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ORIGINAL ARTICLE

Mechanical properties of a thoracic spine mannequin with variable stiffness control*

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Objective: To test the posterior-to-anterior stiffness (PAS) of a new thoracic spine training simulator under different conditions of "fixation."

Methods: We constructed a thoracic spine model using plastic bones and ribs mounted in a wooden box, with skin and soft tissue simulated by layers of silicone and foam. The spine segment could be stiffened with tension applied to cords running through the vertebrae and ribs. We tested PAS at 2 tension levels using a custom-built device to apply repetitive loads at the T6 spinous process (SP) and over adjacent soft tissue (TP) while measuring load and displacement. Stiffness was the slope of the force-displacement curve from 55 to 75 N.

Results: Stiffness in the unconstrained (zero tension) condition over the SP averaged 11.98 N/mm and 6.72 N/mm over the TP. With tension applied, SP stiffness increased to 14.56 N/mm, and TP decreased to 6.15 N/mm.

Conclusion: Thoracic model compliance was similar to that reported for humans. The tension control system increased stiffness by 21.3% only over the SP. Stiffness over the TP was dominated by the lower stiffness of the thicker foam layer and did not change. The mannequin with these properties may be suitable for use in manual training of adjusting or PAS testing skills.

Key Indexing Terms: Palpation; Manipulation, Spinal; Spine; Chiropractic; Education

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INTRODUCTION

The hallmark of chiropractic care is the application of forces into the spine and pelvis, the high-velocity low-amplitude (HVLA) manual method in particular. This is usually referred to as spinal adjustment or spinal manipulation. There is some controversy over the proper way to apply the loads, and many techniques exist.[1] Careful control of the magnitude and speed of thrusts are needed to ensure the safety of the procedure for both the doctor and the patient. A significant portion of the curricula in chiropractic schools is devoted to learning the diagnostic and psychomotor skills needed to deliver HVLA manipulation thrusts in a safe and effective manner.[2]

Safety is a concern for patients and doctors of chiropractic (DC) themselves, who occasionally experience practice-related injuries. There is disagreement among studies as to whether injuries are more common for the low back, hands and fingers, shoulders, wrists and elbows, neck, or mid-back.[3–9] Many studies have emphasized some DC work activities as being labor-intensive[10] and repetitive[9,11] such that injury-related complaints involve cumulative trauma,[9] chronic overuse,[8] or "body stressing".[12] Some studies of work-related injuries for physical therapists[13,14] and osteopaths[12,15] have reported similar complaints and causative factors.

The performance of side-posture manipulation procedures gets much of the blame, but injuries during thoracic spine manipulation and other patient-positioning maneuvers have been reported.[4,5,7,8,11,16] For many DCs, injuries occur early in their careers, even while in chiropractic college.[5,16,17] Hodgetts and Walker[10] commented that injuries to DC students and early career practitioners could have long-term consequences on their ability to perform professionally and on their physical, emotional, and financial well-being. The reduction of injury potential is important for individuals and the chiropractic profession in general. Williams et al[18] found that DCs' self-reported

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burnout had a significant association with whether they had had a work-related injury.

Thus far, the above discussion has addressed DCs in the role of providing patient care, but a number of studies have reported students being injured while playing the role of patient in chiropractic technique classes.[6,16,17,19] There are, therefore, some advantages to having students' early education in delivering adjustive thrusts to involve inanimate models rather than solely with living humans. Historically, chiropractic educators have used a variety of training tools ranging from simple automobile tires to so-called speeder-boards to more advanced electronic devices, such as the Dynadjust.[20] There has been an upsurge recently of mannequin use in medical and nursing education. From the classic cardiopulmonary resuscitation (CPR) dummy to more advanced electronic devices, mannequins are used to simulate clinical examination, prostate exams, obstetric procedures, and surgery.[21] Use of a mannequin in the training of chiropractors and other manual therapists would allow for numerous thrusts by multiple novice student adjustors under controlled conditions without concern for injury by the recipient.

It seems intuitive that inanimate models used for chiropractic technique education and research should be as much like a human as possible, at least in their force-displacement properties, to provide the best practice for adjusting humans. Passmore et al[22] used bicycle inner tubes inflated to varying intensities in a recent study to show that practitioners modulate the force and speed of their manipulative thrusts based on perceptions of the resistance to deformation of the material into which thrusts are delivered. Hence, it is important to practice adjustments on devices with force-displacement properties similar to the range seen in human paraspinal tissues and perhaps with the ability to simulate changes in compliance.

Lifelike mannequins have been developed for teaching spinal adjustment skills. Chapman used a FlexiMan cervical spine model in a study of cervical spine adjusting.[23,24] Descareaux et al have used a specially modified CPR-style dummy with a force transducer supporting the thoracic section in several studies of chiropractic education.[25–30] Their device provided feedback on the force applied and also incorporated varying stiffness and a breakaway feature by setting the system to release an electromagnetic lock in the support structure when a certain force threshold was reached.

In a study of DC students' progress over 10 weeks, Owens et al[31] used a mannequin composed of a plastic spine and pelvis model enclosed in high-density upholstery foam to simulate a human torso. The same model was used with experienced DCs who were also spinal adjustment technique instructors.[32] Those participants disagreed about whether the mannequin was more than, less than, or similar in compliance to that of a human being. It is clear, however, that the model was not very humanlike. It may be relevant that 2 participants noted hand and arm soreness or fatigue toward the end of their testing session, and 1 reported shoulder pain a few days later.[32]

Several studies have used a Human Analog Mannequin as part of a force-measurement system produced by the Canadian Memorial Chiropractic College.[33–35] The mannequin developers report that the mannequin has properties similar to the human torso but tested only the soft foam covering over the thoracic section, not the compliance of the undercarriage.[36] Our team has received anecdotal reports suggesting that the mannequin is too rigid for regular practice, and other objects are substituted in practice sessions.

At the time of this writing, there is a need for a high-fidelity mannequin for use in technique training. We have reported previously on progress being made in this area, showing efforts to simulate spinal fixation[37] and the results of compliance tests on a lumbar spine model.[38] As part of this large mannequin design effort, the objective of this study was 2-fold: (1) develop a thoracic spine mannequin with lifelike properties, including a method for influencing spinal stiffness, and (2) test the stiffness of the mannequin in comparison to human spines and the control system's influence on stiffness.

METHODS

Mannequin Design

We started with certain design criteria to produce a lifelike mannequin. We wanted the shape and contour to be like a human torso. Another criterion was to have skin and overlying tissues provide a realistic quality to the mannequin. The mannequin also needed to have palpable simulation for anatomical landmarks commonly contacted during evaluation and treatment. We also wanted the stiffness/compliance of covering tissues and undercarriage to be similar to human spines and to have movable joints with realistic kinematics. We also wanted to include a rib cage needed to contribute to structure and for the mannequin to be durable enough to withstand repeated thrusts of up to 1000 N.

The early prototype used the thoracic section (T2–T9) of a typical plastic spine found commonly in school bookstores or online retailers (artificial skeletons, 3B Scientific GmbH, Hamburg, Germany). We disassembled the spine, taking out the central steel rod and replacing it with a more flexible ¼-in elastic (bungee) cord connecting all 8 vertebrae with the original plastic intervertebral discs included. We drilled holes in each vertebra at approximately the location of the costo-vertebral joint on the lateral edge of the vertebral body. Then 2-in sections of polyethylene tubing (outer diameter = ¼ in) were inserted and glued into the holes. Each of these tube sections was inserted into a longer polyethylene tube with a larger diameter (inner ¼ in and outer 3/8 in.) These formed the ribs. We cut out a curved contour resembling a thoracic curve from ¼-in plywood and drilled grooves in the upper edge to hold each rib in place. Plywood cross braces supported the frame (Fig. 1)

A 1/8-in-diameter bungee cord was strung through the whole rib/vertebra assembly. A system of knots in the bungee cord passed through keyholes on the anterior edge of the plywood frame to apply tension to the cord. The bungee tension was increased by pulling more bungee through the keyhole and locking it down with another

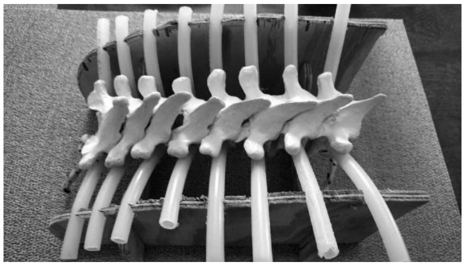

Figure 1 - Photograph of the model constructed of plastic thoracic vertebrae T2–T9 and a rib structure of polyethylene tubing supported in a plywood carriage.

knot farther up on the cord. The bungee tension then could be controlled to any particular vertebra independently and 1 side or the other.

To create the skin and simulated soft tissue, we first sculpted a contoured cavity using modeling clay. The clay was lined with cellophane, and the silicone skin layer was laid down in it (Moldstar 16 Silicone, Smooth-On, Inc, Lower Macungie, PA). We then inverted the spinal model and positioned it in the mold on top of the hardened silicone. Finally, a premixed liquid expanding foam (FlexFoam-iT! V, Smooth-On, Inc) was poured into the mold and allowed to expand to invest the rib/vertebrae assembly. The foam adhered to the silicone layer very well, making a combined soft tissue/skin layer. The mold contours were such that the soft tissue layer was very thin over the vertebral spinous processes, approximately 1 in thick over the transverse processes and thin again at the outer margins, where the ribs were supported by wood.

Testing Apparatus

To measure the compliance of the mannequin construction, we needed to be able to apply loads to specific parts of the model and measure the displacement produced. The system we devised used a wooden frame with joints constructed and pinned such that 1 end could be mounted firmly to an adjusting bench and the other, the "head" end, would be freely movable but would maintain a vertical orientation (Fig. 2). The location of the loading point (P_5 in the figure) was a simple function of the member lengths and the angles between them, as shown by the equation in the figure.

To track the frame position, we used inertial measurement units (IMU) (Myomotion, Noraxon USA, Inc, Scottsdale, AZ) placed on the movable arms. The IMUs provided wireless location tracking when monitored by the MR 3.10 software (Noraxon USA, Inc). Specifically, the software was set up to provide the angles α and β shown in the diagram. The IMU system was capable of measuring angular motion with an accuracy of 0.25°. We tested the system against blocks of wood with known thickness and found the accuracy of the displacement measure to be 0.5 mm.

To measure force, the spinal model was placed directly on a force plate (Model FP4550, Bertec, Columbus, OH). We applied forces manually by pressing the free arm of the testing frame against the model through a 1-cm round disk.

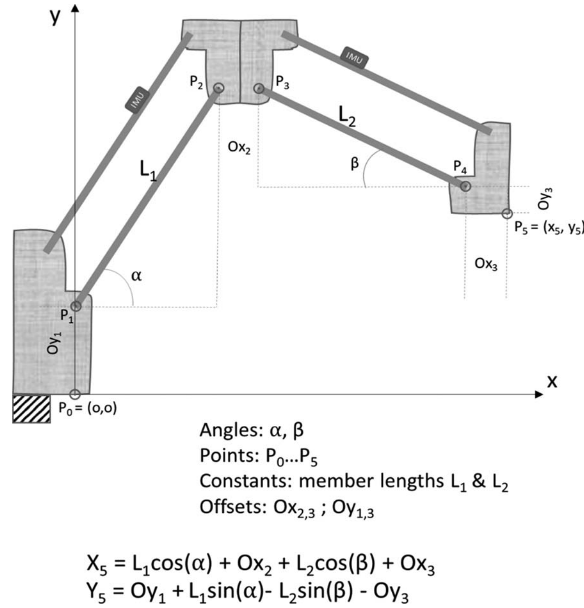

Figure 2 - Schematic of the stiffness testing apparatus. It is built to allow free movement of the arms, but the headpiece at P5 will always remain parallel to the base at P1. Inertial measurement units on the upper arms measure angular motion and allow calculation of the movement of P5 with respect to P1.

While the testing system may seem crude in comparison to computer-controlled systems reported in the literature,[39] it does bear some resemblance to the system used at the Palmer College of Chiropractic to measure stiffness of spinal tissues in low back pain patients.[40] It had the added advantage of making use of equipment already available in the research lab.

Testing Procedures

Stiffness testing consisted of pressing the free arm of the testing apparatus (at P_5 in Fig. 2) against 1 of 2 locations on the model and applying 10 cycles of loading and unloading manually. The operator could observe the loading trace on a computer screen to allow for smooth loading with 1-second cycles and a target maximum load of 100 N. Loads were applied directly over the T6 spinous process and to the foam over the transverse process. Testing was first performed with all the elastic cords in the unloaded state, followed by loading of all cords to a moderate level of tension (2 in of stretch on each 1/8-in-diameter bungee cord). The MR software was set up to record angular motions and force plate loads at a rate of 1500 Hz. Each run of 10 loading cycles was stored as a single file.

Stiffness Calculations

We developed a custom application in Excel (Microsoft Corp, Redmond, WA, USA) with routines to automatically locate the portions of each loading trace where the

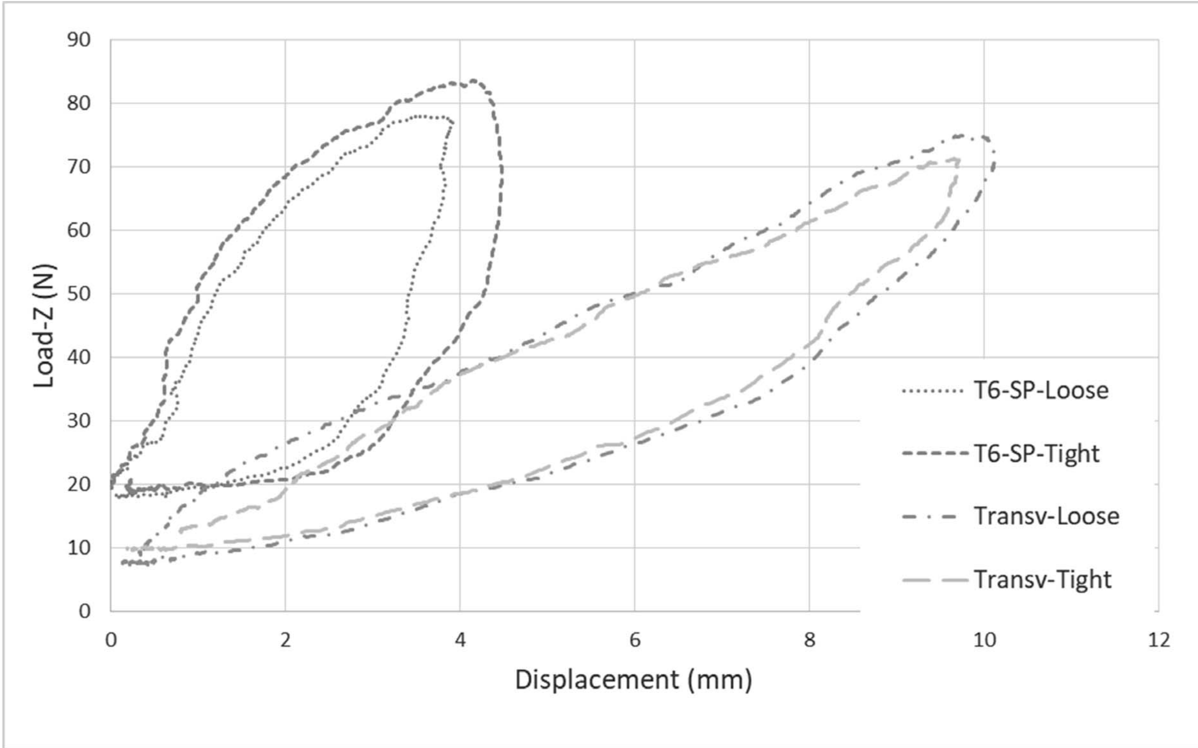

Figure 3 - Plots of 1 loading unloading cycle from each of the 4 test conditions.

force ranged between 55 and 75 N. Mimicking a method used to calculate posterior-to-anterior stiffness (PAS) in the lumbar spine,[40] the software calculated the least-squares slope of that segment to determine stiffness in the form of N/mm.

Loading rate and maximum load per cycle were also calculated. We discarded the first cycle of each loading sequence to avoid artifacts that may have occurred due to shifts in the contact. The final stiffness determination was based on the average of those remaining cycles where the loading rate was between 50 and 100 N/s and the maximum load exceeded the 75 N required to create the calculable 55–75 N force range.

RESULTS

In total, there were 40 cycles of loading/unloading, given 2 conditions of cord tension and 2 locations tested, including 4 discarded cycles as stated above. The stiffness calculations were successful in 25 of the 36 remaining cases. An additional 11 cycles were eliminated due to poor quality in either the force or displacement data or low maximum force that made the stiffness calculation inaccurate.

Plots of load versus displacement for the stiffness tests show hysteresis effects typical of load-displacement curves seen in human spinal stiffness tests.[40] Figure 3 shows plots of 1 characteristic loading/unloading cycle for each of the 4 location/tension conditions tested. Each plot forms a loop where increasing displacement during indentation induces increasing load. Then, when the displacing probe is retracted, the load decreases more rapidly than it did under indentation. This is evidence that some potential energy in the system is lost to friction or viscoelastic effects.

The load-displacement relationship is quite different between occasions when the spinous process (T6–SP) is contacted compared to the transverse process. Force rises much more quickly during indentation over the T6–SP, as indicated by the more steeply inclined traces in the plot. This is evidence that the stiffness over T6–SP is greater than the transverse process. The character of the plots is also somewhat different in that the shape of the curve is convex for T6–SP traces and slightly concave for the transverse process.

The plots comparing the "loose" (no tension on bungees) and "tight" (moderate bungee tension) show a difference in traces of T6–SP stiffness but very little difference over the transverse process. The average stiffness at each location/loading condition is shown in Table 1.

Table 1 - Average Stiffness at Each Location and Tension Condition

	Average Stiffness (N/mm)	
	Spinous	Transverse
No tension	11.98	6.75
Moderate tension	14.56	6.16
% Change	21.5%*	–8.7%

* $p < .001$.

DISCUSSION

Both goals of the study were accomplished in that the stiffness testing was successful in 2 conditions of model tension. Our measures of thoracic spine model stiffness are in the range of those found in living humans. While there is a considerable literature base on the stiffness of the thoracic spine in cadavers,[41] less is found on living humans. Most of the literature on in vivo human spinal stiffness has been done in the low back, but there are a few that have measured thoracic spine stiffness. Edmonston et al[42] used a motorized indenter to measure thoracic spine stiffness of pain-free individuals. Patients were placed on a lightly padded surface during testing. They measured stiffness at the T7 spinous process to average 10.7 N/mm, similar to our finding on the spinous process of the model with no tension applied. They also flipped their study subjects over and measured global rib cage stiffness by using the indenter on the subject's sternum. They measured the rib cage as 7.6 N/mm and concluded that the variation in rib cage stiffness contributes as much as 33% of the variation in spinal stiffness. This observation informed our decision to include riblike structural supports in our model.

Pagé et al[43] used a motorized indenter to measure thoracic spine stiffness in pain-free individuals and those with thoracic pain. Stiffness over the T5–T7 spinous processes was in the range of 7–10 N/mm. They used a similar method for calculating stiffness as we did but tested the patients in the range of only 10–40 N, somewhat lower than the test we used.

Applying tension to the bungees in our model has the effect of decreasing the sliding motion of the tubes that support the spine within the wooden supporting box. Decreased motion during indentation should lead to increased stiffness. It is interesting that applying tension increased the spinous process stiffness by 21% but did not affect the stiffness over the transverse process. We suspect that this is due to the thickness of the foam layer over the transverse process. The foam is quite a bit softer than the thin layer of silicone over the spinous processes. This fact is evident when comparing the stiffness over the TP to that over the SP (6.75 versus 12 N/mm). Apparently, any change in stiffness due to tension in the bungee system is masked by the softness of the foam layer.

The observation that the load-displacement curve over the T6–SP is convex rather than the more typical concave shape is interesting. The model is a complex composite of materials and support structures, making interpretation difficult. Anatomically, the thoracic spine and this model are different from the lumbar spine because the thoracic spinous processes significantly overlap each other. We speculate that early increased stiffness is high because the T6 spinous process is compressed onto the root of the T7 spinous process quickly during displacement under loading. We have not seen similar plots of load displacement for the thoracic spine in the literature.

Kawchuk et al[44] compared the ability of human palpators to detect changes in spinal stiffness to an indenter machine. When testing the stiffness of inflated devices, clinicians were able to detect an 8% change in stiffness on average. Those same clinicians were very good at detecting large differences when comparing stiffness in the thoracic versus lumbar spines of human subjects but not able to detect smaller differences between individual lumbar segments.

Clinicians tested stiffness by pressing on the spinous processes of the subjects' spines, similar to our indenter tests on the spinous process of the mannequin. Kawchuk et al[44] did not report the actual magnitudes of the stiffness measured in either the inflated test device or the human spines but reported only relative magnitudes. Since we could vary the stiffness of our mechanical thoracic spine by as much as 21%, the change should be detectable by a trained clinician.

Future Research

The current thoracic model is to be a subsection of larger full-spine mannequin. When that construction is completed, we will repeat the tests to see if the more anthropic version can produce the same changes in stiffness. We can also vary the tension in the fixation system to decrease the percent change in stiffness, providing a more graded stiffness control system. Such a system can be used in studies with students to show if training methods using the device improve clinicians' ability when performing spinal stiffness measures on human patients.

Limitations

We have compared the stiffness of our model to measures found in the literature using similar but not identical methods and equipment. Measurement methods and equipment have been found to influence results on humans.[45] A better comparison could be made if we used our measurement device on humans in the same setting as the model. Our measures are on the high end of the range seen in the literature. It could be that the model is stiffer, or it could be that the conditions did not match those used in previous studies. For instance, we had no padding between the model and force plate, whereas the human studies use a more conformable treatment bench to support the test subjects. Extra foam padding would have the effect of reducing measured stiffness.

CONCLUSION

The compliance of our thoracic spine mannequin model is similar to that previously measured in humans. The tension control system increased stiffness by 21.3% but only over the SP. Stiffness over the TP is dominated by the lower stiffness of the thicker foam layer. The mannequin with these properties, with further refinement and testing, may be suitable for use in manual training of adjusting or PAS testing skills.

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Author Contributions

Concept development: EO, RH, BR. Design: EO, RH, BR. Supervision: EO. Data collection/processing: EO, RH, BR. Analysis/interpretation: EO. Literature search: EO, RH, BR. Writing: EO, RH, BR. Critical review: EO, RH, BR.

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