JRRD
Volume 42, Number 2, Pages 167–174
March/April 2005
Journal of Rehabilitation Research & Development

Bending stiffness of the lumbar spine subjected to posteroanterior manipulative force

Raymond Y. W. Lee, PhD;[1*] Bonnie Y. S. Tsung, MPhil;[1] Pin Tong, PhD;[2] John Evans, PhD[3]
[1]Department of Rehabilitation Sciences, The Hong Kong Polytechnic University, Hong Kong; [2]Department of Mechanical Engineering, The Hong Kong University of Science and Technology, Hong Kong; [3]Centre for Rehabilitation Science and Engineering, Queensland University of Technology, Queensland, Australia

Abstract—This study measured the bending stiffness of the spine when it is subjected to posteroanterior mobilization force. The lumbar spine was modeled as an initially curved beam column supported over the rib cage and the pelvis. Posteroanterior mobilization was assumed to be three-point bending of the beam. The mobilization force was measured by the mounting of a force plate onto the manipulation couch, where electromagnetic sensors measured the change in spinal curvature. The bending stiffness of the spine was derived from the force and curvature data. The technique developed in this study provided highly repeatable data. The theoretical analysis suggests that the pelvic rotation produced by mobilization may be used clinically to indicate the magnitude of the mobilization force. Future research may employ the present method to determine how back pain may affect the bending stiffness of the spine. The bending stiffness values reported in this study will be valuable to future modeling work.

Key words: back pain, beam analysis, biomechanics, manipulation, physiotherapy, posteroanterior mobilization, rehabilitation, spine, stiffness, three-point bending.

INTRODUCTION

Manipulative techniques are commonly employed in the clinical assessment and treatment of low back pain [1]. Posteroanterior mobilization, a highly popular technique, generally involves the application of vertical forces over the spinous process of a given vertebra while the subject is lying prone (**Figure 1**) [2]. Several research studies examined the mechanical characteristics of posteroanterior mobilization [3–11]. In some of these studies, posteroanterior forces were delivered by a mechanical mechanism and measured by a load cell [6–8]. Other researchers measured the forces by attaching strain gauges to a specially constructed couch on which the subject lay [3–4]. The intervertebral movements produced by mobilization have also been quantified by video and radiographic techniques [5–6,9–10].

For a full understanding of the mechanical response of the spine to mobilization force, measuring the forces and the movements produced simultaneously is essential. Lee and Evans successfully developed a technique for such measurements [6–8]. They found that the load-displacement characteristics of mobilization were nonlinear. The spine was shown to exhibit time-dependent properties such as creep and preconditioning [6–7]. The radiographic study of Lee and Evans showed that under the application of mobilization loads, the lumbar motion segments tended to extend [9]. They also showed that the

Abbreviations: ASIS = anterior superior iliac spine, ICC = intraclass correlation coefficient.

This material was based on work supported by the Hong Kong Research Grant Council, grant PolyU 5195/01E.

*Address all correspondence to Raymond Y. W. Lee, PhD; Department of Rehabilitation Sciences, The Hong Kong Polytechnic University, Yuk Choi Road, Hunghom, Hong Kong; 852-2766-4889; 852-2330-8656. Email: raymond.lee@polyu.edu.hk

DOI: 10.1682/JRRD.2004.01.0002

upper lumbar segments (L1/2–L3/4) translated posteriorly and the L5/S1 segment anteriorly. The L4/5 segment did not exhibit translation in a consistent direction and appeared to be a transitional segment. The movement pattern observed in this study strongly suggests that the spine is subjected to three-point bending under the application of mobilization loads.

Clinically, spinal stiffness is simply determined according to the magnitude of the movement detected or perceived by clinicians. This approach is unsatisfactory because it does not take into account the loads exerted on the spine. Previous research attempted to quantify posteroanterior stiffness by determining the absolute vertical displacement of the mobilized vertebra in space [11]. Such measurement was subject to large error because the vertebral displacement could be affected by factors such as the stiffness of plinth padding rather than the intrinsic stiffness of the spine [12].

Because previous definitions of spinal stiffness are rather imprecise, the effects of symptoms on spinal stiffness remain unclear. These issues must be addressed, because they will provide fundamental information on the mechanisms of action underlying therapy. Posteroanterior stiffness should not be defined simply by the vertical displacement of the spine without consideration of the geometry of the spine and the mechanical effects on the adjacent anatomical structures. One should interpret it as the bending stiffness of the spine that may be modeled as an initially curved beam under three-point bending.

Figure 1.
Application of posteroanterior mobilization over L4 spinous process of subject lying prone.

Previous studies measured the bending stiffness of the vertebral column during flexion, extension, lateral bending, and twisting of the trunk [13–14]. These researchers found that wearing a belt, holding the breath, and receiving anesthesia would significantly change spinal stiffness. Quantifying the bending stiffness of the spine when subjected to mobilization would be clinically useful, but no previous research has attempted to do this. Posteroanterior mobilization was found to induce extension moment and shear on the lumbar spine [6,8–10]. The loading conditions are rather complex, and the stiffness property of the spine under these conditions may be different from that during simple physiological loading.

Therefore, with this study, we developed a method for measuring the bending stiffness of the spine when subjected to posteroanterior mobilization.

METHODS

Theoretical Model

We used a two-dimensional beam-column model to study the mechanical response of the whole lumbar spine under the influence of posteroanterior mobilization loads. The lumbar spine was modeled as an initially curved beam column supported over the rib cage and the pelvis (**Figure 2**) [6,8,10]. The model assumed that no significant compressive forces existed along the spine. Yoon

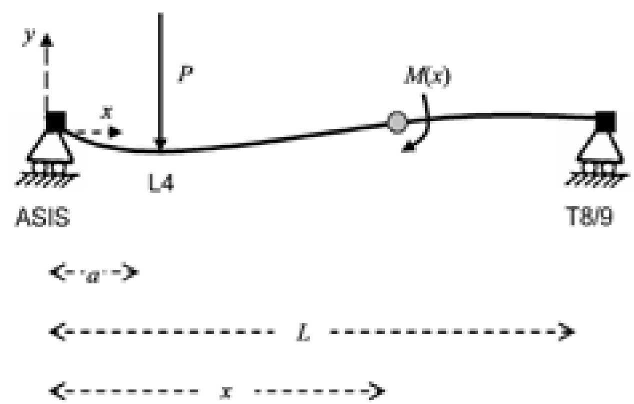

Figure 2.
Biomechanical model of posteroanterior mobilization. Lumbar spine is shown as initially curved beam supported over T8/9 and anterior superior iliac spine (ASIS). Distances *a* and *L* represent distances of point of application of mobilization force (P) and ASIS from T8/9. Solid black squares represent placement of electromagnetic sensors. $M(x)$ refers to moment acting on beam at point *x* (represented by gray circle).

and Mansour showed that the passive hip moment was negligible in the neutral position and with a small angle of rotation [15]. The pelvic support was considered to be a pin joint located effectively over the anterior superior iliac spine (ASIS). A force plate study was carried out that established the boundary condition at the thoracic end of the beam [6]. No significant change in moment was observed at the thoracic cage during the application of mobilization force. The thoracic end was therefore assumed to be a pin boundary. The effective point of support was found to be at T7/8 level.

According to the theory for an elastic beam with bending stiffness b [16], the change in curvature of the beam κ is given by

$$\kappa = \frac{M(x)}{b} , \qquad (1)$$

where $M(x)$ is the bending moment acting on the beam.

The bending moment acting at a point x on the right side of the force P (for $x \geq a$) (**Figure 2**) is given by

$$M(x)_{\text{right}} = Pa\left(1 - \frac{x}{L}\right) , \qquad (2)$$

where L is the distance between the two ends of the beam (T8/9 and ASIS), a is the horizontal distance between the ASIS and the point of application of the force P, and x is the horizontal distance from the ASIS.

The bending moment acting on the beam on the left side of the force P (for $x \leq a$) is given by

$$M(x)_{\text{left}} = Px\left(1 - \frac{a}{L}\right) . \qquad (3)$$

At the point of mobilization ($x = a$), both **Equations (2)** and **(3)** become

$$M(x)_{(x=a)} = Pa\left(1 - \frac{a}{L}\right) . \qquad (4)$$

Combining **Equations (1)** and **(4)**,

$$P = \frac{b}{a\left(1 - \frac{a}{L}\right)}\kappa . \qquad (5)$$

The variables P and κ may be measured experimentally with the use of force transducers and angular displacement sensors attached to the spine. The distances L and a may also be determined by a recording of the positions of the two ends of the beam and the point of force application. The bending stiffness of the spine when subjected to mobilization can then be determined.

According to **Equation (2)**, the differential equation for the right side of the beam (for $x \leq a$) is given by

$$b\frac{d^2y}{dx^2}_{\text{right}} = Pa\left(1 - \frac{x}{L}\right) , \qquad (6)$$

where y is the vertical distance from the ASIS.

Integration of **Equation (6)** yields

$$b\frac{dy}{dx}_{\text{right}} = Pax - \frac{Pax^2}{2L} + C_1 , \qquad (7)$$

where C_1 is a constant. Integration of **Equation (7)** gives

$$by_{\text{right}} = \frac{Pax^2}{2} - \frac{Pax^3}{6L} + C_1x + C_2 , \qquad (8)$$

where C_2 is a constant.

The differential equation for the left side of the beam (for $x \leq a$) is given by

$$b\frac{d^2y}{dx^2}_{\text{left}} = Px\left(1 - \frac{a}{L}\right) . \qquad (9)$$

First integration of **Equation (9)** yields

$$b\frac{dy}{dx}_{\text{left}} = \frac{Px^2}{2}\left(1 - \frac{a}{L}\right) + C_3 , \qquad (10)$$

where C_3 is a constant. Further integration gives

$$by_{\text{left}} = \frac{Px^3}{6}\left(1 - \frac{a}{L}\right) + C_3x + C_4 , \qquad (11)$$

where C_4 is a constant.

To solve the unknown coefficients C_1, C_2, C_3, and C_4, one must establish four boundary conditions:

$$y_{\text{left}(x=0)} = 0 , \qquad (12)$$

$$y_{\mathrm{right}(x=L)} = 0 \ , \qquad (13)$$

$$y_{\mathrm{left}(x=a)} = y_{\mathrm{right}(x=a)} \ , \qquad (14)$$

$$\frac{dy}{dx}_{\mathrm{left}(x=a)} = \frac{dy}{dx}_{\mathrm{right}(x=a)} \ . \qquad (15)$$

Solving **Equations (7)**, **(8)**, **(10)**, and **(11)** with the boundary conditions in **Equations (12)** to **(15)** results in

$$C_1 = Pa\left(-\frac{L}{3} - \frac{a^2}{6L}\right) \ , \qquad (16)$$

$$C_2 = \frac{Pa^3}{6} \ , \qquad (17)$$

$$C_3 = Pa\left(-\frac{L}{3} + \frac{a}{2} - \frac{a^2}{6L}\right) \ , \text{ and} \qquad (18)$$

$$C_4 = 0 \ . \qquad (19)$$

At the pelvic (left) end of the beam ($x = 0$), according to **Equation (10)**,

$$b\frac{dy}{dx}_{\mathrm{left}(x=0)} = C_3 \ . \qquad (20)$$

Hence, for small rotation of the pelvis $(\theta - \theta_0)$ with θ_0 as the initial position of the pelvis,

$$b = \frac{C_3}{(\theta - \theta_0)} \ . \qquad (21)$$

Substituting **Equation (18)** into **Equation (21)**,

$$b = \frac{Pa\left(-\frac{L}{3} + \frac{a}{2} - \frac{a^2}{6L}\right)}{\theta - \theta_0} \ . \qquad (22)$$

Equation (22) defines the mathematical relationship between the bending stiffness of the spine, the magnitude of mobilization force, the geometry of the spine, and the magnitude of pelvic rotation. Therefore, the bending stiffness of the spine is shown to be proportional to the magnitude of the mobilization force but inversely proportional to the magnitude of pelvic rotation.

Experimental Study

Twenty subjects (12 men and 8 women, mean age = 20 ± 2 yr, mean height = 1.59 ± 0.09 m, mean weight = 67.5 ± 4.9 kg) agreed to participate in this study. They were in good health, with no history of back pain or leg pain that could be attributed to the back within the last 12 months. They were excluded if they had undergone previous back surgery, or had a fracture, dislocation, or any structural defects of vertebral structures.

A nonconductive force plate (4060-NC Bertec Corporation, Columbus, OH) was mounted underneath the manipulation couch. Subjects were requested to lie on the couch face down. The forces experienced by the force plate represented the forces exerted onto the lumbar spine. An electromagnetic tracking system (FASTRAK, Polhemus Navigation, Colchester, VT) was used to measure the change in curvature of the spine. Such a system could be adversely affected by the presence of metals or other conductive materials, and thus the force plate used in this study was nonconductive. Our earlier research showed that the system was highly reliable in recording spinal motion [17–18]. A physiotherapist with specialized training in manipulation and with more than 5 years clinical experience applied posteroanterior mobilization force to the L4 spinous process. The clinical technique (Grade III, which was clinically defined as large amplitude of oscillations at the end of the range), described by Maitland et al. [2], involved cyclic applications of vertical forces over the spinous process while the subject was lying prone, as shown in **Figure 1**. The physiotherapist was instructed to perform the technique that was deemed appropriate for Grade III. No instructions were given to control the rate of mobilization and the magnitude of force so that the technique would resemble how it would be carried out in a normal clinical situation.

Two electromagnetic sensors were attached with double adhesives to the skin overlying the T7/8 vertebral junction and the position of the spine that corresponded to the level of the ASIS. These sensors allowed for a determination of the change in curvature of the spine. The orientations of the sensors were initially recorded when no mobilization force was applied. These orientations defined the initial curvature of the spine. After mobilization had been performed for 30 s, data were recorded for three oscillatory cycles of mobilization.

A computer program was developed to acquire the force and motion sensor data. The force signal was low-pass filtered digitally with the use of a second-order Butterworth filter with a cutoff frequency of 6 Hz in both the forward and reverse directions so there would be zero phase distortion. Data were converted analog to digital (DT3001, Data Translation Inc., Marlboro, MA) and acquired by a personal computer at a sampling rate of 30 Hz. The motion sensor data were acquired via the serial port at 115.2 kB with the same sampling rate as the force data (that is, 30 Hz per sensor). The force and motion sensor data were synchronized with the use of an external pulse to trigger data collection of both signals at the same instant.

The positions of T7/8 junction, the ASIS, and the L4 spinous process (where the mobilization load was applied) were recorded by a stylus attached to an electromagnetic sensor. The distances a and L can be determined from these digitized points. All measurements were repeated three times so that the reliability of the data obtained could be assessed.

Analysis of Experimental Data

The mobilization force (P) was plotted against the change in curvature of the spine. Data were fitted with a linear regression equation with the use of the least squares method. According to **Equation (5)**, the bending stiffness of the spine (b) could be derived from the slope of the regression line:

$$b = \text{slope} \times a\left(1 - \frac{a}{L}\right) \quad . \qquad (\mathbf{23})$$

An intraclass correlation coefficient (ICC) was used for the examination of the repeatability of the bending stiffness data obtained in the three measurement trials.

RESULTS

The results of this experiment are presented in the **Table**. Three cycles of mobilization force and the change in spinal curvature of one of the subjects (subject 20, 19 yr old, male, weight = 65.6 kg, height = 1.55 m) is shown in **Figure 3**. The force and movement data generally follow a sinusoidal pattern, and this was observed in all subjects. The mean maximum posteroanterior force of all subjects was found to be 178 ± 30 N (range = 141–273 N), and the mean frequency of mobilization was 1.2 ± 0.6 Hz (range = 0.3–2.5 Hz).

Figure 4 shows a typical plot of the mobilization force against the change in curvature (subject 20). The relationship was almost linear, as predicted by **Equation (5)**. The line of best fit, as obtained by linear regression, and the bending stiffness value, which was derived from the slope of the line, are also shown in **Figure 4**. The force-curvature plot was found to follow a linear relationship in all subjects. The **Table** shows the slopes and bending stiffness of all subjects, together with their summary statistics. The mean bending stiffness of the spine was found to be 15.1 ± 4.3 Nm^2 (range = 8.7–25.9 Nm^2). The mean ICC was found to be 0.98 ± 0.01 (range = 0.97–0.99) (**Table**), showing that the stiffness values derived in the three measurements were highly similar. The measurement method was highly reliable.

DISCUSSION

This paper reports a new method of measuring the bending stiffness of the lumbar spine when subjected to posteroanterior mobilization. The method, found to provide highly repeatable data in all the subjects as demonstrated by the high ICCs, could therefore be recommended, with confidence, for clinical uses and future research studies. The **Table** shows that the experimental observations were highly consistent among all the subjects examined. This would enable researchers to draw conclusions on the mechanical properties of the spine in a group of young healthy subjects. The bending stiffness data reported may be used as reference values for comparison with those of an older population and subjects with low back pain.

The bending stiffness values obtained in this study were similar to those reported in previous work [13–14]. However, we should note that previous work examined the stiffness of the spine when subjected to physiological loadings (flexion/extension, lateral bending and twisting) rather than posteroanterior mobilization. Our study also revealed a large variation in the bending stiffness values among the subjects. This observation was also consistent with the results of previous studies [13–14].

The physiotherapist was asked to perform the mobilization in a manner that would be appropriate according to his clinical experience. Large amplitudes of sinusoidal oscillations of force and movement signals were observed, reflecting the clinical definition of Grade III mobilization proposed by Maitland et al [2]. Similar force and movement patterns were reported in previous work [7,17]. The magnitude of the force and the frequency of mobilization were not controlled in this study

Table.
Results of experimental study. Personal characteristics of subjects (age, weight, and height), geometric data of trunk (*a* and *L*), slope of line of best fit, bending stiffness, maximum mobilization force, frequency of mobilization, and intraclass correlation coefficients (*R*) are shown for each subject. Summary statistics of variables are provided in last four rows of table.

Subject No.	Age	Weight (kg)	Height (m)	*a* (m)	*L* (m)	Slope ($Nrad^{-1}$)	Bending Stiffness (Nm^2)	Maximum Mobilization Force (N)	Mobilization Frequency	*R*
1	20	67.1	1.6	0.072	0.257	390.5	20.2	154.6	1.9	0.99
2	18	61.2	1.4	0.068	0.237	378.3	18.3	159.8	2.0	0.97
3	21	62.5	1.55	0.067	0.241	232.2	11.2	167.9	0.9	0.98
4	18	78.2	1.78	0.074	0.266	311.4	16.6	187.7	1.0	0.99
5	22	69.9	1.62	0.068	0.239	199.5	9.7	169.5	1.9	0.99
6	24	72.4	1.64	0.076	0.264	334.5	18.1	188.8	0.7	0.98
7	21	69.2	1.52	0.069	0.250	335.5	16.8	157.1	1.2	0.98
8	19	63.2	1.54	0.071	0.251	171.2	8.7	155.0	2.5	0.97
9	20	68.3	1.65	0.066	0.240	327.1	15.7	273.2	1.7	0.97
10	19	65.7	1.66	0.069	0.244	266.7	13.2	148.3	1.5	0.99
11	19	73.3	1.72	0.073	0.263	275.6	14.5	229.1	0.9	0.98
12	21	73.9	1.74	0.053	0.247	333.5	13.9	194.4	2.0	0.97
13	23	65.2	1.52	0.062	0.223	245.5	11.0	178.4	1.3	0.97
14	20	62.1	1.55	0.062	0.227	353.0	15.9	187.0	0.3	0.98
15	19	65.9	1.52	0.064	0.226	221.7	10.2	172.5	1.2	0.97
16	22	63.4	1.59	0.070	0.249	345.6	17.4	176.1	0.8	0.99
17	24	75.4	1.67	0.069	0.246	211.2	10.5	141.1	0.4	0.98
18	21	64.1	1.57	0.070	0.245	389.6	19.5	150.3	0.5	0.98
19	18	62.7	1.5	0.067	0.235	304.2	14.6	190.5	0.8	0.99
20	19	65.6	1.55	0.095	0.267	423.4	25.9	187.1	1.2	0.98
Mean ± SD	20 ± 2	67.5 ± 4.9	1.59 ± 0.09	0.069 ± 0.008	0.246 ± 0.013	302.5 ± 71.4	15.1 ± 4.3	178.4 ± 30.4	1.2 ± 0.6	0.98 ± 0.01
Min	18	61.2	1.40	0.053	0.223	423.4	25.9	141.1	0.3	0.97
Max	24	78.2	1.78	0.095	0.267	171.2	8.7	273.2	2.5	0.99

a = horizontal distance between anterior superior iliac spine (ASIS) and point of application by mobilization force.
L = horizontal distance between two ends of beam (T8/9 and ASIS).
SD = standard deviation.

so that they could reflect the loading conditions in a clinical situation. The values obtained in this study were similar to those reported elsewhere [7–8].

Posteroanterior mobilization should not be interpreted as posteroanterior translation of one vertebra upon the other. Previous research attempted to define the bending stiffness of the spine using the vertical displacement of the spine at the point of force application [11–12]. This definition was unsatisfactory because this could be affected by the geometry of the spine and the stiffness of the plinth padding [12]. The method reported in this study employed a beam model that took into account the geometry of the spine (*a* and *L*). The bending stiffness of the spine was derived from the relationship between the mobilization force and the change in spinal curvature.

An interesting finding of the theoretical analysis is that, according to **Equation (22)**, if the bending stiffness of the spine remains constant, the magnitude of the mobilization force is proportional to the magnitude of pelvic rotation. Clinically, quantifying the forces applied to the spine so that treatment can be implemented accurately is important. However, load cells or force plate may not be

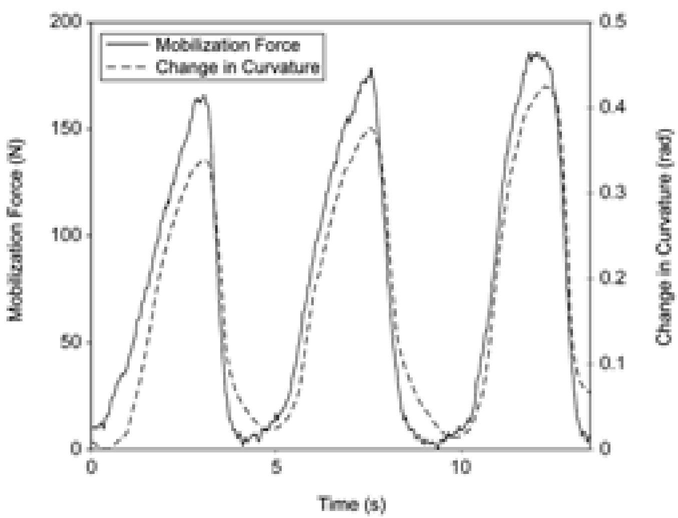

Figure 3.
Typical pattern of three cycles of mobilization force and changes in curvature of spine in subject 20.

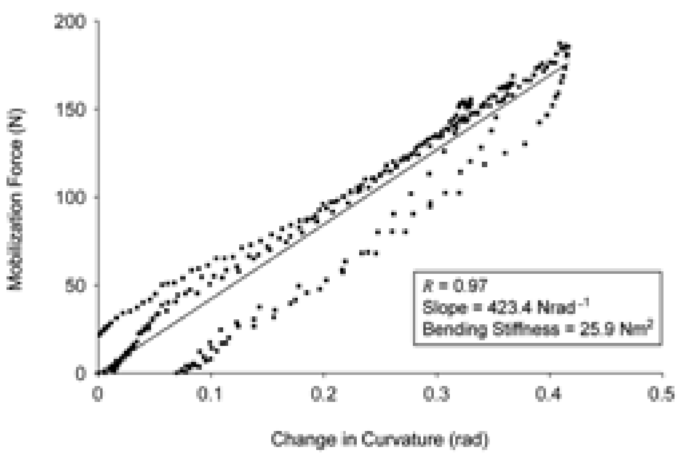

Figure 4.
Typical plot of mobilization force against change in curvature in subject 20. Line of best fit is also shown.

available in the clinical environment. In this case, a simple tilt sensor or an inclinometer attached to the sacrum may be employed for measurement of pelvic rotation, indicating the force applied. Such a simple device may also be used as a visual aid in the training of student therapists. However, we should note that the these arguments are based on the theoretical analysis only, and future research should examine the feasibility of using the sacral inclinometer in analyzing mobilization loads.

Equation (22) may also be used to determine the bending stiffness of the spine with knowledge of the geometry of the spine (*a* and *L*), the mobilization force applied, and the magnitude of pelvic rotation. In the clinical environment, all these variables can be easily measured with inexpensive devices. The mobilization force may be determined with the therapist standing on a bathroom scale. The distances *a* and *L* may be measured with a ruler and the pelvic rotation with an inclinometer as described previously. Future research should further explore this approach.

CONCLUSION

The method described in this paper is recommended for the measurement of the bending stiffness of the spine when subjected to posteroanterior mobilization. Bending stiffness should be derived from the relationship between the mobilization force and the change in spinal curvature. Clinically, medical staff may measure pelvic rotation to indicate the bending stiffness of the spine. In future studies, researchers may employ the present method to study various factors that may affect the bending stiffness of the spine—for instance, breathing and muscular contraction. The effect of back pain on the bending stiffness should also be investigated. The bending stiffness values reported in this study will be valuable to future modeling work.

ACKNOWLEDGMENT

We wish to thank Rachel Boyd for her assistance with data collection.

REFERENCES

1. Foster NE, Thompson KA, Baxter GD, Allen JM. Management of nonspecific low back pain by physiotherapists in Britain and Ireland. A descriptive questionnaire of current clinical practice. Spine. 1999;24:1332–42.
2. Maitland GD, Hengeveld E, Banks K, English K. Vertebral manipulation. 6th ed. London: Butterworth-Heineman; 2000.
3. Harms MC, Milton AM, Cusick G, Bader DL. Instrumentation of a mobilization couch for dynamic load measurement. J Med Eng Technol. 1995;19:119–22.
4. Triano J, Schultz AB. Loads transmitted during lumbosacral spinal manipulative therapy. Spine. 1997;22:1955–64.
5. Gal J, Herzog W, Kawchuk G, Conway PJ, Zhang YT. Movements of vertebrae during manipulative thrusts to unembalmed human cadavers. J Manip Physiol Therapeut. 1997;20:30–40.

6. Lee RYW. The biomechanical basis of spinal manual therapy [dissertation]. Glasgow (Scotland): University of Strathclyde; 1995.
7. Lee RYW, Evans JH. Load-displacement-time characteristics of the spine under posteroanterior mobilization. Aust J Physiother. 1992;38:115–23.
8. Lee RYW, Evans JH. Towards a better understanding of posteroanterior mobilization. Physiotherapy. 1994;80:68–73.
9. Lee RYW, Evans JH. An in-vivo study of the intervertebral movements produced by posteroanterior mobilization. Clin Biomech. 1997;12:400–8.
10. Lee RYW, Evans JH. The role of spinal tissue in resisting posteroanterior forces applied to the lumbar spine. J Manip Physiol Ther. 2000;23:551–56.
11. Lee M, Sevensson N. Measurement of stiffness during simulated spinal physiotherapy. Clin Phys Physiol Meas. 1990; 11:201–7.
12. Latimer J, Holland M, Lee M, Adams R. Plinth padding and measures of posteroanterior lumbar stiffness. J Manip Physiol Therapeut. 1997;20:315–19.
13. Scholten PJM, Veldhuizen AG. The bending stiffness of the trunk. Spine. 1986;11(5):463–67.
14. McGill S, Seguin J, Bennett G. Passive stiffness of the lumbar torso in flexion, extension, lateral bending, and axial rotation. Effect of belt wearing and breath holding. Spine. 1994;19(6):696–704.
15. Yoon YS, Mansour JM. The passive elastic moment at the hip. J Biomech. 1982;15(12):905–10.
16. Gere JM, Timoshenko SP. Mechanics of materials. Boston (MA): PWS Publishing Co; 1997.
17. Lee RYW. Kinematics of rotational mobilization of the lumbar spine. Clin Biomechan. 2001;16(6):481–88.
18. Lee RYW, Wong TKT. Relationship between the movements of the lumbar spine and hip. Hum Mov Sci. 2002; 21(4):481–94.

Submitted for publication January 5, 2004. Accepted in revised form September 17, 2004.