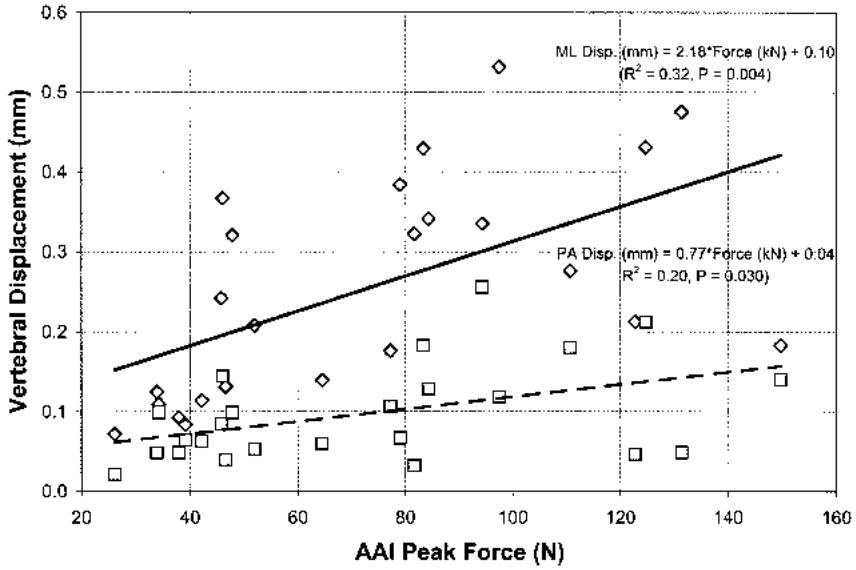

Fig 5. *Peak vertebral displacement response versus peak applied force obtained for posterior-anterior SMTs over the facet joints at the level of the pin. Medial-lateral (ML) displacement response (shaded diamonds) and posterior-anterior (PA) displacement response (shaded squares) showed a statistically significant linear relationship with respect to the amplitude of the anterior-interior vectored PA Activator adjusting instrument (AAI) force. Axial (AX) displacements were not significantly correlated (R^2 = 0.10, P = .14) to the applied PA force. Results shown are for the instrumented AAI SMTs (8 facet joint thrusts at the level of the pin each for patients 3, 6, and 8). Linear regression equation, coefficient of determination (R^2), and statistical significance (P-value) are shown for the ML (solid line) and PA (dashed line) responses.*

Table 3. *Vertebral segment peak-peak motion response summary*

	Medial-Lateral (ML)		Axial (AX)		Posteroanterior (PA)	
Thrust location	SP	FJ	SP	FJ	SP	FJ
Displacement (mm)	0.18 (0.09)	0.53 (0.27)	0.46 (0.24)	0.37 (0.23)	0.66 (0.30)	0.66 (0.24)
Velocity (mm/s)	44.6 (19.7)	140.8 (77.1)	147.1 (55.8)	105.9 (50.0)	163.0 (53.8)	116.3 (32.3)
Acceleration (m/s^2)	21.8 (11.7)	61.1 (36.6)	96.4 (35.1)	53.5 (29.3)	151.9 (55.8)	74.1 (40.8)

Mean (SD) for maximum setting Activator II Adjusting Instrument (AAI) thrusts over the spinous processes (n = 18) and facet joints (n = 54).
SP, spinous process; *FJ*, facet joint.

peak-peak acceleration, velocity, and displacement responses obtained for SMTs delivered to the SPs and FJs are summarized in Table 3.

Collectively (all 126 thrusts), the ML, PA, and AX peak-to-peak displacements for the SMTs ranged from 0.03 mm to 1.30 mm (mean = 0.44 mm), 0.10 mm to 1.28 mm (mean = 0.56 mm), and 0.06 mm to 1.32 mm (mean = 0.33 mm), respectively. For SMTs delivered to the FJs at the level of the pin, both ML and PA vertebral displacements increased in a relatively linear manner with increasing AAI force setting (Fig 5). PA SMTs resulted in statistically significant increases in peak-to-peak ML (settings 2, 3), PA (settings 1, 2, 3), and AX (settings 2, 3) vertebral displacements compared with sham (setting 0) thrusts ($P < .002$). SMTs delivered to the FJs resulted in approximately 3-fold greater ML displacements compared with SMTs delivered to the SPs ($P < .001$). No statistically significant differences were observed for PA vertebral displacements during SMTs on the SPs and FJs. SMTs to the SPs resulted in significantly ($P < .05$) greater (22%) AX displacements compared with SMTs applied to the FJs. The influence of thrust force magnitude and location are graphically summarized in Figure 6 and Figure 7, respectively.

Seventy-five percent of the SMTs resulted in a positive CAP response (peak-peak response > 2.5× baseline). The majority of SMTs that resulted in positive CAP responses were characterized by a single evoked action potential (Fig 8). Using the force-accelerometer instrumented AAI in 3 subjects, the mean temporal relationship between the initiation of the SMT and initiation of a positive CAP response was 12.0 ms (range 8.2-17.3 milliseconds). Collectively, the combined left + right (L+R), peak-peak CAP magnitude

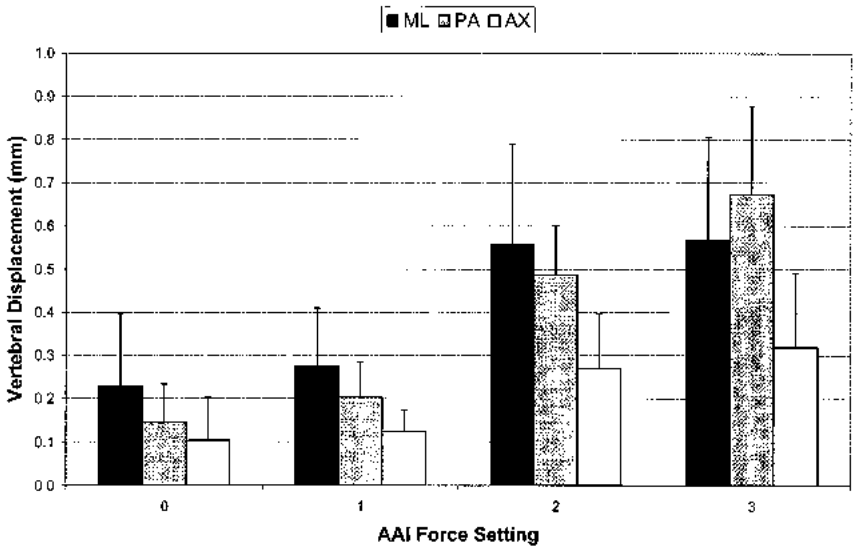

Fig 6. *Mean lumbar vertebral segmental displacement response to posterior-anterior (PA) anterior-inferior thrusts over the facet joints at the level of the pin. Medial-lateral (ML), posterior-anterior (PA), and axial (AX) motion responses to the 4 force settings (defined in text) are shown.* Error bars *indicate SDs.*

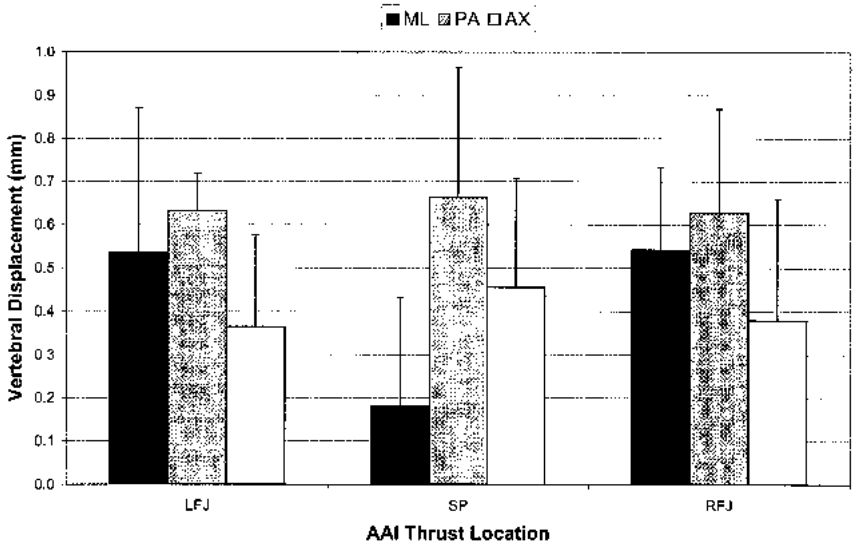

Fig 7. *Mean lumbar vertebral segmental displacement responses to posterior-anterior (PA) anterior-inferior thrusts over the left facet joint (LFJ), right facet joint (RFJ), and spinous process (SP). Medial-lateral (ML), posterior-anterior (PA), and axial (AX) motion responses to the maximum force setting are shown.* Error bars *indicate SDs.*

was significantly ($P < .01$) greater for max setting, anterior-inferior vectored SMTs (n = 2 sides × 3 locations × 9 subjects = 54) compared with similarly vectored sham setting SMTs (n = 2 × 27 = 54). No significant differences in the magnitude of L+R CAP responses were observed for SMTs delivered to the SPs in comparison with the FJs. Mean left S1 nerve root and right S1 nerve root CAP responses for each of the 8 protocol 2 SMTs are summarized in Figure 9. The percentage of positive CAP responses for each of the SMT contact points is summarized in Table 4. In the case of patients with left side symptoms, positive CAP responses were seen more commonly on the contralateral side of lumbar radiculopathy (Table 5).

Discussion

This clinical biomechanical study confirmed that spinal manipulation induces spinal motion and concomitant spinal nerve root responses. This line of investigation is the first to simultaneously measure vertebral movements and nerve root responses during SMT in human subjects. Such neuromechanical responses may be related to the mechanisms

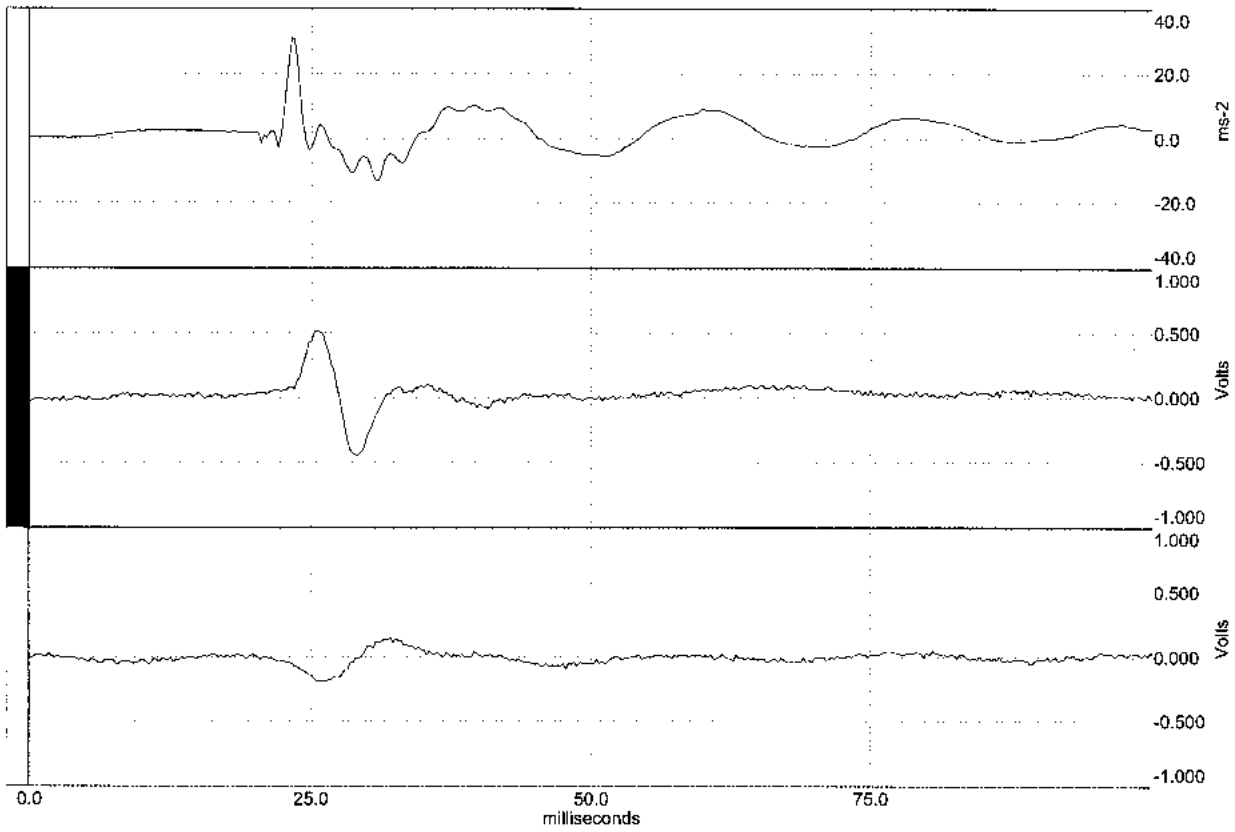

Fig 8. *Typical axial (AX) displacement (z-axis acceleration, top graph) and S1 spinal nerve roots compound action potential (CAP) responses (L-S1 nerve root, middle graph; R-S1 nerve root, bottom graph) for a maximum posterior-anterior (PA) anterior-inferior spinal SMT on the right facet joint of patient 008. Initiation of the AX acceleration response occurred approximately 2.2 ms following initiation of the SMT. A positive bilateral nerve root CAP response is illustrated. Nerve root CAP responses were acquired using a biopotential amplifier and digitally filtered using the protocol described in the text.*

of spinal manipulation as administered in routine clinical practice.

Biomechanical Findings

Due to the invasiveness necessary to quantify spinal motions during spinal manipulation, previous research has typically been limited to cadaver studies.[1,17,18,46] Gál et al[17] measured relative movements between vertebral bodies during PA thoracic SM. In this study, steel bone pins were embedded in the vertebral bodies of 2 unembalmed postrigor cadavers (aged 77 years each) at the levels of T10, T11, and T12. High-speed cinematography measured spinal motions during SM delivered at the level of T11. Preload and peak forces were approximately 80 N and 525 N, respectfully, in their study. These authors reported statistically significant mean relative translations and rotations ranged from 0.3 mm ± 0.2 mm to 0.6 ± 0.4 mm and 0.0 ± 0.3° to 1.9 ± 0.2°, respectively, between the 2 subjects. Similarly, Maigne and Guillon[46] measured relative lumbar spinal motions during lumbar spinal manipulation in 2 unembalmed cadavers (aged 49 and 71 years) by implanting accelerometers in the vertebral bodies. Using side-posture manipulation, the authors reported a maximum approximation between the L4-5 functional spinal unit of 1.1 mm, which is consistent with the magnitudes of relative vertebral movements observed in the current study. The ML, PA, and AX peak-to-peak vertebral displacements in this study are also of the same magnitude as previously reported in situ and in vivo relative or intervertebral motion studies.[22] Differences in the vertebral displacement response for the current study reflect subject differences, recording and sampling methodologies utilized, SMT force magnitude and duration, and segmental versus intersegmental nature of measurements.

Differences in vertebral motion responses associated with thrusts applied on various anatomical landmarks are important to clinicians who apply forces to the spine. In the current study, SMTs delivered to the FJs resulted in significantly (approximately 3-fold) greater ML motions as compared with SMTs delivered to the SPs. Because the SMT force vector was similar for thrusts on SPs and FJs, it is apparent that the segmental contact point has a direct influence on the vertebral motion response that is elicited. For clinicians, ML motion during spinal manipulation is accom-

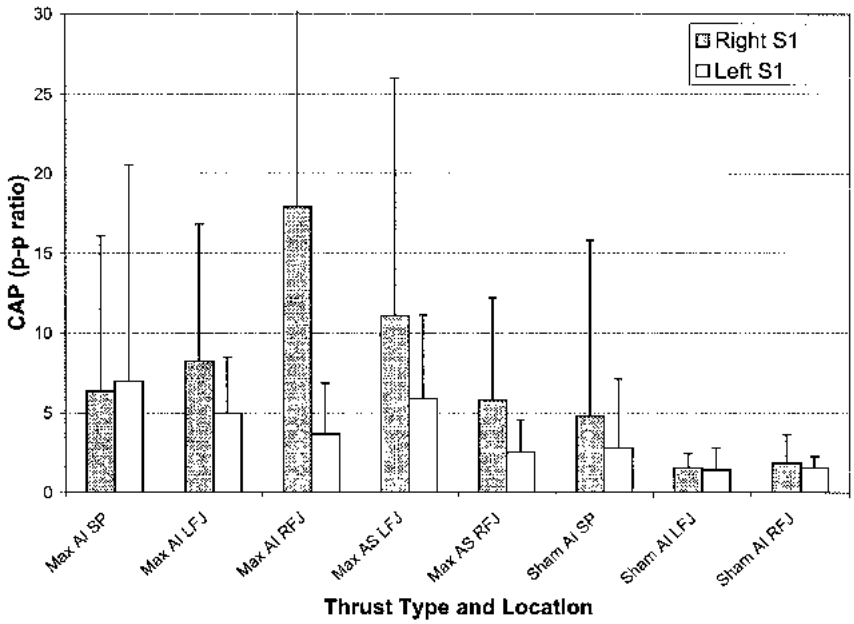

Fig 9. *Compound action potential (CAP) responses to maximum setting (max) and zero setting (sham) posterior-anterior (PA) thrusts over the left facet joint (LFJ), right facet joint (RFJ), and spinous process (SP). The CAP peak-peak (p-p) ratio was defined as the ratio of the peak-peak amplitude obtained during the 100-ms interval following the peak axial displacement and the peak-peak amplitude of the baseline signal prior to the SMT. AI, anterior-inferior force vector; AS, anterior-superior force vector.*

Table 4. *Effects of SMT contact point on positive CAP responses*

SMT contact point	Left S1 CAP	Right S1 CAP
SP	44.4	77.8
LFJ	64.7	70.6
RFJ	33.3	77.8
FJ (L+R)	48.6	74.3

Percent of thrusts >2.5 × baseline.
SMT, spinal manipulative thrust; *CAP*, compound action potential; *SP*, spinous process; *LFJ*, left facet joint; *RFJ*, right facet joint; *FJ*, facet joint; *L*, left; *R*, right.

Table 5. *Effects of lumbar radiculopathy on CAP responses to maximum force SMT delivered over spinous processes and facet joints*

Side of symptoms	Left S1 CAP	Right S1 CAP
Left (6 subjects)	56.7	73.3
Right (1 subject)	75.0	100
Bilateral (2 subjects)	10.0	70.0

Percent of thrusts >2.5× baseline.
CAP, compound action potential; *SMT*, spinal manipulative thrust.

plished by applying the SMT to the FJ as opposed to the SP. Moreover, in the case of the impulsive-type forces (force-time period « natural frequency) produced during MFMA SMT, the vertebral displacement response increased in a relatively linear manner with increasing force amplitude (constant preload).

A limitation of the current study was the fact that we did not quantify the precise thrust angle and FJ segmental contact points during the SMTs. Both of these factors may influence the motion response, but the surgical setting and the complexity of the motion and neurophysiological measurements performed precluded such measurements. Care was taken to perform the SMTs in a consistent and routine clinical manner, namely PA anterior-inferior or anterior-superior angulations of 20° ± 5° and offset of 10 mm to 15 mm from the midline (thrusts over FJs). Our aim was to quantify the lumbar vertebral motion response associated with spinal manipulation as it is performed in routine clinical chiropractic practice. According to computer simulations performed by Keller et al,[47] a 5° angulation difference (−15° versus −20°) and 5-mm contact point offset are predicted to result in less than a 0.1-mm difference in the peak-to-peak PA and axial motion responses to impulsive forces. Thus, lumbar spine PA and AX motion responses to impulsive forces are thought to be relatively insensitive to thrust angle/contact point variations of 20°/5 mm or less. While imaging technology is currently available to identify the underlying segmental contact points during biomechanical assessments,[10,48] we do not believe that this specificity would have assisted our aim of quantifying vertebral motions during clinically applied SMT. Nevertheless, the influence of variations in precisely controlled force vector and contact point on the in vivo motion response deserves further consideration.

The MFMA instrument used for the SMTs produced a very short time duration (impulsive) force that induced a

transient dynamic oscillatory motion response. For a given force amplitude, impulsive forces are associated with smaller displacements in comparison with longer duration, nonperiodic forces, such as those commonly applied during manual manipulation.[47] Consequently, high-precision, low-noise, dynamic accelerometers were used in this study to quantify the dynamic motion response of individual segments. The posterior-anterior, medial-lateral, and axial acceleration responses and displacements derived from the acceleration responses indicate that the method yields results comparable with other kinematic measurement methods, including spatial linkage sensors.[23] Additional work is needed to determine the reproducibility of the acceleration-based vertebral motion analysis method.

In the current study, we did not transform the Cartesian components of acceleration (x, y, z) to account for rotations of the vertebral segments or to estimate the flexion-extension rotation and medial-lateral rotation of the segments. Such transformations require knowledge of the location of the rotation axes relative to the accelerometer axes, and although we obtained fluoroscopic images of the pin-accelerometer sites, the image quality and image coverage was insufficient to perform these measurements in a manner precise enough to warrant transformation. Given the small absolute x, y, and z vertebral displacements measured (< 1 mm), vertebral rotations would be predicted to be extremely small, and therefore the transformed vertebral motions would not be expected to vary appreciably from that reported in this study. The absolute intervertebral flexion-extension rotations ($< 1°$) reported by Nathan and Keller[22] and vertebral and intervertebral flexion-extension rotations reported by Keller et al[47] support this assumption. A 6–degree-of-freedom motion measurement system (3 translations and 3 rotations) would provide a more precise description of vertebral displacements and could be used to obtain vertebral rotations.

Our results are presented for patients undergoing surgery for significant spinal disorders and therefore should not be considered "normal lumbar segment motion responses." As previously noted, investigations into spinal motions during spinal manipulation are in their infancy, so readily available data regarding spinal motions in normal subjects as opposed to subjects with spinal disorders are sparse.[22] A number of studies indicate that it is likely that spinal motions are highly dependent on the force-time input of the directed thrust,[14,49,50] as well as a variety of clinical factors, such as pain,[7,13,51] spinal morphology,[52] the presence of degeneration,[16,53,54] and muscular stiffness.[55,56] Therefore, vertebral motions observed in the spinal surgery patients are not expected to be representative of normal or asymptomatic subjects. Recent work by Kaigle et al[57] examined in vivo spinal motions and muscular responses in patients and asymptomatic subjects performing unresisted flexion-extension tasks. They found that intervertebral motions and trunk mobility were significantly lower in the patients than controls both in terms of range and pattern of motion. Still other factors such as intra-abdominal pressure,[58] cycle of breathing,[59] spinal level being tested,[22,60] vector of applied force,[61-63] and spinal positioning during testing[64] have all been found to be important variables of spinal motion. In the current study, we accounted for many of these variables by placing patients in the same position on the same frame, standardizing the segmental level, vector, and cycle of breathing during performance of the SMTs. Further work in this regard with respect to understanding spinal motion differences among patients and asymptomatic subjects is warranted.

The results obtained from this study provide basic biomechanical information that is useful to both clinicians and researchers. The dynamic motion response data, force dependence, and coupling characteristics of the spinal segments to PA thrusts reported in this study will also assist researchers in the development and validation of computer models that aim to simulate the static and dynamic motion response of the spine.[47,65-67] Based on the results of this study, a recent model developed by Keller et al[47] is currently being refined to include motion coupling in each of the orthogonal axes of the spine.

Neurophysiological Findings

Based on the knowledge of the presence of mechanosensitive afferents in the discoligamentous and muscular spinal tissues, we assumed that mechanical stimulation of viscoelastic structures during SMT would result in physiologic responses in human subjects.[25,26,29] Prior research has demonstrated that mechanical and electrical stimulation of spinal articulations results in neurophysiological and neuromuscular responses, but such research has mostly been limited to the laboratory utilizing animal models.[36-39,68] Intraoperative monitoring techniques are currently used in spinal surgery[69-73] and offer promise for evaluating neurophysiological responses during SMT,[40] albeit limited to the research setting. Thus, the objective of the current study was to measure intraoperative neuromechancial responses with a commonly used conservative therapeutic approach—spinal manipulation.

Because our measurements were taken just adjacent to the dorsal root ganglion, it is likely that the SMT-induced CAPs observed in the S1 spinal nerve roots were afferent traffic resulting from the stimulation of mechanosensitive afferent fibers in the viscoelastic spinal tissues. Sensory receptors within a tissue such as spinal ligaments, facets, disks, and muscles can initiate neural outflow to the spinal cord during application of various mechanical stimuli (eg, pressure, elongation, vibration, friction, tissue crushing) and application of chemical stimulants.[31] However, we were not able to directly ascertain the exact source of the neurophysiological responses, as is routinely performed in animal studies.[74,75] Rather, intraoperative monitoring of compound action potentials was performed, which represents the alge-

braic sum of action potentials arising from respective mechanosensitive axons passing through the epineuria of the dorsal spinal nerve roots. Because the CAP represents many axons with differing thresholds of excitation, the CAP response is graded with a magnitude that is proportional to the intensity of stimulation.

We originally hypothesized that neurophysiological and biomechanical responses would be related to the magnitude and location of the SMT, with differential responses dependent on patient symptoms. Indeed, we found that variable intensity SMTs produced CAP responses of different amplitudes. Moreover, the magnitude of the CAP responses was significantly greater for SMTs compared with sham thrusts, indicating that the CAP response was not a product of preload. However, because we observed no difference in CAP response for MFMA SMTs delivered to the SPs or FJs, our findings indicate that spinal nerve root responses may not be sensitive to segmental contact point. Larger force magnitudes as delivered in other forms of manual SMT may cause more frequent and larger amplitude biomechanical and neurophysiological responses.[76] Further investigation into the effects of force-time profiles and segmental contact points on neuromechanical responses is warranted.

The mean reflexogenic time duration (SMT-to-peak positive CAP response) obtained in this study is similar to the work of others who have stimulated spinal structures and recorded physiological responses.[32,33,36,69] Some researchers have used electrical stimulation to measure reflexogenic activity in the adjacent spinal musculature. Indahl et al[36,68] reported time durations of 4 ms to 8 ms in a porcine model on stimulating the intervertebral disk and sacroiliac joint. Kang et al[74] also reported similar stimulus-to-response times of about 10 ms in feline preparations. Solomonow et al[33] measured stimulus-to-response time durations of 5 ms to 10 ms in human subjects on electrical stimulation of the supraspinous ligament. Stimulus-to-response times in the current study corroborate these time durations in our human subjects. It is likely that the CAP response represents afferent traffic from multiple mechanosensitive units in the muscular and discoligamentous soft tissues. The average 12-ms delay between the SMT and positive CAP response in the current study are expected due to the time it takes for the stimulus to travel along the Ia fibers, through the dorsal root ganglion, to the spinal cord. Neurologic deficits inherent in the patient population of the current study may have resulted in stimulus-to-response delays or the absence of positive CAP responses altogether. Indeed, a significant percentage of SMTs did not elicit positive neurophysiological responses in the patients. However, with the current methodology, it was not possible to ascertain whether the presence (or absence) and amplitude of CAP responses were specifically related to the neurologic status of the patient.

Nevertheless, it would not be unreasonable to expect neurologic deficits from damaged tissues. Three fourths of patients in this study had radiculopathy in the left lower extremity. Such clinical presentation might help to explain the greater number of right-sided (asymptomatic side) positive S1 CAP responses, as opposed to those measured from the left S1 spinal nerve root. This is consistent with the findings of Solomonow et al[33] who reported an absence of electromyography (EMG) responses during intraoperative stimulation of the supraspinous ligament. Hence, neurological deficit among patients may explain the decreased number of positive neurophysiological responses to SMT. In assessing the CAP response, positive responses were based on a threshold level of 2.5 $\times$ baseline. Responses at lower levels were not counted as "positive." In a previous study,[44] peak-peak EMG reflex responses to PA thrusts were categorized according to 8 different baseline thresholds: $>1.5\times$, $>2.0\times$, $>2.5\times$, $>3.0\times$, $>3.5\times$, $>4.0\times$, $>4.5\times$, and $>5.0\times$ the baseline p-p surface electromyography (sEMG) values. Here baseline refers to the resting or reference noise level of the biopotential (CAP in this study). A 1.5-fold increase ($1.5\times$) represented a very weak reflex response, whereas a 5-fold increase ($5.0\times$) represented a very strong reflex response. A $2.5\times$ response was chosen for this study to ensure that the CAP responses were substantially greater than the background noise level. The clinical relevance of CAP threshold needs to be clarified further. A larger patient population will assist in clarifying the neuromechanical effects of SMT, including the effects of force vectoring, force-time profiles, and segmental contact points on neuromechanical responses. In particular, investigation of traditional manual SMT procedures[9] is necessary to better describe the neuromechanical responses of SMT.

Controversy may arise over our terminology reporting the use of "sham" SMT, since the so-called sham setting produces a 30 N peak impulse force. This setting has been referred to as a sham SMT by us and other investigators.[77,78] Subsequently, both biomechanical and clinical studies have been performed using the zero (sham) and max settings of the device. Noteworthy, Keller and Colloca[77] found that the trunk muscle function assessed using erector spinae muscle electromyography was significantly improved in patients who received a max setting AAI SMT intervention. These authors found that there was no functional improvement in trunk muscle function for patients who received sham (0 setting) AAI SMTs or control (no intervention) treatment.

In the current study, the CAP response was temporally related to the onset of the MFMA SMTs and not to the initiation of the preload force. Although we did not include a control protocol that applied a preload force without engaging the AAI, our previous research showed that CAP responses were not elicited during the application of a preload force alone.[40] In this work, other control experiments, wherein the CAP electrode was intentionally moved on the spinal nerve root, were not found to produce a CAP response. Thus, we feel confident that the CAP responses observed in the current study are not experimental artifacts.

From a data analysis point of view, engaging the AAI also helped to facilitate the neuromechanical temporal and amplitude measurements performed in this study.

Neurophysiologic models theorize that SMT may stimulate or modulate the somatosensory system and subsequently may evoke neuromuscular reflexes.[38,79-81] Such reflexes are thought to inhibit hyperactive musculature, inhibit nociceptive traffic, and improve spinal function. The current line of investigation assists in understanding the relationships between the mechanical stimulation as delivered in SMT and the concomitant biomechanical and neurophysiological (neuromechanical) responses. In attempting to understand such neuromechanical relationships, often overlooked is the clinical status of the patient. The highly individualized neuromechanical response characteristics among patients in this study serves to highlight the need to clinically correlate the neuromechanical response characteristics with patient clinical status. Identifying such clinical relevance and understanding just how SMT may be related to inhibition or stimulation of the central nervous system in modulating nociception in humans awaits clarification. Our current work and the work of others aim to investigate such issues.[82-84]

Conclusion

In vivo PA impulsive force SMTs in human subjects were found to produce spinal nerve root responses that were temporally related to the onset of vertebral motion. These findings suggest that vertebral motions produced by spinal manipulation may play a prominent role in eliciting physiologic responses. Patient clinical status also appears to have a prominent role in the presence of neurophysiological responses. Further work, particularly examination of the force magnitude and frequency dependency of SMT, is necessary to elucidate the clinical relevance of enhanced or absent CAP responses in patients. Knowledge of biomechanical and neurophysiological events that occur during spinal manipulation assists in formulating a theoretical framework to understand the mechanisms of spinal manipulation.

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References

1. Gal JM, Herzog W, Kawchuk GN, Conway PJ, Zhang YT. Forces and relative vertebral movements during SMT to unembalmed post-rigor human cadavers: peculiarities associated with joint cavitation. J Manipulative Physiol Ther 1995;18:4-9.
2. Kawchuk GN, Herzog W. Biomechanical characterization (fingerprinting) of five novel methods of cervical spine manipulation. J Manipulative Physiol Ther 1993;16:573-7.
3. Herzog W, Conway PJ, Kawchuk GN, Zhang Y, Hasler EM. Forces exerted during spinal manipulative therapy. Spine 1993;18:1206-12.
4. Kawchuk GN, Herzog W, Hasler EM. Forces generated during spinal manipulative therapy of the cervical spine: a pilot study. J Manipulative Physiol Ther 1992;15:275-8.
5. Triano J. The mechanics of spinal manipulation. In: Herzog W, editor. Clinical biomechanics of spinal manipulation. Philadelphia: Churchill Livingstone; 2000. p. 92-190.
6. Keller TS, Colloca CJ, Fuhr AW. Validation of the force and frequency characteristics of the activator adjusting instrument: effectiveness as a mechanical impedance measurement tool. J Manipulative Physiol Ther 1999;22:75-86.
7. Colloca CJ, Keller TS. Stiffness and neuromuscular reflex response of the human spine to posteroanterior manipulative thrusts in patients with low back pain. J Manipulative Physiol Ther 2001;24:489-500.
8. Hessell BW, Herzog W, Conway PJ, McEwen MC. Experimental measurement of the force exerted during spinal manipulation using the Thompson technique. J Manipulative Physiol Ther 1990;13:448-53.
9. Triano J, Schultz AB. Loads transmitted during lumbosacral spinal manipulative therapy. Spine 1997;22:1955-64.
10. Kawchuk GN, Elliott PD. Validation of displacement measurements obtained from ultrasonic images during indentation testing. Ultrasound Med Biol 1998;24:105-11.
11. Kawchuk GN, Fauvel OR, Dmowski J. Ultrasonic indentation: a procedure for the noninvasive quantification of force-displacement properties of the lumbar spine. J Manipulative Physiol Ther 2001;24:149-56.
12. Latimer J, Goodsel MM, Lee M, Maher CG, Wilkinson BN, Moran CC. Evaluation of a new device for measuring responses to posteroanterior forces in a patient population, part 1: reliability testing. Phys Ther 1996;76:158-65.
13. Latimer J, Lee M, Adams R, Moran CM. An investigation of the relationship between low back pain and lumbar posteroanterior stiffness. J Manipulative Physiol Ther 1996;19:587-91.
14. Latimer J, Lee M, Adams RD. The effects of high and low loading forces on measured values of lumbar stiffness. J Manipulative Physiol Ther 1998;21:157-63.
15. Shirley D, Ellis E, Lee M. The response of posteroanterior lumbar stiffness to repeated loading. Man Ther 2002;7:19-25.
16. Kawchuk GN, Kaigle AM, Holm SH, Rod FO, Ekstrom L, Hansson T. The diagnostic performance of vertebral displacement measurements derived from ultrasonic indentation in an in vivo model of degenerative disc disease. Spine 2001;26:1348-55.
17. Gál J, Herzog W, Kawchuk G, Conway PJ, Zhang YT. Movements of vertebrae during manipulative thrusts to unembalmed human cadavers. J Manipulative Physiol Ther 1997;20:30-40.
18. Gál J, Herzog W, Kawchuk G, Conway P, Zhang YT. Measurements of vertebral translations using bone pins, surface markers and accelerometers. Clin Biomech 1997;12:337-40.
19. Smith DB, Fuhr AW, Davis BP. Skin accelerometer displacement and relative bone movement of adjacent vertebrae in response to chiropractic percussion thrusts. J Manipulative Physiol Ther 1989;12:26-37.
20. Fuhr AW, Smith DB. Accuracy of piezoelectric accelerometers measuring displacement of a spinal adjusting instrument. J Manipulative Physiol Ther 1986;9:15-21.

21. Lee R, Evans J. Load-displacement-time characteristics of the spine under posteroanterior mobilization. Aust J Physiother 1992;38:115-23.
22. Nathan M, Keller TS. Measurement and analysis of the in vivo posteroanterior impulse response of the human thoracolumbar spine: a feasibility study. J Manipulative Physiol Ther 1994; 17:431-41.
23. Kaigle AM, Pope MH, Fleming BC, Hansson T. A method for the intravital measurement of interspinous kinematics. J Biomech 1992;25:451-6.
24. Mendel T, Wink CS, Zimny ML. Neural elements in human cervical intervertebral discs. Spine 1992;17:132-5.
25. Roberts S, Eisenstein SM, Menage J, Evans EH, Ashton IK. Mechanoreceptors in intervertebral discs. Morphology, distribution, and neuropeptides. Spine 1995;20:2645-51.
26. Jiang H, Russell G, Raso VJ, Moreau MJ, Hill DL, Bagnall KM. The nature and distribution of the innervation of human supraspinal and interspinal ligaments. Spine 1995;20:869-76.
27. McLain RF, Pickar JG. Mechanoreceptor endings in human thoracic and lumbar facet joints. Spine 1998;23:168-73.
28. McLain RF. Mechanoreceptor endings in human cervical facet joints. Spine 1994;19:495-501.
29. Cavanaugh JM, Ozaktay AC, Yamashita T, Avramov A, Getchell TV, King AI. Mechanisms of low back pain: a neurophysiologic and neuroanatomic study. Clin Orthop 1997; 335:166-80.
30. Cavanaugh JM, Ozaktay AC, Yamashita HT, King AI. Lumbar facet pain: biomechanics, neuroanatomy and neurophysiology. J Biomech 1996;29:1117-29.
31. Cavanaugh JM. Neural mechanisms of lumbar pain. Spine 1995;20:1804-9.
32. Stubbs M, Harris M, Solomonow M, Zhou B, Lu Y, Baratta RV. Ligamento-muscular protective reflex in the lumbar spine of the feline. J Electromyogr Kinesiol 1998;8:197-204.
33. Solomonow M, Zhou BH, Harris M, Lu Y, Baratta RV. The ligamento-muscular stabilizing system of the spine. Spine 1998;23:2552-62.
34. Solomonow M, Zhou BH, Baratta RV, Lu Y, Harris M. Biomechanics of increased exposure to lumbar injury caused by cyclic loading: part 1. Loss of reflexive muscular stabilization. Spine 1999;24:2426-34.
35. Solomonow M, He ZB, Baratta RV, Lu Y, Zhu M, Harris M. Biexponential recovery model of lumbar viscoelastic laxity and reflexive muscular activity after prolonged cyclic loading. Clin Biomech 2000;15:167-75.
36. Indahl A, Kaigle AM, Reikeras O, Holm SH. Interaction between the porcine lumbar intervertebral disc, zygapophysial joints, and paraspinal muscles. Spine 1997;22:2834-40.
37. Indahl A, Kaigle A, Reikeras O, Holm S. Electromyographic response of the porcine multifidus musculature after nerve stimulation. Spine 1995;20:2652-8.
38. Pickar JG, McLain RF. Responses of mechanosensitive afferents to manipulation of the lumbar facet in the cat. Spine 1995;20:2379-85.
39. Pickar JG, Wheeler JD. Response of muscle proprioceptors to spinal manipulative-like loads in the anesthetized cat. J Manipulative Physiol Ther 2001;24:2-11.
40. Colloca CJ, Keller TS, Gunzburg R, Vandeputte K, Fuhr AW. Neurophysiologic response to intraoperative lumbosacral spinal manipulation. J Manipulative Physiol Ther 2000;23:447-57.
41. Fuhr AW, Colloca CJ, Green JR, Keller TS. Activator methods chiropractic technique. St. Louis: Mosby;1997.
42. Osterbauer PJ, Fuhr AW, Hildebrandt RW. Mechanical force, manually assisted short lever chiropractic adjustment. J Manipulative Physiol Ther 1992;15:309-17.
43. Weiner BK, Fraser RD, Peterson M. Spinous process osteotomies to facilitate lumbar decompressive surgery. Spine 1999; 24:62-6.
44. Colloca CJ, Keller TS. Electromyographic reflex response to mechanical force, manually-assisted spinal manipulative therapy. Spine 2001;26:1117-24.
45. Colloca CJ, Keller TS, Gunzburg R. Neuromechanical characterization of in vivo lumbar spinal manipulation. Part II: neurophysiological response. J Manipulative Physiol Ther 2003;26:579-91.
46. Maigne JY, Guillon F. Highlighting of intervertebral movements and variations of intradiskal pressure during lumbar spine manipulation: a feasibility study. J Manipulative Physiol Ther 2000;23:531-5.
47. Keller TS, Colloca CJ, Beliveau JG. Force-deformation response of the lumbar spine: a sagittal plane model of posteroanterior manipulation and mobilization. Clin Biomech 2002;17:185-96.
48. Kawchuk GN, Fauvel OR, Dmowski J. Ultrasonic quantification of osseous displacements resulting from skin surface indentation loading of bovine para-spinal tissue. Clin Biomech (Bristol, Avon) 2000;15:228-33.
49. Colloca CJ, Keller TS, Seltzer DE, Fuhr AW. Mechanical impedance of the human lower thoracic and lumbar spine exposed to in vivo posterior-anterior manipulative thrusts. Dublin: The Royal College of Surgeons; 2000, p. 171.
50. Lee M, Svensson NL. Effect of loading frequency on response of the spine to lumbar posteroanterior forces. J Manipulative Physiol Ther 1993;16:439-46.
51. Shirley D, Lee M. A preliminary investigation of the relationship between lumbar posteroanterior mobility and low back pain. J Manipulative Man Ther 1993;1:22-5.
52. Lundberg G, Gerdle B. Correlations between joint and spinal mobility, spinal sagittal configuration, segmental mobility, segmental pain, symptoms and disabilities in female homecare personnel. Scand J Rehabil Med 2000;32:124-33.
53. Colloca CJ, Keller TS, Peterson TK, Seltzer DE. Comparison of dynamic posteroanterior spinal stiffness to plain film radiographic images of lumbar disc height. J Manipulative Physiol Ther 2003;26:233-41.
54. Burton AK, Battie MC, Gibbons L, Videman T, Tillotson KM. Lumbar disc degeneration and sagittal flexibility. J Spinal Disord 1996;9:418-24.
55. Colloca CJ, Keller TS, Seltzer DE, Fuhr AW. Muscular and soft-tissue contributions of dynamic posteroanterior spinal stiffness. Proceedings of the 2000 International Conference on Spinal Manipulation. Bloomington, MN: 2000. p. 159-60.
56. Shirley D, Lee M, Ellis E. The relationship between submaximal activity of the lumbar extensor muscles and lumbar posteroanterior stiffness. Phys Ther 1999;79:278-85.
57. Kaigle AM, Wessberg P, Hansson TH. Muscular and kinematic behavior of the lumbar spine during flexion-extension. J Spinal Disord 1998;11:163-74.
58. Kawchuk GN, Fauvel OR. Sources of variation in spinal indentation testing: indentation site relocation, intra-abdominal pressure, subject movement, muscular response, and stiffness estimation. J Manipulative Physiol Ther 2001;24:84-91.
59. Shirley D, Hodges PQ, Eriksson AE, Gandevia SC. Spinal stiffness changes throughout the respiratory cycle. J Appl Physiol 2003;95:1467-75.

60. Viner A, Lee M, Adams R. Posteroanterior stiffness in the lumbosacral spine. The correlation between adjacent vertebral levels. Spine 1997;22:2724-9.
61. Caling B, Lee M. Effect of direction of applied mobilization force on the posteroanterior response in the lumbar spine. J Manipulative Physiol Ther 2001;24:71-8.
62. Allison G. Effect of direction of applied mobilization force on the posteroanterior response in the lumbar spine. J Manipulative Physiol Ther 2001;24:487-8.
63. Allison GT, Edmondston SJ, Roe CP, Reid SE, Toy DA, Lundgren HE. Influence of load orientation on the posteroanterior stiffness of the lumbar spine. J Manipulative Physiol Ther 1998;21:534-8.
64. Edmondston SJ, Allison GT, Gregg CD, Purden SM, Svansson GR, Watson AE. Effect of position on the posteroanterior stiffness of the lumbar spine. Man Ther 1998;3:21-6.
65. Lee M, Kelly DW, Steven GP. A model of spine, ribcage and pelvic responses to a specific lumbar manipulative force in relaxed subjects. J Biomech 1995;28:1403-8.
66. Solinger AB. Theory of small vertebral motions: an analytical model compared to data. Clin Biomech 2000;15:87-94.
67. Keller TS, Colloca CJ. A rigid body model of the dynamic posteroanterior motion response of the human lumbar spine. J Manipulative Physiol Ther 2002;25:485-96.
68. Indahl A, Kaigle A, Reikeras O, Holm SH. Sacroiliac joint involvement in activation of the porcine spinal and gluteal musculature. J Spinal Disord 1999;12:325-30.
69. Mochida K, Komori H, Okawa A, Shinomiya K. Evaluation of motor function during thoracic and thoracolumbar spinal surgery based on motor-evoked potentials using train spinal stimulation. Spine 1997;22:1385-93.
70. Matsui H, Kitagawa H, Kawaguchi Y, Tsuji H. Physiologic changes of nerve root during posterior lumbar discectomy. Spine 1995;20:654-9.
71. Matsui H, Kanamori M, Kawaguchi Y, Kitagawa H, Nakamura H, Tsuji H. Clinical and electrophysiologic characteristics of compressed lumbar nerve roots. Spine 1997;22:2100-5.
72. Hormes JT, Chappuis JL. Monitoring of lumbosacral nerve roots during spinal instrumentation. Spine 1993;18:2059-62.
73. Sine RD, Merrill D, Date E. Epidural recording of nerve conduction studies and surgical findings in radiculopathy. Arch Phys Med Rehabil 1994;75:17-24.
74. Kang YM, Choi WS, Pickar JG. Electrophysiologic evidence for an intersegmental reflex pathway between lumbar paraspinal tissues. Spine 2002;27:E56-E63.
75. Holm S, Indahl A, Solomonow M. Sensorimotor control of the spine. J Electromyogr Kinesiol 2002;12:219-34.
76. Symons BP, Herzog W, Leonard T, Nguyen H. Reflex responses associated with activator treatment. J Manipulative Physiol Ther 2000;23:155-9.
77. Keller TS, Colloca CJ. Mechanical force spinal manipulation increases trunk muscle strength assessed by electromyography: a comparative clinical trial. J Manipulative Physiol Ther 2000;23:585-95.
78. Hawk C, Azad A, Phongphua C, Long CR. Preliminary study of the effects of a placebo chiropractic treatment with sham adjustments. J Manipulative Physiol Ther 1999;22:436-43.
79. Colloca CJ. Articular neurology, altered biomechanics, and subluxation pathology. In: Fuhr AW, Colloca CJ, Green JR, Keller TS, editors. Activator methods chiropractic technique. St. Louis: Mosby Year-Book, Inc.; 1997. p. 19-64.
80. Wyke B. Articular neurology and manipulative therapy. In: Glasgow E, Twomey L, Scull E, Kleynhans A, Idczak R, editors. Aspects of manipulative therapy. New York: Churchill-Livingstone; 1985. p. 72-7.
81. Herzog W, Scheele D, Conway PJ. Electromyographic responses of back and limb muscles associated with spinal manipulative therapy. Spine 1999;24:146-52.
82. Dishman JD, Bulbulian R. Spinal reflex attenuation associated with spinal manipulation. Spine 2000;25:2519-25.
83. Dishman JD, Bulbulian R. Comparison of effects of spinal manipulation and massage on motoneuron excitability. Electromyogr Clin Neurophysiol 2001;41:97-106.
84. Dishman JD, Ball KA, Burke J. First prize-central motor excitability changes after spinal manipulation: a transcranial magnetic stimulation study. J Manipulative Physiol Ther 2002;25:1-9.