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Thoracic Kyphosis Affects Spinal Loads and Trunk Muscle Force

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Background and Purpose

Patients with increased thoracic curvature often come to physical therapists for management of spinal pain and disorders. Although treatment approaches are aimed at normalizing or minimizing progression of kyphosis, the biomechanical rationales remain unsubstantiated.

Subjects

Forty-four subjects (mean age [±SD]=62.3±7.1 years) were dichotomized into high kyphosis and low kyphosis groups.

Methods

Lateral standing radiographs and photographs were captured and then digitized. These data were input into biomechanical models to estimate net segmental loading from T2–L5 as well as trunk muscle forces.

Results

The high kyphosis group demonstrated significantly greater normalized flexion moments and net compression and shear forces. Trunk muscle forces also were significantly greater in the high kyphosis group. A strong relationship existed between thoracic curvature and net segmental loads (r=.85–.93) and between thoracic curvature and muscle forces (r=.70–.82).

Discussion and Conclusion

This study provides biomechanical evidence that increases in thoracic kyphosis are associated with significantly higher multisegmental spinal loads and trunk muscle forces in upright stance. These factors are likely to accelerate degenerative processes in spinal motion segments and contribute to the development of dysfunction and pain.

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The thoracic spine can be a source of pain and dysfunction for many individuals across the life span.[1-3] Pain in this region has been associated with reduced quality of life and functional capacity.[2,4,5] Compared with the cervical and lumbosacral spine, there is less research directed toward thoracic spine biomechanics. This may reflect a lower incidence of thoracic spine pain compared with cervical or lumbar pain, as well as the technical difficulties associated with experimentation in this area,[6] particularly in terms of anatomical complexity. The degree of morbidity associated with thoracic dysfunction, however, is likely to be comparable to other spinal levels.

The shape and design of the spine affords efficient distribution and balancing of body mass. There is minimal spinal muscle involvement required for maintaining static equilibrium in erect stance.[7] Changes in spinal shape, however, are likely to disrupt this balance. An increase in sagittal curvature may alter physiologic loading through the spine as a consequence of a shift in trunk mass, leading to increased flexion moments and compression and shear forces imposed on spine segments.[8] In addition to increased mechanical loading, changes in spinal posture may compromise back extensor strength (force-generating capacity)[9] and the normal function of paraspinal musculature,[10] perhaps due to alterations in length-tension relationships, moment arm lengths, and force vector orientations.[11,12]

A recent *in vivo* study demonstrated that an increase in lumbar flexion (stooping with hands on thighs or hands on knees) was associated with a 13% to 24% increase in L5–S1 compression force and peak bending moment, which was attributable to the erector spinae musculature and, to a lesser extent, to the abdominal musculature.[13] Altered load transmission through spinal motion segments is likely to contribute to fatigue and creep deformity and to changes in load transmission through the intervertebral disk (thereby potentially accelerating degenerative processes) and through the endplates and adjacent ligamentous network.[14-16]

The effects of naturally occurring thoracic kyphosis on segmental vertebral loading and the relationship between the magnitude of kyphosis and vertebral loading remain inadequately explored. These factors are particularly relevant to an elderly population, where the prevalence of increased thoracic kyphosis is greater. Previous studies[17,18] have examined thoracic mechanics and spinal curvature using simple anatomic models with input data derived from young populations; thus, the applicability of the findings to an older population remain uncertain.

A recent study evaluated the effect of voluntary anterior shifts in thorax position on intervertebral disk loads in a young population with normal thoracic curvature.[17] The adopted anterior thoracic posture was associated with significantly greater shear and compressive stresses imposed on the intervertebral disks. The anterior shift in thorax position (C7 relative to S1) by 81.5±39.2 mm did not significantly change the sagittal curvature of the spine. The anterior thorax posture reduced thoracic angle (kyphosis) by a mean of 13.1± 10.3 degrees. The effect of naturally occurring kyphosis on a loading profile of the spine has not been investigated *in vivo* in an elderly population using comprehensive biomechanical models. However, this remains an important consideration for musculoskeletal rehabilitation, considering the large number of conditions that affect the thoracic spine.

One of the shortcomings of previous biomechanical models is that moment and force estimations are limited to single lumbar motion segments and often model the trunk as one, nondeformable unit. Even studies that do consider multilevel loading continue to model the thoracic spine as a single unit and are limited to a narrow anthropometric range of participants who are usually healthy.[19,20] To achieve a more comprehensive understanding of spinal loading, multiple vertebral levels should be examined and multiaxial loading considered.

The aim of the current study was to evaluate the biomechanical effects of increased thoracic kyphosis on the loading profile of the thoracolumbar spine *in vivo* during upright stance. We hypothesized that an increase in naturally occurring thoracic kyphosis would significantly increase all segmental load parameters and trunk muscle forces and that a strong, positive relationship would exist between thoracic kyphosis and spinal load.

Method
Participants
Forty-four elderly participants (1 male, 43 female) with and without osteoporosis were recruited to provide heterogeneity in measures of thoracic kyphosis. This rationale has been used previously.[21] Based on bone densitometry classification criteria developed by the World Health Organization,[22] 31 participants had a diagnosis of osteoporosis, defined as a T score of less than −2.5. Participants were divided into 2 groups (high kyphosis [n=21], and low kyphosis [n=23]) based on a median split of kyphosis of 31.5 degrees measured between T4–9 using the vertebral centroid angle from lateral radiographs (Fig. 1) acquired in a standing position.[23,24] Back pain was assessed at the time of data collection using a 10-cm visual analog scale (VAS). The VAS scores ranged from 0 to 2 out of 10 and were not significantly different between the groups
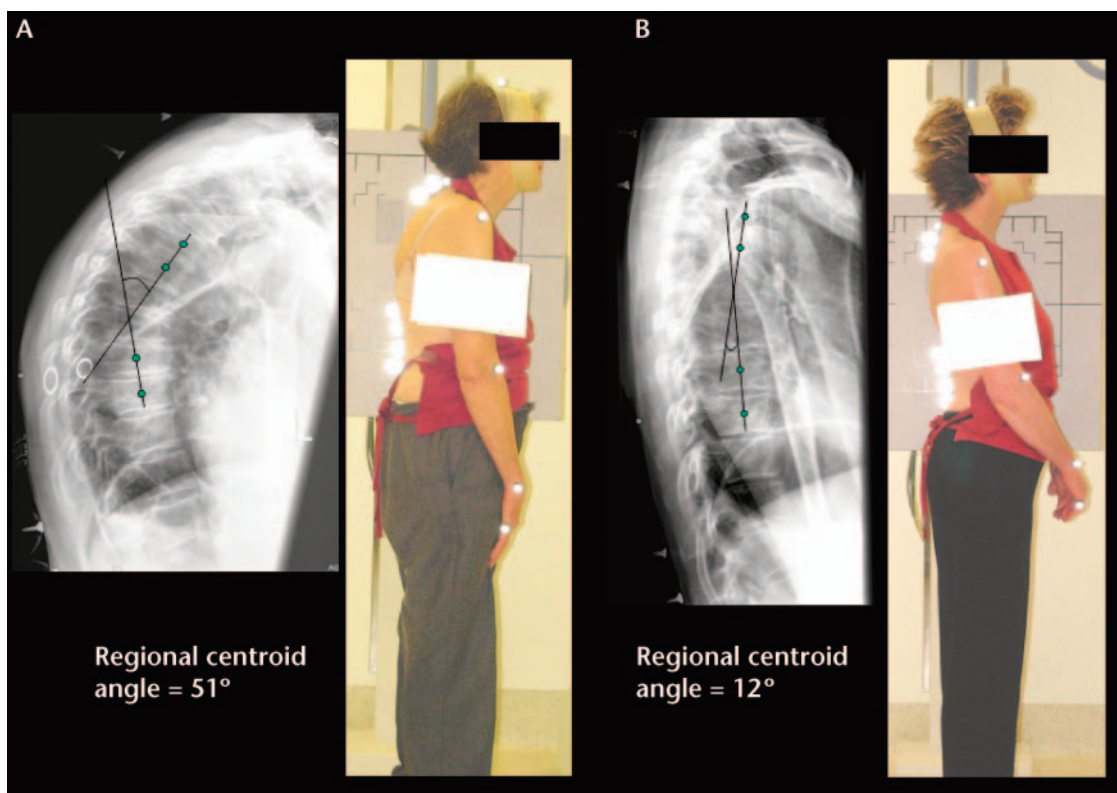
Figure 1.
Participants with (A) high kyphosis and (B) low kyphosis in a standing posture with their respective lateral thoracic radiograph. Measurement lines for the regional vertebral centroid angle are superimposed on the radiographs. The angle is measured at the intersection between lines joining the centroids of vertebral bodies T4–5 and T8–9.

($P>.05$). Physical characteristics between the high and low kyphosis groups were explored with independent t tests. Compared with the low kyphosis group, the high kyphosis group had a kyphosis angle that was 12.6 degrees greater ($P<.01$), it was 4 cm shorter in height ($P<.01$), and it had nonsignificant trends for being older in age and lighter in weight. The physical characteristics for each group are presented in Table 1.

All participants provided written, informed consent. The biomechanical model used in this study has been described previously[25,26] and is discussed briefly hereafter. A previous study by our group presented some data from the same cohort to examine the effect of vertebral fracture on spinal loads.[25]

Biomechanical Model
The steps involved in estimating spinal loads and muscle forces using the biomechanical model are described below and summarized in Figure 2.

Anthropometric data. In addition to measuring thoracic kyphosis, spinal radiographs were used to derive data on vertebral morphology for each participant that would be input into the model. Lateral radiographs of the thoracic and lumbar spine were captured at a fixed film-to-focus distance of 100 cm while participants adopted a relaxed, self-defined standing posture. At the time the radiographs were taken, a digital image of the participant also was captured at a distance of 4 m. Photographic-reflective markers were attached to anatomic landmarks (Fig. 1) on the upper limbs (head of humerus, lateral humeral epicondyle, ulnar styloid, head of the fifth metacarpal bone), neck (C7 spinous

Table 1.
Descriptive Statistics of Sample Characteristics Expressed as the Mean (SD)

Group	Age (y)	Height (cm)[a]	Mass (kg)	Kyphosis (°)[a]
High kyphosis (n=21)	63.3 (8.4)	158.5 (4.5)	63.7 (11.3)	37.6 (4.6)
Low kyphosis (n=23)	61.0 (5.6)	162.8 (5.3)	66.3 (10.2)	25.0 (5.1)

[a] Significant difference ($P<.01$, 2-tailed).
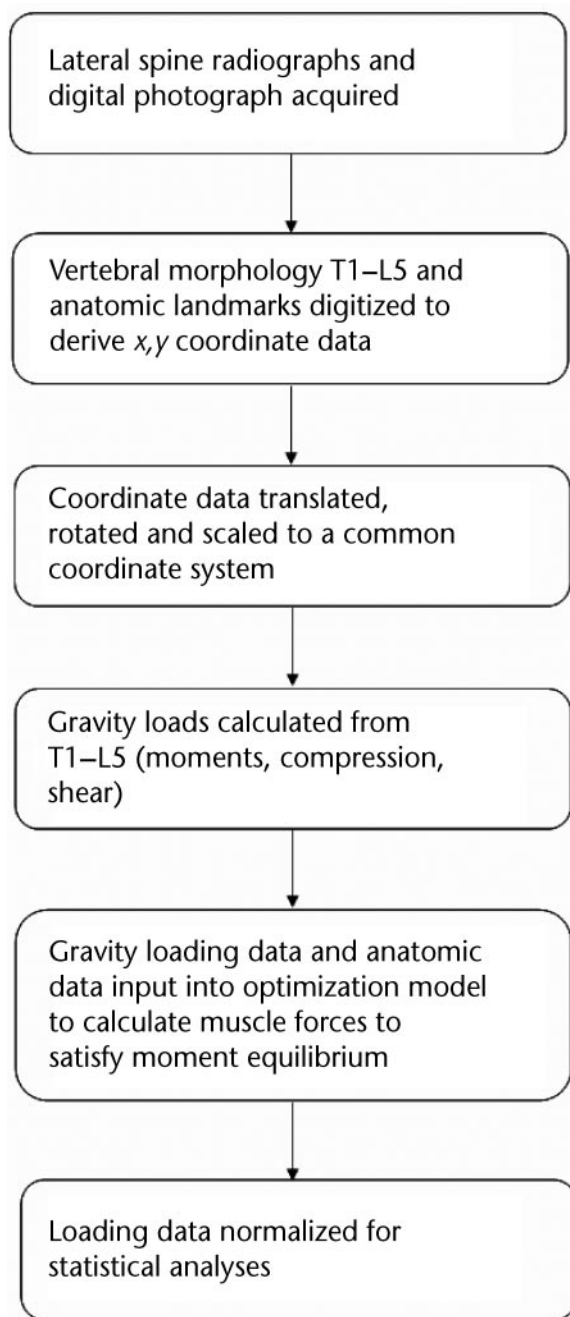
Figure 2.
Sequential steps in estimating net segmental loads and muscle forces for each participant.

process), and head (tragus) to define lengths and positions of these segments.[27,28]

Image analysis software (Image J, version 1.3*) was used to digitize Cartesian x,y coordinate data from radiographs and photographs. Coordinates of the 4 vertebral body corners of T1–L5 visible from the lateral radiograph were digitized and were used to calculate the coordinates of the vertebral centroids for T1–L5. Coordinates of the anatomic landmarks also were digitized and were used to calculate segment positions and lengths for each participant, providing a unique data set for each participant.

The thoracic and lumbar images also included a radiograph of a vertically hanging, radio-opaque ruler in a fixed position for scaling and for transforming image coordinate data to a common system. An image of this ruler also was captured in the digital photographs. Coordinate data from all images were then transformed to a common, floor-fixed coordinate system; that is, the floor was defined as the origin (0,0) of the coordinate system.

Anatomic data. Muscular anatomy was modeled with 11 bilateral trunk muscles (thoracic multifidus, lumbar multifidus, longissimus pars lumborum, iliocostalis pars lumborum, longissimus pars thoracis, iliocostalis pars thoracis, psoas, quadratus lumborum, external oblique, internal oblique, and rectus abdominis muscles), consisting of 180 muscle elements crossing T12–S1 and the longissimus and iliocostalis muscles extending to T1. These data were based on a previously published comprehensive anatomic model of the trunk in which comprehensive muscle geometry data were

reported (Fig. 3).[29] The latissimus dorsi, transversus abdominis, cervicothoracic, and shoulder complex muscles were not included.

The anatomic model also included data for the bony anatomy of the spine and thorax to which the trunk muscles were attached. Muscle geometry data were referred to the vertebrae, allowing it to be geometrically transformed to fit a specified spine shape (assuming a rigid attachment between the muscle and bony anatomy). The anatomic model, therefore, could be customized to suit the individual spinal geometries observed in the current study, based on data derived from participant-specific radiographs.

Estimation of vertebral forces due to gravity. Loads due to gravity were calculated about the vertebral centroids for T1–L5. Load parameters included segmental flexion moments, compressive forces and shear forces. Data on center of mass (COM) positions and percentage of total body mass for body segments and vertebral levels were extracted from previously published studies of elderly men and women.[27,28,30] Moment arm lengths from the vertebral centroids to the COM location at each level were scaled to participant body height, which was consistent with previous approaches.[17,18,31] From these data, gravitational flexion moments about each centroid could be calculated for each participant individually. The gravitational force at a given level included the weight force from superior vertebral levels and the head, neck, and arms. The gravitational force at each level was decomposed into compression and shear vectors based on the angle of the superior endplate tilt at each level, as determined from the radiographs. Segmental spinal loads due to gravity were input into a model used to estimate muscle forces on a
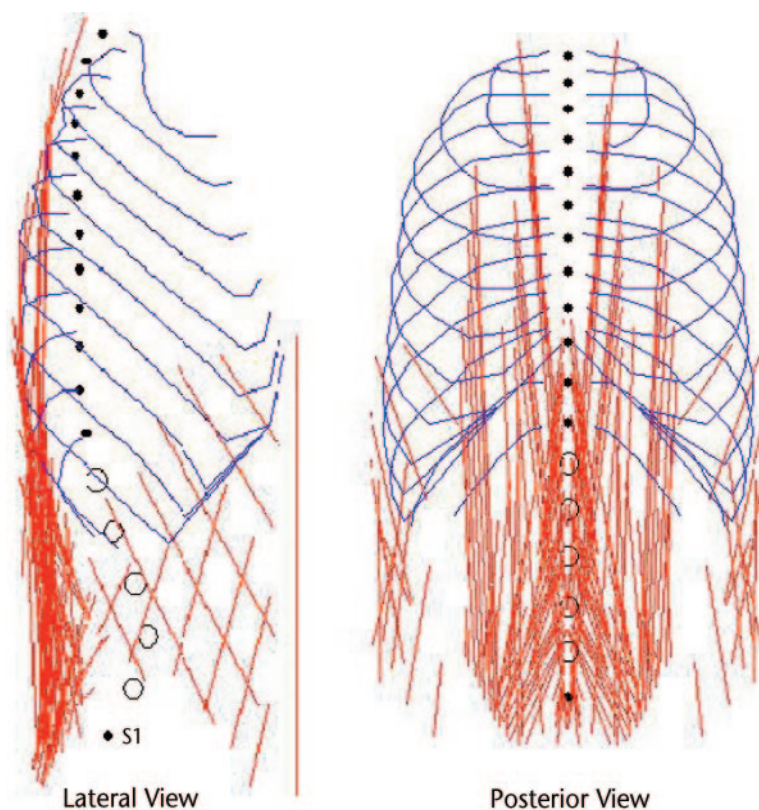
Figure 3.
Trunk muscular anatomy used in the model illustrating the positions and orientations of 180 muscle fascicles that cross T12–S1 and erector spinae musculature that cross above T12. Figure constructed using data from Stokes and Gardner-Morse.[29]

per-participant basis to satisfy the assumption of moment equilibrium.

Estimation of muscle forces: optimization routine and its constraints. Optimization is a distribution class of biomechanical modeling in which calculations of loads in individual muscles and supporting structures are performed. The complexity of musculature and passive tissue organization in the trunk creates a situation where an infinite number of possible force-producing options are available to balance external loads imposed on the system (in this case, segmental loading due to gravity). Mathematical optimization solves this situation of indeterminacy, and it provides a unique set of muscle forces from a feasible set within certain con-

straints and according to a specified criterion (cost function) aimed at maximizing physiologic efficiency. In this way, optimization models attempt to mimic muscle recruitment patterns using a similar criterion to that which the central nervous system is believed to select.[32]

Continuous, single-objective, constrained, nonlinear mathematical optimization was used to calculate trunk muscle forces from T2–L5. The model operated using a cost function aimed at minimizing muscle fatigue[33] and has been validated previously for the trunk with electromyography (EMG).[26,34,35] We consider this cost function to be appropriate given the postural role of trunk musculature. Indeed, a recent histological study[36] showed a predominance of type I

Table 2.
Details of Polynomial Functions for Each Load Parameter in Each Group and Results of t Tests Between Coefficient Terms

Polynomial Parameter	Flexion Moment			Compression Force			Shear Force		
	High Kyphosis Group	Low Kyphosis Group	P	High Kyphosis Group	Low Kyphosis Group	P	High Kyphosis Group	Low Kyphosis Group	P
x^3	9.54×10^{-6}	2.24×10^{-6}	<.0001				0.0007	0.0006	<.0001
x^2	−0.0004	−0.0002	<.0001	0.0012	0.0009	.0054	−0.0162	−0.0132	.0003
x	0.0044	0.0028	<.0001	0.0100	0.0102	.4974	0.0847	0.0704	.0282
Constant	0.0027	0.0043	.0212	0.1473	0.1599	.1075	−0.0071	−0.0118	.3974
df^{a}	353	387		333	365		332	364	
R^2	0.55	0.60		0.83	0.89		0.80	0.75	

^{a} Degrees of freedom (residuals).

muscle fibers in the thoracic and lumbar erector spinae group.

The model was constrained in terms of muscle stress ($0 \leq \sigma_{musc} \leq 50$ N/cm^2), intervertebral displacement and rotation (<1 mm, <1°) and moment equilibrium for T12–L5. Passive stiffness attributable to motion segments was modeled according to a previous study to stimulate motion segment stiffness.[37] Optimization calculations were performed using the constrained optimization function ("fmincon") in MATLAB 6.5[†] to determine a vector of muscle activation that minimized the cost function described above. Muscle activation was modeled to balance external loads (postural loading due to gravity), and muscle force estimates were decomposed into compression and shear vectors on a per-participant basis.

Data Analysis
Moments were normalized to body weight (BW) × height (Ht), and compression and shear forces were normalized to BW for each participant.

Net loading profiles. Profiles of normalized segmental load parameters from T2–L5 in each group were described with least squares polyno-

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[†] The Mathworks Inc, 3 Apple Hill Dr, Natick, MA 01760-2098.

mial regression functions. For both groups, cubic functions were fitted to segmental flexion moments and shear forces, whereas quadratic regression models described compression forces. The polynomial functions described a significant proportion of variance in load parameters for both groups (P<.0001).

The coefficient terms of the polynomial functions are detailed in Table 2. To compare differences in loading profiles between groups for each load parameter, corresponding coefficient terms in the polynomial functions were compared using independent t tests. For the polynomial functions to be considered statistically different, a significant difference between one or more corresponding coefficient terms was required. Regression functions were plotted to interpret the nature of the difference between groups. This rationale has been used previously and is an accepted statistical approach for hypothesis testing.[25,38]

Net force and muscle force in spine sections. Normalized net force and muscle force were compared between groups within spinal sections using mixed models analyses. The spine was divided into 4 anatomically functional sections (up-

per thoracic: T2–5; middle thoracic: T6–9; lower thoracic: T10–L1; lumbar: L2–5), and the mean net force and muscle force within each section were compared between groups. If muscles crossed more than one spine section, their contribution to net muscle force was included in all relevant spine sections. Compression and shear net and muscle forces were treated as dependent variables; "group" was treated as a fixed factor and "spine section" as a random factor. Interaction between the factors also was tested, and when significant, group differences in each section were explored with independent t tests.

Relationship between force and curvature. The strength of association between thoracic curvature (centroid angle) and net segmental load using pooled data (n=44) was explored with a canonical correlation. Correlations were performed separately for flexion moments, compression forces, and shear forces. The canonical correlation is a class of correlation that expresses the correspondence between sets of variables, rather than individual variables. Thus segmental loading at multiple levels (ie, a set of variables) can be correlated to another set of variables or a single variable (eg, curvature).

Data were analyzed with SPSS 12.0[‡] with the level of significance (α) set at .05. Given the directional nature of our hypotheses, *P* values are reported as 1-tailed.
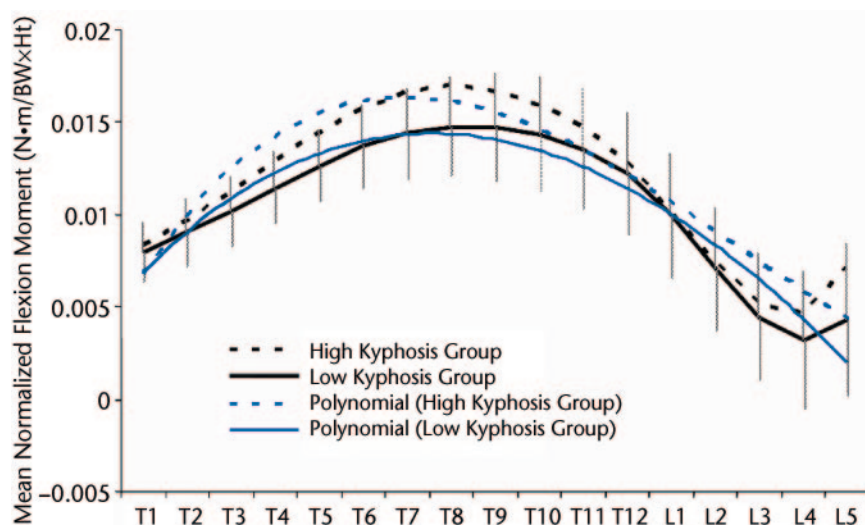
Results
Net Loading
The high kyphosis group demonstrated systematically greater flexion moments compared with the low kyphosis group, with moments peaking in the mid-thoracic spine ($P<.0001$; Tab. 2, Fig. 4). The peak mean flexion moment in both groups occurred at T8 with normalized values of 0.017 N·m/BW×Ht in the high kyphosis group and 0.015 N·m/BW×Ht in the low kyphosis group. The percentage difference in mean flexion moments between the groups from T1–L5 ranged from 1.1% to 65.6%.

A significant difference between net compression force profiles was established ($P=.005$; Tab. 2, Fig. 5). Normalized compression forces increased as a function of vertebral level in the high kyphosis (0.17–0.65 N/BW) and low kyphosis (0.18–0.57 N/BW) groups from T2–L4. The high kyphosis group had 2% to 14.4% greater mean compression from T7–L5 and 0.3% to 6.4% lower mean compression from T2–T6 compared with the low kyphosis group.

A significant difference between shear force profiles was established ($P<.0001$; Tab. 2, Fig. 6). Generally, the mean anterior and posterior shear forces of the high kyphosis group were greater than those of the low kyphosis group by 8.3% to 119.3%. Mean shear forces in the high kyphosis group were lower than those of the low kyphosis group only at L2 and L3 (17.9% and 96.8%, respectively). Within the thoracic spine, normalized anterior shear force was at a maximum at T3 in the

high kyphosis group (0.12 N/BW) and at T2 in the low kyphosis group (0.096 N/BW). Posterior shear force was at a maximum at T12 in the high kyphosis group (0.15 N/BW) and at L1 in the low kyphosis group (0.12 N/BW).

Net and Muscle Force in Spinal Sections
There was no significant difference in net force between groups for compression or shear ($P>.05$). However, a significant group × spine section interaction was established for
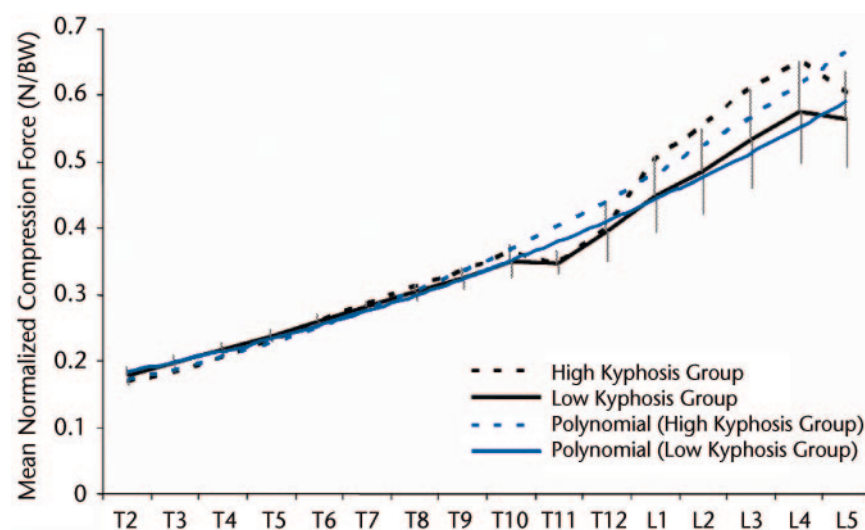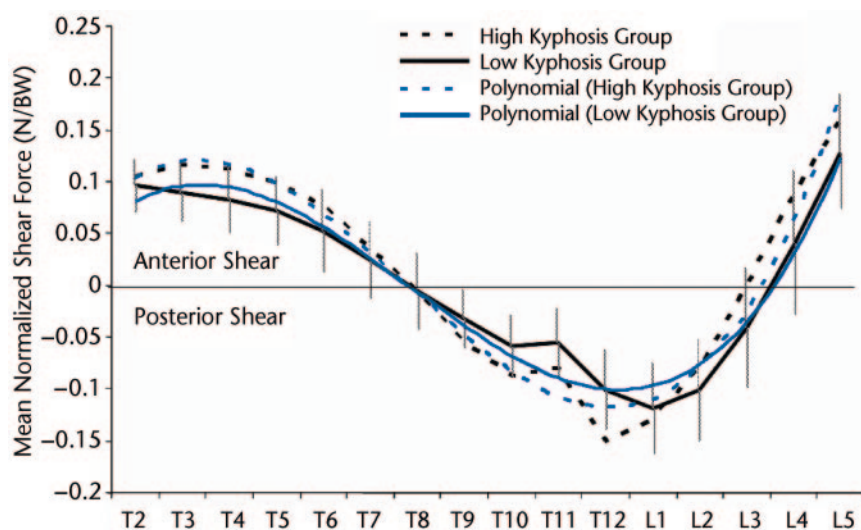

Figure 4.
Mean normalized gravitational flexion moments per segment for each group. Vertical bars represent 1 SD for the low kyphosis group. Cubic regression functions are superimposed in blue. Refer to Table 2 for details of the polynomial equation. BW=body weight, Ht=height.

Figure 5.
Mean normalized net compression force per segment for each group. Vertical bars represent 1 SD for the low kyphosis group. Quadratic regression functions are superimposed in blue. Refer to Table 2 for details of the polynomial equation. BW=body weight.

Figure 6.
Mean normalized net shear force per segment for each group. Normalized force >0 N/BW represents anterior shear and <0 N/BW posterior shear. Vertical bars represent 1 SD for the low kyphosis group. Cubic regression functions are superimposed in blue. Refer to Table 2 for details of the polynomial equation. BW=body weight.

both compression and shear muscle forces ($P<.0001$ and $P<.001$, respectively), representing differences between groups in the upper and lower spine sections (Figs. 7A and 7B). *Post hoc* testing revealed that the high kyphosis group had lower net compression force in the upper thoracic spine section and higher net compression force in the lumbar spine section ($P<.05$; Fig. 7A) compared with the low kyphosis group. There was a trend for the high kyphosis group to have higher net compression force in the mid- and lower thoracic spine sections compared with the low kyphosis group. Net shear force was greater in the high kyphosis group for the upper thoracic, lower thoracic, and lumbar spine sections ($P<.05$; Fig. 7B).

There was a significant difference in muscle force between groups for compression ($P=.02$), but not for shear ($P=.13$). A significant group × spine section interaction was established for both compression and shear muscle forces ($P<.0001$ and $P=.026$, respectively). These interac-

tions represented increasing differences in muscle force between the groups in the lower thoracic and lumbar spine (Figs. 8A and 8B). *Post hoc* testing revealed that the high kyphosis group had higher muscle compression force in all spinal sections ($P<.05$; Fig. 8A) compared with the low kyphosis group. Muscular shear force also was greater in the high kyphosis group for the upper thoracic, mid-thoracic, and lumbar sections ($P<.05$; Fig. 8B), with a trend toward greater shear force in lower thoracic section.

Force and Curvature
Strong, positive associations were found between normalized net segmental load parameters from T2–L5 and thoracic curvature ($P<.0001$). The canonical correlation coefficients (r) for each load parameter with curvature were .93, .89, and .85 for flexion moments, compression forces, and shear forces respectively. A moderately strong, positive association ($P<.01$) was observed between curvature and normalized muscle

force for compression ($r=.70$) and shear ($r=.82$) from T2–L5.

Discussion
To our knowledge, this is the first *in vivo* study to establish that naturally occurring thoracic kyphosis significantly affects spine loading profiles and the force required by the trunk muscles to maintain erect stance. The relationship between load and kyphosis was found to be strongly linear. Normalized net loading was greater in the high kyphosis group compared with the low kyphosis group for segmental flexion moments, compression force, and shear force. The normalization approach accounted for differences in body height and mass. Therefore, the loading observations are the direct consequence of anterior translation of trunk mass associated with increasing thoracic curvature. The anterior translation of mass increases the moment arm distance from the vertebral centroid to the composite COM. This results in an increased flexion moment that is counterbalanced by higher muscle forces, which, added to gravity, can be trigonometrically decomposed into greater shear and compression force vectors.

Qualitatively, the nature of the loading profiles was similar between groups. Greater loads borne by the high kyphosis group are in agreement with previous studies of lumbar spine mechanics, where greater lumbar flexion is associated with higher spinal loads.[39,40] The fact that flexion moments peaked at T8 was not surprising, considering that T8 is likely to be the apex of curvature of the thoracic spine. Compression was the dominant force vector in terms of net and muscle-derived load magnitude (Figs. 5, 7, and 8). This can be attributed to the principally axial orientation of muscle lines of action that run parallel to the spinal column (Fig. 3).

The significantly greater net and muscle shear loading in the high kyphosis group can be explained by the decrease in the verticality of muscle lines of action with increasing kyphosis. The difference in mean net force between groups was in the range of 14% for compression and 119% for shear, and correlation statistics suggest a strong, linear association between load and curvature, which is in agreement with an earlier study.[41] Therefore, greater differences between groups would be expected if kyphosis values were larger. Kyphosis in the current study ranged from 12 degrees to 51 degrees as measured by the vertebral centroid method illustrated in Figure 1.

The optimization model was constrained to satisfy moment equilibrium in the sagittal plane from T12–L5. Therefore, higher external loads (gravitational loads) in the high kyphosis group resulted in significantly greater muscle-derived extensor moments to maintain equilibrium. The muscle-derived moments were decomposed to compression and shear forces, thus explaining the higher muscle forces and net forces per vertebral level in the high kyphosis group. These findings are supported by previous studies, in which progressive increases in thoracic kyphosis were associated with greater motion segment loading and paraspinal muscle force.[18,41]

Vertebral loading due to muscle force was greater in the high kyphosis group in all spine sections (although not statistically significant for shear forces in the lower thoracic spine). Muscle-derived shear forces were low in the upper and mid-thoracic spine, highlighting that compressive force is the predominant mode of muscle loading in this area of the spine, which is attributable to the axial orientation of the paraspinal muscles. Muscular loading was less than gravitational loading in terms of
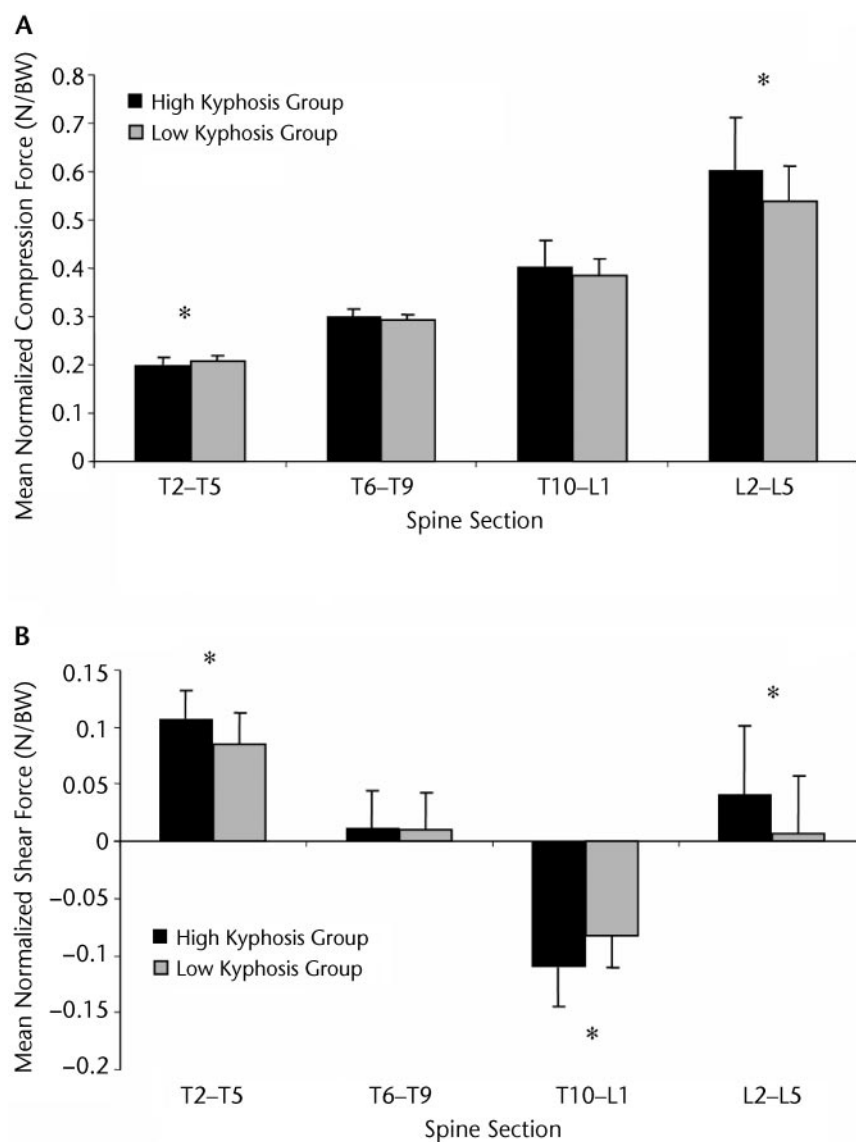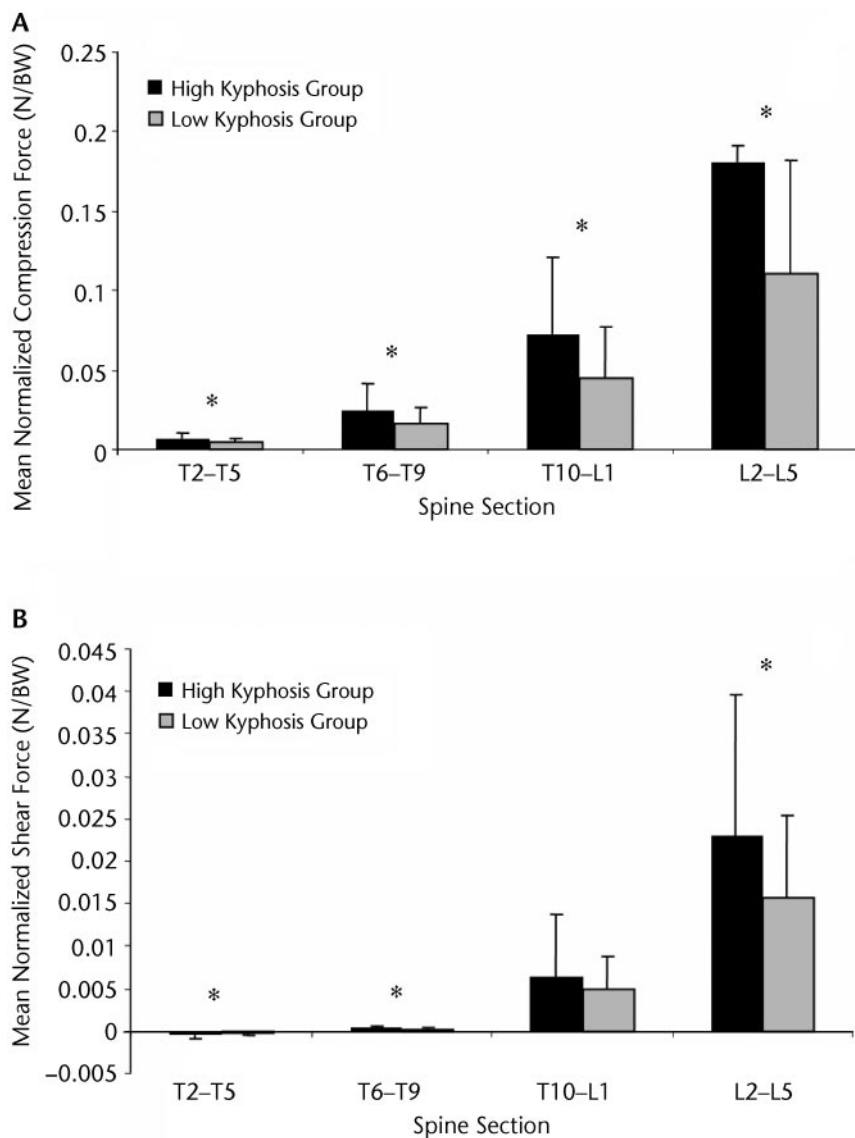

Figure 7.
Mean, normalized net compression force (A) and net shear force (B) in each group for each spinal section. The asterisk (*) denotes a significant difference (*P*<.05, 1-tailed). BW=body weight.

magnitude. However, considering the short lever arm of paraspinal muscles, large forces would be expected in these structures during functional activities (eg, lifting or manipulating objects anterior to the body) or if the magnitude of thoracic kyphosis increased.

This hypothesis is supported by the correlation results between load and curvature and a recent study where

moderately increased cervico-thoracic flexion caused an increase in the myoelectric activity (EMG) of the paraspinal musculature.[42] Although this likely translates to greater muscle force, it is uncertain given the confounding effect of muscle length and contraction velocity on the relationship between EMG amplitude and force output.

Figure 8.
Mean, normalized muscle compression force (A) and muscle shear force (B) in each group for each spinal section. The asterisk (*) denotes a significant difference ($P<.05$, 1-tailed). BW=body weight.

A range of functional and degenerative changes may occur as spinal loading resulting from kyphosis progresses. The increased vertebral and motion segment loading associated with higher kyphosis is likely to contribute to the development or progression of spinal musculoskeletal impairments and ultimately pain in cervical, thoracic, and lumbar levels.[43] A recent study[44] determined that mechanical loading of the spine in a progressively flexed posture from neutral caused significantly earlier fatigue failure of vertebral motion segments. This finding may suggest that, in addition to spinal tissue degeneration, individuals with higher kyphosis are likely to approach motion segment fatigue failure earlier. Increased spinal loading has been associated with degeneration and fatigue of the intervertebral disk, contributing to disruptions in normal cellular metabolism within the annulus and nucleus and to the development of osteoarthritis.[45,46]

Intervertebral disk degeneration also has been associated with changes in load transmission through the vertebral body, potentially increasing the risk of vertebral failure in individuals with compromised bone strength.[47] Alterations in axial load transmission as a consequence of this degeneration have been associated with architectural changes in vertebral trabecular bone.[48] It is likely that degeneration of thoracic motion segments will influence the range of motion, nature of movement, and patterns of coupled movements in the thoracic spine.[49]

In addition to the mechanical loading implication of kyphosis, sustained curvature increases the likelihood of soft tissue creep,[50] zygapophyseal joint capsule strain,[16] and ossification of spinal ligaments.[51] Functional implications of sustained thoracic kyphosis include limitations in rib cage expansion,[52] compromised balance, and, therefore, an increased falls risk in older populations,[53] and advancement of back extensor weakness.[9,54] Muscle weakness, in turn, can lead to earlier onset of fatigue, allowing the thoracic curvature to increase further and thereby exacerbate the impairments mentioned above. These consequences of elevated and sustained tissue loading, secondary to increased thoracic kyphosis, highlight a biomechanical rationale for treatment modalities aimed at minimizing thoracic kyphosis.

Treatments for kyphosis may include manual therapy,[6,55] exercise therapy,[56-60] postural re-education,[56] taping and orthoses,[56,59] and, where indicated, balance retraining.[57,59] A study examining the kinematic effects of manual therapy techniques on thoracic motion segments demonstrated that manual posteroanterior (PA) force application, in a

mode equivalent to a Maitland-like mobilization, caused extension of thoracic motion segments.[61] Ultimately, manual therapy may help reduce kyphosis by restoring or increasing motion segment mobility and by the stretching of anterior vertebral soft tissues, allowing greater range of motion for active and passive thoracic extension.

Exercise therapy aimed at increasing strength and endurance of back extensors and postural muscles would complement postural re-education in a population with increased kyphosis. External support such as taping and orthoses may complement exercise therapy regimes. Unpublished data by our group demonstrates that therapeutic taping significantly reduces thoracic kyphosis in the short term by a mean of 8%, whereas another study has demonstrated the efficacy of an orthotic brace in reducing kyphosis by 11% over 6 months.[62]

The study has several strengths. First, a clinical population was recruited to ensure heterogeneity in naturally occurring kyphosis. Our results boast greater generalizability to an elderly population, where naturally occurring kyphosis is more common, although not necessarily related to osteoporosis.[43,63] Previous studies[17,18] have induced voluntary shifts in trunk mass and modeled grades of vertebral deformity. These studies used young participants who were healthy and relied on a much simpler anatomic model. Second, the analytic approach examined net load profiles between the groups, rather than isolating force comparisons to a single motion segment. We believe analysis of a load profile to be more meaningful than an isolated load estimate, considering the high functional interdependence between spinal structures and the nature of physical therapy treatment, which incorporates more than one spinal segment.

Finally, the anatomic model used was comprehensive for the trunk, ensuring a more physiologically realistic intermuscular force distribution compared with simpler models.[35] Unlike other studies, our model considers anterior musculature and motion segment passive stiffness and does not assume a constant COM position per vertebral level.[17,18,41] Furthermore, the previously published inertial data for the trunk used in this study were derived from elderly males and females and subsequently scaled to individual participant height to maximize the physiologic accuracy of this study.[30]

Biomechanical and physiologic assumptions, however, are inherent with all models and should be considered with the results presented here. The anatomical model we used did not consider the role of transversus abdominis, latissimus dorsi, scapulothoracic, or cervicothoracic muscles; however, we did not consider these muscles to be significant moment generators in the sagittal plane.[64] The effect of muscle length was ignored in muscle force generation because we did not expect this parameter to significantly influence muscle force in upright stance. This is an accepted assumption in trunk muscle modeling for isometric muscle force estimates in erect stance, even in cases of spinal deformity.[65]

The contribution of spinal ligaments in resisting flexion moments was not considered in isolation, since this was considered negligible relative to the contribution of muscle.[66] However, passive stiffness of the motion segments was modeled on a previous study.[37] The cost function used in this study does not consider abdominal co-contraction, because the aim of the cost function is to maximize physiologic energy efficiency. However, a recent study[35] demonstrated a limited effect of co-contraction when calculating muscle

forces using this cost function compared with an EMG-driven model where co-contraction was measured. Furthermore, the effect would be systematic and not affect the primary results of the study.[34] Certainly, optimization routines using different cost functions may yield force estimates different from those reported here. However, the cost function we used has been shown to have excellent correspondence with an EMG-driven model.[26] Optimization routines do not allow comparison of neuromuscular control between individuals because a generic muscle recruitment strategy is utilized by the model based on anatomic data and the cost function employed. Future studies should use EMG-driven models to explore the associations between kyphosis and neuromuscular control. Finally, the data presented relate only to upright stance. Future studies should examine differences between groups during functional tasks and functional postures.

Conclusion

Increased curvature in the thoracic spine is associated with higher spinal loads attributable to gravity and muscle force, and a strong linear relationship exists between the magnitude of load and thoracic kyphosis. Several musculoskeletal impairments may arise as a consequence of kyphosis-induced loading and, therefore, physical therapy interventions directed to decrease kyphosis or minimize its progression are worth further investigation.

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