Eur Spine J (1997) 6:45–53

ORIGINAL ARTICLE

A. Kiefer
A. Shirazi-Adl
M. Parnianpour

Stability of the human spine in neutral postures

Received: 6 December 1995
Revised: 16 April 1996
Accepted: 24 April 1996

A. Kiefer
Division of Applied Mechanics, Department of Mechanical Engineering, Ecole Polytechnique, Montreal, Quebec, Canada, H3C 3A7
Fax 514-340 4176;
e-mail: kiefer@grbb.polymtl.ca

M. Parnianpour
Department of Industrial, Welding and Systems Engineering, The Ohio State University, Columbus, OH 43210, USA

A. Shirazi-Adl (✉)
Division of Applied Mechanics, Department of Mechanical Engineering, Ecole Polytechnique, P.O. Box 6079, Station Centre-Ville, Montreal, PQ, Canada, H3C 3A7

Abstract The present study aimed to identify some of the mechanisms affecting spinal compressive load-bearing capacity in neutral postures. Two spinal geometries were employed in the evaluation of the stabilizing mechanisms of the spine in standing neutral postures. Large-displacement finite-element models were used for parametric studies of the effect of load distribution, initial geometry, and pelvic rotation on the compression stability of the spine. The role of muscles in stabilization of the spine was also investigated using a unique muscle model based on kinematic conditions. The model with a realistic load configuration supported the largest compression load. The compressive load-bearing capacity of the passive thoracolumbar spine was found to be significantly enhanced by pelvic rotation and minimal muscular forces. Pelvic rotation and muscle forces were sensitive to the initial positioning of T1 and the spinal curvatures. To sustain the physiological gravity load, the lordotic angle increased as observed in standing postures. These predictions are in good agreement with in vitro and in vivo observations. The load-bearing potential of the ligamentous spine in compression is substantially increased by controlling its deformation modes through minimal exertion of selected muscles and rotation of the pelvis.

Key words Thoracolumbar spine · Pelvic rotation · Muscle force · Compression load-bearing capacity · Finite element

Introduction

The nature of the mechanisms by which physiological gravity loads are resisted by the active and passive components of the human trunk is an important and yet unresolved issue in the biomechanics of the spinal column. The standing or sitting neutral posture is a body position sustained in the workplace and throughout daily activities for prolonged periods of time. The stability of the spine in a neutral posture has been the subject of numerous studies. Specimens of the normal cadaveric thoracolumbar ligamentous spine (TLS), extending from T1 to S1 and devoid of musculature, have been found to carry up to 19.5 N vertical load in the lateral plane at T1 with T1 free to move, and up to 170 N with T1 prevented from moving horizontally [14]. Recent in vivo loading experiments on volunteers with asymptomatic spines found relatively low levels of superficial muscle EMG activity while an erect posture was maintained, with or without weights of up to 223 N carried in each hand [18]. The low levels of muscular activity and low stiffness of the TLS are indicative of accessory mechanisms employed to enhance its compression load-bearing capactiy (i.e., stability).

Muscles acting on the TLS can be divided into two systems: global muscles that act on the TLS through the rib cage, which control the overall response (e.g., erector spinae, abdominal muscles, etc.) and local muscles that are attached directly to the lumbar spine, which control the response of the lumbar spine (e.g., intertransverse, multifidi, etc.). Using a statically determinate model of the TLS and incorporating the influence of selected muscles in sagittal and lateral directions, conditions for stability under sagittally symmetric loading in terms of minimal muscular activity in global and local systems have been formulated [4]. An optimization method used on a continuous beam model with selected muscles pointed to the stabilizing potential of lateral moments in the frontal plane in the lumbar region when carrying 300 N load in one hand [10]. An alternative concept of the spine, as an arch in which muscles control the spinal configuration in addition to their load-resting function, has been proposed [3].

Experimental studies on the positioning accuracy of the head in a free standing posture indicate that the T1 vertebra can be centered horizontally to within 10 mm [17]. Studies on asymptomatic subjects [11, 25] and low-back patients [11] have suggested a possible correlation between sacral slope (angle between the plumb line and the line parallel to the back of proximal sacrum), lordosis, kyphosis, and the sagittal position of the T1 vertebra in both groups. The decrease in lordosis is accompanied by a decrease in sacral slope, thus preserving the horizontal sagittal distance between the T1 vertebra and the sacrum [11]. Correlation of the sacral slope with the lordosis is postulated to be an important factor in an efficient standing position [7]. Postural adjustments have also been observed in astronauts during flights under microgravity conditions; these involve flattening of the lumbar spine and a decrease in the sacral slope [15]. Moreover, lumbar lordosis has been reported to increase when external loads are added to the hands while the subject is in an erect posture [8, 18].

In a neutral standing posture, the centre of mass at various vertebral levels has been found to be located up to 30 mm anterior to the corresponding vertebral centre [13]. This anteriorly off-centered position of the physiological gravity load generates, in addition to axial forces, sagittal flexion moments on the TLS. In our previous biomechanical studies, it was found that an optimal value of sagittal flexion moment applied at the L1 vertebra increases the load-bearing capacity of the lumbar spine to 400 N with minimal horizontal displacements [21]. A similar stabilizing effect has also been observed in subsequent finite-element (FE) studies on the whole TLS [22]. The presence of sagittal and lateral moments at the T1–L5 levels in conjunction with pelvic rotation (with the pelvis considered as a rigid body) significantly improves the load-bearing capacity and stability of the TLS in axial compression [22]. A neural control system has been proposed that takes advantage of mechanisms such as upper body weight configuration and pelvic rotation to increase the load-bearing capacity of the TLS with minimal muscular effort, i.e., at minimum energy cost. The previous studies [21, 22], despite recognising the importance of moments in compression load-bearing capacity of the TLS, did not precisely identify the generating source of the moments. These studies also considered only the isolated TLS, taking no account of muscles. Moreover, no specific control parameters or corresponding cost functions were considered.

The present study attempts to overcome some of the limitations and shortcomings of our previous studies [21, 22]. A new approach to computing the required muscle activation and pelvic rotation is introduced. The likely role of muscles is then evaluated by two idealized muscle models. In the proposed approach, the neural controller is postulated to maintain T1 in a steady position in the horizontal plane [11, 17]. Full exploitation of the passive load-bearing capacity of the TLS is postulated to minimize the muscular effort required in maintaining neutral postures. Therefore, the main objectives of this study are as follows:

1. To create computationally efficient FE models of the human torso incorporating the physiological load-distribution configuration, pelvic rotation, and idealized muscle models
2. To perform a parametric study on the effect of load distribution and pelvic rotation on stability of the torso
3. To develop a new, kinematic-based approach to modelling muscles, and to apply it to identify the likely role of two idealized muscle groups in the response in neutral postures
4. To determine the influence of the two different spinal geometries and the initial positioning of T1 on the stabilizing pelvic rotation and muscle forces

Materials and methods

Finite-element model

Spinal behavior was evaluated using ABAQUS FE structural analysis software. Both the Timoshenko beam element, including transverse shear deformation, and the Euler beam element with no transverse shear deformation were employed to represent the intervertebral disks. In the range of load considered in this study, no significant difference was observed in the response predictions based on either of the two elements, and the Euler beam element was used for subsequent studies. Rigid body elements were used for the vertebral bodies and lever arms for the off-centered application of loads. To quantify the instability behavior, large-deformation analysis was performed.

Two geometries of the TLS (curves passing through centroids of disks and vertebrae), SP1 and SP2 extending from T1 to S1, with a total height of 468 mm, were used in the present study (Fig. 1). The SP1 geometry was based on lumbar spine CT measurements and mean anthropometric data for normal subjects from the literature [22, 25] (Table 1). The spinal curve for SP2 was traced directly from sagittal and lateral radiographs of a subject in the standing neutral posture. Its dimensions were then rescaled to obtain identical disk and vertebral heights as those in SP1. The two geometries exhibited the following angulations between the verte-

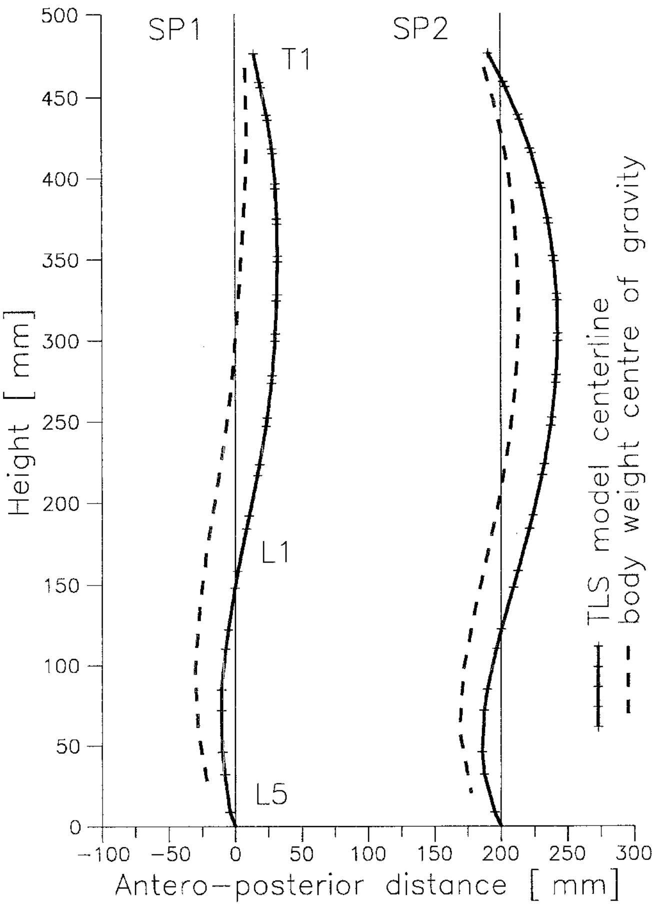

Fig. 1 Sagittal profile of models of the thoracolumbar ligamentous spine (TLS). Both models SP1 and SP2 are fixed at the distal end

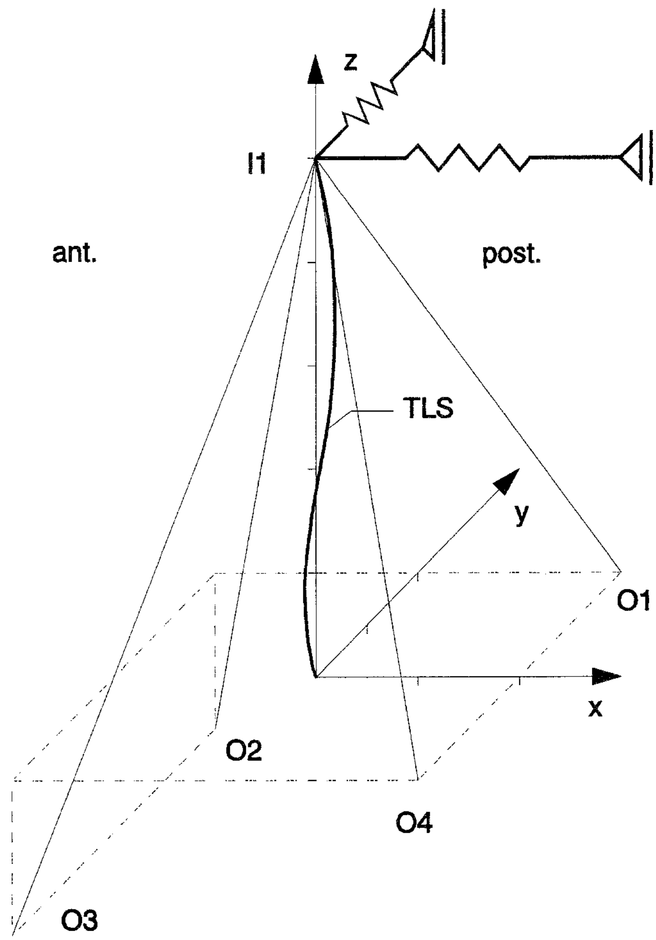

Fig. 2 Simplified muscle model. Only the insertion point ($I1$) at T1 with its points of origin at the pelvis ($O1$, $O2$, $O3$, $O4$) is shown. Springs at the point of insertion are added to evaluate muscle activations and allow vertical movement only

Table 1 Structural properties of beams simulating intervertebral disks (Ix, Iy area moments of inertia, Jz polar moment of inertia, h disk height)

Disk	Area (mm^2)	Ix, Iy (mm^4)	Jz (mm^4)	h (mm)
T1–T2	570	25855	51709	2.88
T2–T3	610	29611	59222	2.93
T3–T4	660	34664	69328	2.97
T4–T5	725	41828	83656	2.99
T5–T6	775	47796	95529	3.20
T6–T7	840	56150	112300	3.50
T7–T8	850	57495	114988	4.00
T8–T9	875	60927	121853	4.49
T9–T10	950	71819	143637	4.67
T10–T11	980	76426	152852	5.02
T11–T12	1190	112690	225379	6.63
T12–L1	1270	128351	256701	8.31
L1–L2	1310	136563	273126	10.37
L2–L3	1385	152647	305295	11.92
L3–L4	1425	161592	323184	13.00
L4–L5	1455	168467	336935	13.53
L5–S1	1555	192420	384841	8.70

bral centres: kyphotic (between T2 and T12) 29° and 48° for SP1 and SP2 respectively; lordotic (between L1 and S1) 39° and 48° for SP1 and SP2, respectively. These values are in the range of normal values as specified by mean anthropometric data [25]. The sagittal position of the centre of the T1 vertebra with respect to the centre of the S1 proximal endplate was 16.48 mm posteriorly in the SP1 curve and 3.69 mm anteriorly in the SP2 curve.

The models were created by interconnecting 17 3D beam elements for the intervertebral disks and 17 rigid bodies for the vertebrae at discrete points along the spinal curves. The structural properties were taken from the literature [21, 22] and are listed in Table 1 (Young's modulus E = 7MPa, Shear modulus G = 3MPa). The presence of the rib cage was simulated by a five-fold increase in the Young's modulus of the thoracic disks T1–T11 [2, 22]. The FE models were fixed at the S1 level. The orientation of the global coordinates system with respect to the FE models is shown in Figs. 1 and 2.

Loading conditions

Three types of load, concentrated axial load at the centre of the T1 vertebra (CF), axial load applied at the centreline of the TLS distributed between individual vertebrae (CL), and load applied at the

Table 2 Positions and magnitudes of LA load (the model most closely simulating real physiological gravity load) along the height of the spine

Vertebra	Axial load (N)	Eccentricity (mm)
T1	8.80	8.66
T2	9.52	13.35
T3	10.16	17.45
T4	10.92	20.97
T5	11.56	23.91
T6	12.32	26.26
T7	12.96	28.04
T8	13.72	29.23
T9	14.48	29.84
T10	15.12	29.87
T11	15.84	29.32
T12	16.52	28.18
L1	17.24	26.46
L2	17.88	24.16
L3	18.64	21.27
L4	19.28	17.81
L5	20.08	13.76

lever arms extending from the centre of gravity of the body slices to the corresponding vertebral centres (LA), were used in the present analysis to investigate the importance of the load modelling on the load-bearing capacity of the TLS. In these three load scenarios, CF represents an idealized concentrated load application, CL represents a trunk load distributed over individual segments but with no eccentricity, while LA is the closest simulation of the real physiological gravity load. Distribution of the trunk load over individual segments using their lever arms (the LA scenario) was based on the data reported by King-Liu et al. [13] rescaled to fit the height of the present TLS models, as listed in Table 2. The distribution of load in the CL and LA scenarios was calculated for a total body weight of 75 kg, from which 40 kg were assigned to the TLS. The head and neck load was 8 kg and was applied to the centre of T1. The load of the arms, 7.5 kg, was equally distributed between the T1–T6 vertebrae through the action of the shoulder girdle and the rib cage. The location of these loads was radiographically evaluated to be about 50 mm anterior to the centroid of the T4 vertebra. The total weight of 12 trunk slices was 24.5 kg. The anterior distance of the centroid of each slice from its corresponding vertebral centre is given in Table 2.

Pevlic rotation simulation and initial geometry alterations

An optimization method was used to evaluate the effect of pelvic rotation on the response of the TLS. With the T1 vertebra prevented from moving horizontally from its initial position [11, 17], horizontal reaction at T1 was taken as the cost function dependent on the magnitude of the pelvic rotation. The required values of optimal pelvic rotation were then evaluated at different loads by minimizing the cost function. To study the influence of the sagittal position of T1 with respect to the centre of the S1 proximal endplate on the response, additional sets of initial geometry were obtained by rigid body rotation of SP1 in the range 0.5° anterior to 0.4° posterior, corresponding to 4.48 mm anterior to 3.52 mm posterior horizontal translations at the T1 vertebra. For the SP2 model, the S1 base was rotated by 1.7° posteriorly, corresponding to a 14 mm posterior translation at T1.

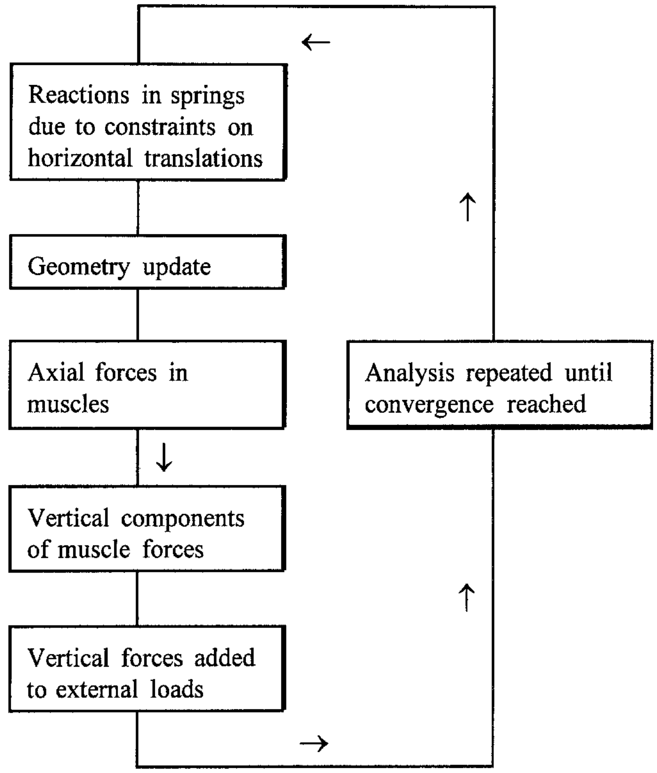

Fig. 3 Iterative procedure for muscle force calculation

Idealized muscle model

A technique for determining minimal muscle forces without antagonistic action based on static equilibrium and taking into account the elastic properties of the TLS has been developed. The complex anatomy of the individual spinal muscles is idealized by one local and one global muscle system. The division of muscles in this study into a local and a global group [4, 23] is based on the natural double-curved shape of the spine, where minimal forces acting in two critical regions along the spinal height can significantly enhance the TLS load-bearing capacity. In the present study, the T1 and L1 vertebrae were selected as the respective points of force application (i.e., muscle application) of the global and local system.

Forces in each muscle group are oriented in four possible directions. Lines of action extending from the pelvis to the TLS are chosen so as to satisfy the bodys anatomical limitations. Coordinates [x, y, z] of the four points of origin at the pelvis in millimeters are (Fig. 2): right posterior $O1$ = [72,5, 100, –20], right anterior $O2$ = [–121, 100, –102], left anterior $O3$ = [–121, –100, –102], and left posterior $O4$ = [72.5, –100, –20]. The insertion points on the TLS are the centroid of T1 for the global muscle system and the centroid of L1 for the local muscle system.

Two springs are added at each insertion point in the direction of the global horizontal x and y axes, as shown in Fig. 2. The spring endpoints are constrained to move with the insertion points so as to preserve the horizontal orientation of the springs. The reactions in the springs represent the horizontal force necessary to restrain horizontal translation at the insertion point. The two directions in which the active idealized muscles exert a force are selected from among the four possible directions according to the computed horizontal forces at the insertion point, to yield the tensile forces in the muscles. At the end of each load increment, therefore, the horizontal reactions at the insertion points can be resolved into two active directions, allowing the evaluation of muscular forces. In the subsequent iteration, the vertical component of the muscle forces is applied at each insertion point as an additional axial load on the TLS. The procedure is repeated until changes in the muscular forces between two consecutive iterations are negligible and convergence is reached (see Fig. 3 for the flow chart). This procedure exemplifies the nature of muscular contribution where balancing horizontal forces are accompanied by adverse compressive forces. The purpose of the springs attached at the insertion points is primarily to control the deformation modes of the TLS and secondarily to identify the degree of activity in the global and local muscle systems. The forces in the muscle systems thus become a function of the TLS configuration, loading, and elastic properties of the TLS. It is to be emphasized that in this study the muscle modelling was idealized to evaluate the overall role of muscles in the stabilization of the spine.

Results

The isolated TLS exhibits transition into hypermobility under axial loads considerably smaller than physiological loads. As the loading configuration approaches its physiological distribution (i.e., the LA scenario), a marked improvement in the stiffness and stability response of the TLS is observed, as depicted in the SP1 model in Fig. 4. The response to the CF and CL loadings is smooth with gradually decreasing stiffness of the TLS and without a sharp bifurcation-type transition into instability. The LA loading, however, generates considerably higher initial stiffness with an axial translation of around 1 mm under the first 75 N load, followed by a sudden transition into hypermobility. The instability occurs primarily in the sagittal plane, which is the least stiff plane.

Pelvic rotation substantially stiffens the response of the TLS in all three scenaries. For the LA load, a 2° anterior pelvic rotation allows the TLS to carry axial compressive loads of up to 400 N while undergoing only 7 mm axial displacement at T1, as shown in Fig. 4. Optimal pelvic rotation depends on the magnitude of axial compression and the initial geometry of the TLS, in particular the sagittal position of T1 with respect to S1. A posterior positioning of T1 increases the optimal pelvic rotation in the anterior direction, while a posterior pelvic rotation is needed for an anterior positioning of T1, as shown in Fig. 5. The calculated results indicate that the magnitude of the compressive load and the vertical alignment on of T1 have a marked effect on the position of the stabilizing pelvic rotation. It can be noted that within the range of vertical alignments considered, as the compressive load increases, in some cases the optimal pelvic rotation reaches zero and reverses its direction from posterior to anterior. Without any initial sagittal translation at T1, this transition occurs at a load of 200 N, whereas with 2.15 mm initial anterior sagittal translation, the transition occurs at about 400 N, which corresponds to the assumed physiological gravity load of the upper body weight. The horizontal positioning of T1 also influences the distribution of sagittal moments along the spinal height. An anterior shift of T1, for example, causes a decrease in flexion moments in the lumbar region and an increase in extension moments in the thoracic region, as shown in Fig. 6. These are accompanied by corresponding changes in spinal curvatures, i.e., decreased lordosis at the lumbar region and increased kyphosis at the thoracic region, as shown in Fig. 7.

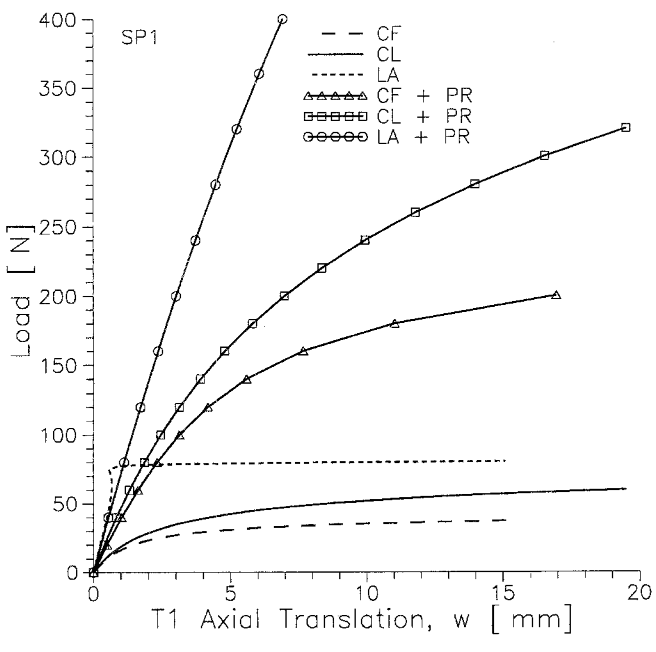

Fig. 4 Predicted response of the SP1 model, with the rib cage, under CF, CL, and LA loads (see Loading conditions) with and without pelvic rotation *(PR)*. The CL and LA loads are equal to the sum of all individual loads applied at different levels

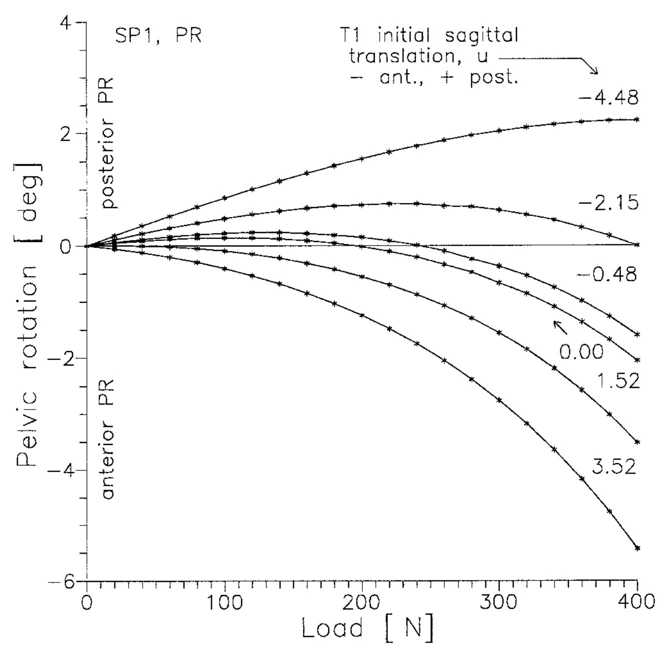

Fig. 5 Predicted variation in optimal pelvic rotation under LA load for the SP1 model as a function of the initial sagittal position of T1

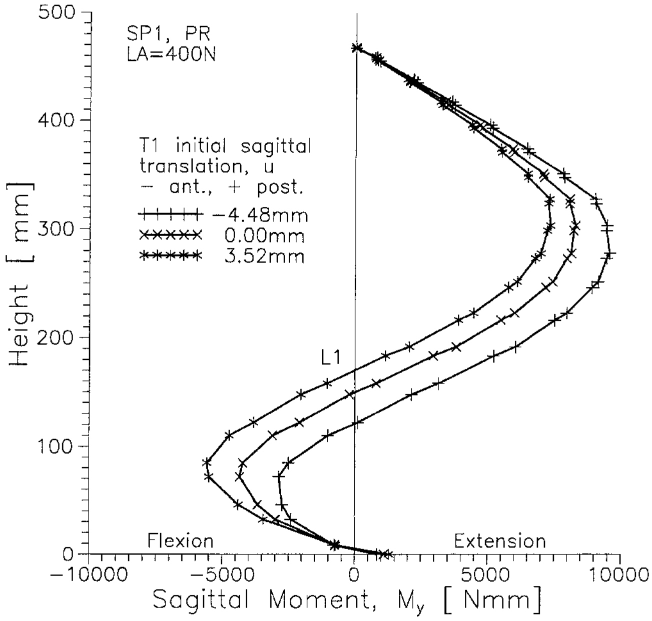

Fig. 6 Predicted variation in the sagittal moment along the height for the SP1 model stabilized by pelvic rotation under 400 N axial load as a function of initial positioning of T1

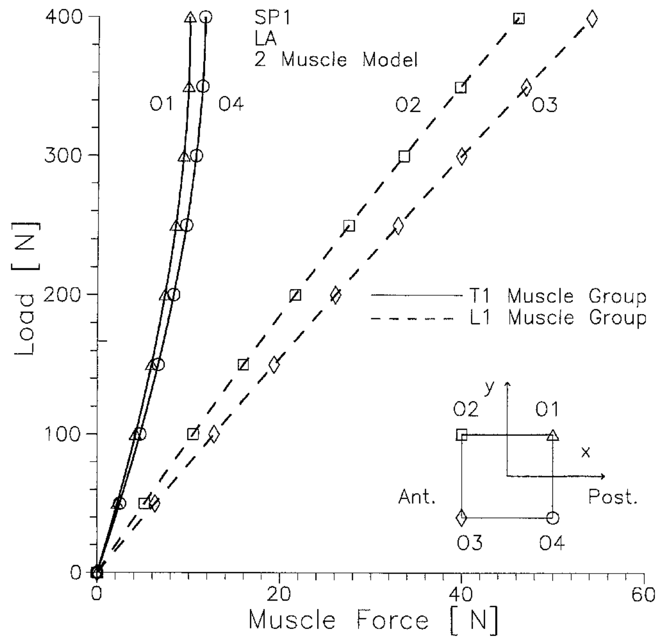

Fig. 8 Predicted variation in forces in both muscle groups for the two-muscle model with the LA load for the SP1 geometry. The pelvic points of origin for both muscle groups are given in Fig. 2

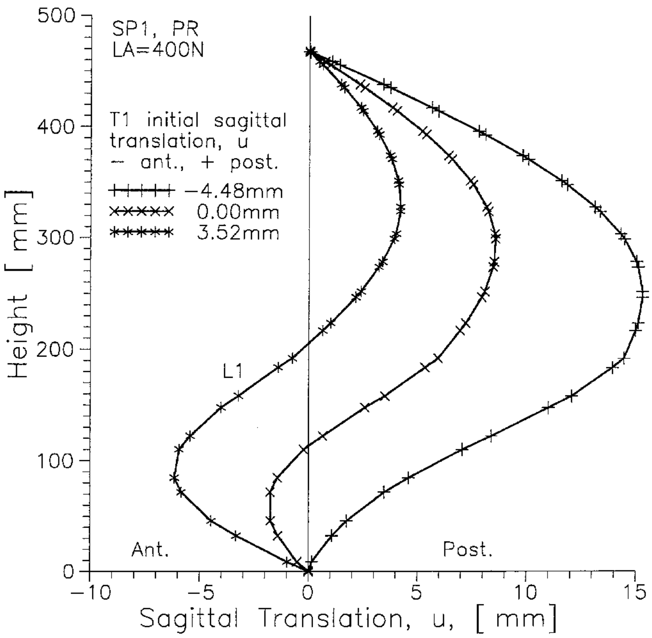

Fig. 7 Variation in sagittal displacements in the SP1 model stabilized by pelvic rotation under 400 N axial load as a function of the initial positioning of T1

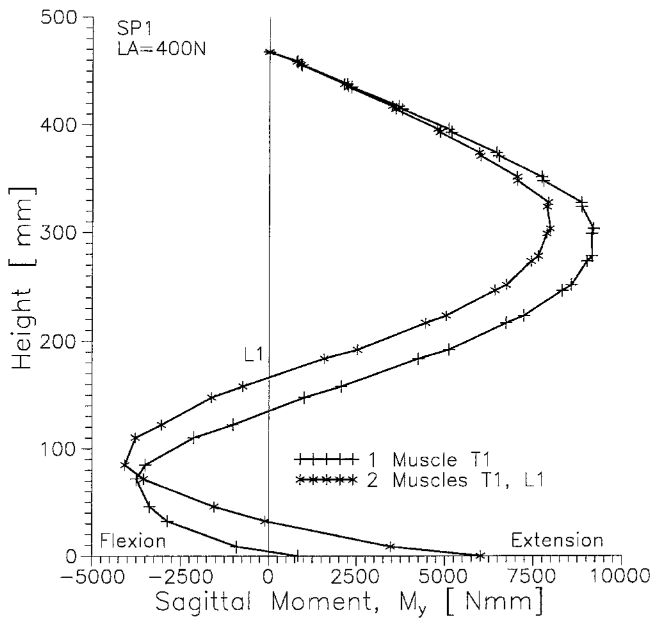

Fig. 9 Predicted variation in the sagittal moment along the height of the SP1 model under 400 N LA load for both one- and two-muscle models

Using the one-muscle model, under LA loading, the calculated muscular activity in the global system (T1 insertion) is minimal (less than 5 N each) under loads of up to 400 N. The pattern of muscular activity is asymmetric with respect to the sagittal plane due to the initial lateral deviation of spinal geometry in the models. Very small values of muscle forces (less than 1 N each) predicted at an axial load of about 200 N correspond to the transition region of nearly zero pelvic rotation without initial sagittal translation that was already identified in Fig. 5. Simultaneous activation of the global (T1 insertion) and local (L1 insertion) systems increases muscle forces, with

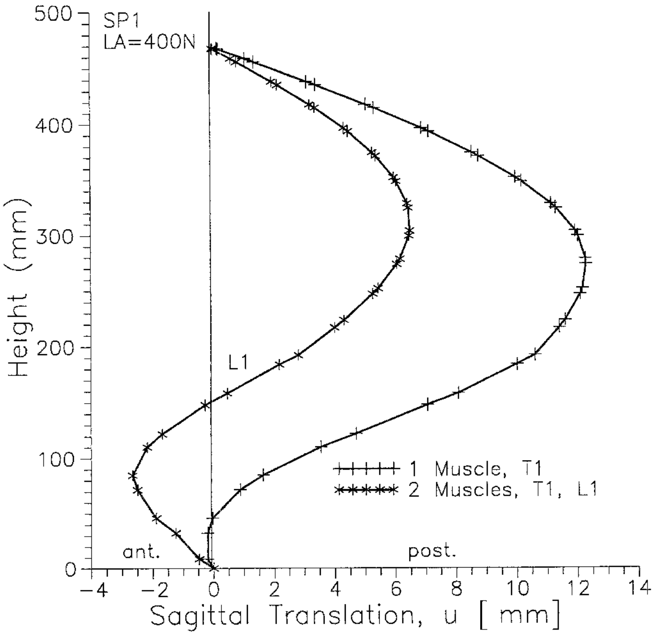

Fig. 10 Variation in sagittal translations along the height of the SP1 model under 400 N LA load for both one- and two-muscle models

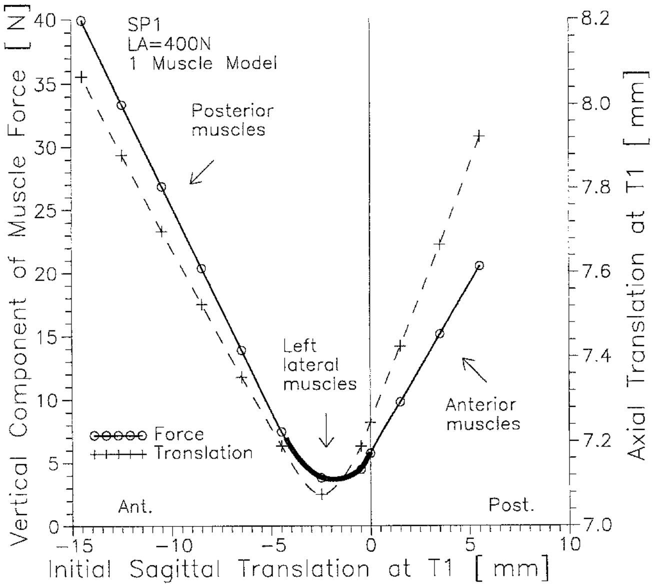

Fig. 11 Variation in T1 vertical movement and the vertical component of muscle force in the one-muscle model under 400 N LA load as a function of initial sagittal positioning of T1

forces in the local system being about five times higher than those in the global system, as shown in Fig. 8. The T1 group shows activation of the posterior muscles, while the anterior muscles are active in the L1 group. The lateral asymmetry in muscular forces is nearly 15%. The effect of muscle models on distribution of sagittal moment along the height of the TLS is most marked at the pelvic level, as shown in Fig. 9. Activation of the local muscle system considerably increases the sagittal moment at the pelvic level, thereby altering the spinal curvatures, as shown in Fig. 10. A significant reduction in sagittal translations in the thoracic region is seen to be accompanied by a small increase in anterior translations in the lumbar region. The variations in the vertical component of muscle forces in the model based on one muscle group attached at T1, and in the associated axial translations at T1 as a function of initial sagittal translation at T1 under 400 N load, are shown in Fig. 11. These are found to be in accordance with earlier predictions shown in Fig. 5. Their minima occur close to –2.2 mm initial anterior sagittal translation at T1, which is also the value for the initial sagittal translation requiring nearly no pelvic rotation for equilibrium under 400 N load, as seen in Fig. 5.

Analysis of the foregoing cases with the SP2 rather than the SP1 model yields similar trends, but with different magnitudes. The SP2 model is more flexible than the SP1 model, with or without pelvic rotation or muscles. It requires larger pelvic rotation, 7° posteriorly rather than 2° anteriorly, as shown in Fig. 5 for the SP1 under 400 N load. As for the influence of the T1 initial positioning on pelvic rotation at 400 N load, a posterior shift of T1 reduces the above pelvic rotation so that no rotation is needed at about 7 mm posterior shift of T1, while anterior pelvic rotations is required for T1 posterior shifts beyond 7 mm. For example, about 3° anterior pelvic rotation is computed for the scenario of a 10 mm posterior shift at T1. The results of both muscle models exhibit considerably larger muscle forces with more pronounced asymmetry as compared with those computed for the SP1. For example, at 400 N load, total forces of 41 N and 96 N are predicted at the T1 muscle in the one- and two-muscle models, respectively. In the latter, a total force of 407 N is calculated at the L1 muscle. Finally, a posterior shift of 7 mm in the initial position of T1 is found to substantially reduce these muscle forces, similar to the trend shown in Fig. 11 for the SP1 model. For example, in the case of an initial 7 mm posterior and 10 mm right lateral T1 shift, the above-mentioned forces diminish to 1 N, 30 N, and 160 N, respectively.

Discussion

The present study is aimed to identify some feasible mechanisms for maintaining spinal stability in neutral postures. These were: a realistic distribution of the upper body weight [13, 24, 26, 27], rotation at the pelvis [8, 11, 22], initial positioning of T1 [11, 17], and the role of some muscles [4]. The TLS models with reasonable structural properties [2, 22–24, 26, 27] used in this study were both sufficiently practical to make possible a large number of nonlinear analyses coupled with constraint equations and sufficiently accurate to determine the stability response in upright postures under relatively small axial compressive forces. The model developed for this study was motivated by observation of low activity of the trunk muscles during in vivo tests when an upright posture is maintained. The simplified muscle representation was based on previous works, suggesting the distinct effect of the local and global muscle groups on spinal behavior [4, 6]. The muscle recruitment strategy allowed maximum utilization of the elastic resistance of the passive tissues while minimizing the muscle force required to stabilize the torso within the physiological limits of displacement. The goal of this idealized muscle representation was to perform preliminary investigation of the overall function of muscles in the response of the spine, rather than to model a detailed muscle anatomy. The proven applicability of the present numerical model calls for future refinement in muscle modelling. Finally, the nonlinear large-displacement analysis in this study allowed for the accurate evaluation of muscle forces and rigorous quantification of the instability response. It is to be noted that, due to the lack of bifurcation or limit phenomena, the term "instability" was used throughout this study to indicate large displacements or hypermobility that occur due to loss of global stiffness in axial compression. Moreover, "load-bearing capacity" refers to the notion of stability in axial compression [21, 22] and not the compressive strength of the system. Finally, the equilibrium configurations computed in this study are stable, and remain so far as the pelvic rotation and/or muscle activations are allowed to react to perturbations in the system.

In both the SP1 and SP2 models, the distribution of the trunk gravity load among individual vertebrae along the spinal centerline improved the compression load-bearing capacity of the TLS when compared with the case of a single concentrated load at T1. Similar trends have already been indicated by previous studies [20-22, 24, 26, 27]. The predicted critical load under a single load at T1 is in agreement with the reported values [2, 14, 22]. The response was further improved when the loads were shifted anteriorly at the centre of the mass at each level, as measured in cadaveric specimens [13] (see Fig. 4).

Evaluation of the optimal pelvic rotation and muscle forces was perfomed using a unique kinematic-based constraint system of equations. In the present study, the horizontal translation at each vertebral level was constrained by adding artificial horizontal springs. The magnitude of pelvic rotation at the base was then evaluated by minimizing the spring forces. In the muscle model, the axial load in each muscle was calculated from the spring forces, using the static equilibrium equations followed by an iterative procedure until the convergence was reached. Due to the directions in which muscle forces acted, the stabilizing horizontal forces were accompanied by adverse vertical forces, which increased the axial load on the spine (Figs. 2, 3). The T1 horizontal movement was selected as a control parameter based on the findings of earlier studies [11] that the position of the head remains constant and that a change in lordosis accompanies a change in sacral slope, thus preserving the horizontal sagittal distance of the T1 vertebra from the sacrum. In the two-muscle group model, the L1 horizontal displacements were also assumed to be constrained to provide the required equations to evaluate the added muscle forces.

In spite of the larger resistance of the TLS when considering physiological load configuration (LA loading), the magnitude of total compression at hypermobility remained well below the upper body weight. The presence of optimal pelvic rotation was found to substantially stiffen the TLS so that the physiological loads could be carried with relatively small displacements (Fig. 4). These rotations have been observed in vivo in neutral postures and are affected by the magnitude of the compressive load and spinal geometry [9, 18]. The predicted results also point to the strong dependence of optimal pelvic rotation on spinal configuration. Pelvic rotation appears to stabilize the TLS by increasing lordosis. The initial lordotic angle of 39° in SP1 increased to 45.2° under 400 N axial load. Initial posterior placement of T1 further increased this trend, while an anterior placement decreased it. The stabilizing role of pelvic rotation was more evident in this study than in our earlier one [22]. This is because a more efficient way of simulating pelvic rotation was used in this study.

The compression load-bearing capacity of the TLS substantially increased with the addition of muscle groups at T1 alone or at T1 and L1. Relatively small axial forces were generated in muscles at 400 N load, especially in the SP1 model, for the global group with T1 insertion. As expected, the overall horizontal displacements were smaller in the two-muscle system although at the expense of increased sagittal moments in the lower lumbar region. In both muscle systems for SP1, the initial lordosis increased from 39° to about 43°. Similar to the optimal pelvic rotation, the geometry of the spine markedly affected the muscle forces required for spinal stability. This indicates that there is an optimal initial positioning of T1 that minimizes the required muscle forces (Fig. 11). The simultaneous presence of pelvic rotation and muscle activity, not considered in this study, could further enhance the spinal load-bearing capacity with smaller pelvic rotation and less muscle activity.

The results for the SP2 curve indicated similar trends to those of SP1. Quantitative differences in the predicted results occurred primarily because of the larger values of kyphotic and lordotic angulations in SP2, compared with SP1. Changes in optimal pelvic rotation can be attributed to differences in the sagittal position of the T1 vertebra with respect to the centre of the S1 proximal endplate. The positioning of T1 influences the magnitudes of muscle forces in a way similar to that shown in Fig. 11 for the SP1 curve.

Comparison of results (Figs. 7, 10) demonstrates the similarities between the stabilizing influence of pelvic rotation and that of muscles in the spine under axial compression. Both mechanisms were found to increase the initial lordotic angle of the TLS in neutral posture. Interestingly, the lumbar spine has been found to have a larger lordotic angle in standing postures than in stress-free cadaveric specimens [1]. Similar trends have been observed under microgravity conditions [15] and in sideways lying positions [12] free of axial loads, were flattening of the spine occurs, as compared with the a neutral standing position under gravity loads. This lordotic posture, although a function of load and T1 positioning, likely enhances the compression stability of the spine.

In summary, the findings of this work indicate that small muscle activations and pelvic rotation fully exploit the passive load-bearing potential of the TLS [9, 11] by controlling its deformation modes. The trunk in free standing posture under physiological gravity loads exploits the TLS double curvature and decreases its effective buckling length by the action of muscles and pelvic rotation, thereby increasing its compression load-bearing capacity while decreasing its horizontal displacements. To move into and maintain higher modes of deformation, only relatively low muscular forces are required when applied at critical regions along the height of the TLS (i.e. the inflection points). In contrast to the strategy for stabilization of upper and lower extremity joints with the presence of high coactivations, the absence of muscle coactivation in the trunk upright posture [18] indicates that a more intelligent control strategy is used for the stabilzation of the spine. The actions of muscles and pelvic rotation are postulated to be coordinated by a neural controller, with the horizontal translation at T1 being a likely feedback parameter. The results suggest that pelvic rotation, muscle activation, and the off-centre placement of the line of gravity are exploited to stabilize the passive spinal system in neutral postures. The evaluation of the stability of the spine at higher exertions and more complex loading conditions deserves a separate analysis with inclusion of more realistic muscle anatomies.

Acknowledgement This study was supported by the Natural Sciences and Engineering Research Council of Canada (NSERC).

References

1. Adams MA, Dolan P, Hutton WC (1988) The lumbar spine in backward bending. Spine 13:1019–1026
2. Andriacchi T, Schultz A, Belytschko T (1974) A model for studies of mechanical interactions between the human spine and rib cage. J Biomech 7:497–507
3. Aspden RM (1992) Review of the functional anatomy of the spinal ligaments and the lumbar erector spinae muscles. Clin Anat 5:372–387
4. Bergmark A (1989) Stability of lumbar spine. Acta Orthop Scand [Suppl 230] 60:1–54
5. Bernhardt M, Bridwell KH (1989) Segmental analysis of the sagittal plane aligment of the normal thoracic and lumbar spines and thoracolumbar junction. Spine 14:717–721
6. Cholewicky J (1993) Mechanical stability of the in vivo lumbar spine. Thesis, University of Waterloo, Waterloo, Canada
7. Duval-Beaupere G, Schmidt C, Cosson P (1992) A barycentremetric study of the sagittal shape of spine and pelvis: the conditions required for an economic standing position. Ann Biomed Eng 20:451–462
8. Gracovetsky S (1988) The spine engine. Springer, Vienna New York, pp 151–152
9. Gracovetsky S, Farfan H, Helleur C (1985) The abdominal mechanism. Spine 10:317–324
10. Hjalmars S (1988) A beam model of the human spine under muscular action. J Tech Phys 29:43–49
11. Jackson RP, McManus AC (1994) Radiographic analysis of sagittal plane alignments and balance in standing volunteers and patients with low back pain matched for age, sex and size: a prospective controlled clinical study. Spine 19:1611–1618
12. Keegan JJ (1951) Alterations of the lumbar curve related to posture and seating. J Bone Joint Surg [Am] 35: 589–603
13. King-Liu YK, Wickstrom JK (1973) Estimation of the inertia property distribution of the human torso from segmented cadaveric data. Perspect Biomed Eng, pp 203–213
14. Lucas DB, Bresler B (1961) Stability of the ligamentous spine. Biomechanics Laboratory, Berkeley, Report 40-WI-CA 4361
15. Massion J, Gurfinkel V, Lipshits M, Obadia A, Popov K (1993) Axial synergies under microgravity conditions. J Vestib Res 3:275–287
16. McGill SM (1991) Kinetic potential of the lumbar trunk musculature about three orthogonal orthopaedic axes in extreme postures. Spine 16:809–815
17. McGlashen K, Ashton-Miller JA, Green M, Schultz AB (1990) Trunk positioning accuracy in the frontal and sagittal planes. J Orthop Res 9:576–583
18. Parnianpour M, Shirazi-Adl A, Sparto P, Dariush B (1994) The effect of compressive load on myoelectric activities of ten selected trunk muscles. In: Proceedings of the 12th TCIEA, vol 3. Toronto, Canada, August 15–19, 1994, pp 119–121
19. Pearsall DJ, Reid JG (1992) Line of gravity relative to upright vertebral posture. Clin Biomech 7:80–86
20. Scholten PJM, Veldhuizen AG, Grootenboer HJ (1988) Stability of the human spine: a biomechanical study. Clin Biomech 3:27–33
21. Shirazi-Adl A, Parnianpour M (1993) Nonlinear response analysis of the human ligamentous lumbar spine in compression – on mechanisms affecting the postural stability. Spine 18:147–158
22. Shirazi-Adl A, Parnianpour M (1996) Stabilizing role of moments and pelvic rotation of human spine in compression. J Biomech Eng 118:26–31
23. Stokes IAF, Gardner-Morse M (1995) Lumbar spine maximum efforts and muscle recruitment patterns predicted by a model with multijoint muscles and joints with stiffness. J Biomech 28:173–186
24. Takashima ST, Singh SP, Haderspeck KA, Schultz AB (1979) A model for semi-quantitative studies of muscle actions. J Biomech 12:929–939
25. Voutsinas SA, MacEven GD (1984) Sagittal profiles of the spine. Clin Orthop 210:235–242
26. Yettram AL, Jackman MJ (1980) Equilibrium analysis for the forces in the human spinal column and its musculature. Spine 5:402–411
27. Yettram AL, Jackman MJ (1982) Structural analysis for the forces in the human spinal column and its musculature. J Biomed Eng 4:118–124