Novel model to analyze the effect of a large compressive follower pre-load on range of motions in a lumbar spine

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Abstract

A 3-D finite element model (FEM) of the lumbar spine (L1–S1) was used to determine the effect of a large compressive follower pre-load on range of motions (ROM) in all three planes. The follower load modeled in the FEM produced minimal vertebral rotations in all the three planes. The model was validated by comparing the disc compression at all levels in the lumbar spine with the corresponding results obtained by compressing 10 cadaveric lumbar spines (L1–S1) using the follower load technique described by Patwardhan et al. [1999]. A follower load increases the load-carrying capacity of the lumbar spine in compression. Spine 24(10), 1003–1009. Further validation of the model was performed by comparing the lateral bending and torsion response without pre-load and the flexion–extension response without pre-load and with an 800 N follower pre-load with those obtained using cadaver lumbar spines. Following validation, the FEM was subjected to bending moments in all three planes with and without compressive follower pre-loads of up to 1200 N. Disc compression values and the flexion–extension range of motion under 800 N follower pre-load predicted by the FEM compared well with \textit{in vitro} results. The current model showed that compressive follower pre-load decreased total as well as segmental ROM in flexion–extension by up to 18%, lateral bending by up to 42%, and torsion by up to 26%.

Keywords: Follower load; Lumbar spine; Kinematics; Finite element

1. Introduction

During tasks of daily living the lumbar spine withstands compressive loads of very high magnitude along with significant amounts of motion. Compressive loads can easily approach several thousand newtons during some lifting tasks (Nachemson, 1966; Schultz, 1987; Rohlmann et al., 1997). Physiologic compressive loads have been successfully applied to individual lumbar motion segments and the stiffening effect of compressive pre-load on single functional spine units has been investigated (Cripion et al., 2000, 2004; Gardner-Morse and Stokes, 2004; Janevic et al., 1991). However, difficulty arises in terms of stability of the lumbar spine when physiologic compressive loads are applied to the entire lumbar spine (Crisco et al., 1992a, b; Patwardhan et al., 1999).

Various mathematical models have been used to show that trunk muscle activation effectively allows for mechanical stable equilibrium of the lumbar spine (Bergmark, 1989; Crisco and Panjabi, 1990; Gardner-Morse et al., 1995). Patwardhan et al. (2001) used a mathematical model to identify muscle co-activation...
patterns in which the vector sum of trunk muscle forces produced a single internal force vector that acted tangent to the curvature of the lumbar spine passing through each segmental center of rotation, allowing the model to withstand 1200 N compressive loads with minimal change in lordosis. This “follower” load path tangent to the curvature of the spine thus mimics the physiologic compressive loads seen in vivo.

Using the follower load technique the compressive load is applied using bilateral cables attached to L1 passing freely through cable guides attached to the vertebral bodies of L2–L5 and over pulleys attached to S1. The path of each cable is adjusted such that the cable runs tangent to the lordotic curvature of the lumbar spine in neutral posture (Fig. 1). This load path is consistent with the balance point of lumbar motion segments where a compressive load can be applied that minimizes coupled motions (Wilder et al., 1989). Using the follower load technique 1200 N loads, replicating the compressive load on the lumbar spine during light tasks of daily living, have been successfully applied to the lumbar spine (Nachemson, 1966; Patwardhan et al., 1999; Patwardhan et al., 2003; Rohlmann et al., 2001). The effect of compressive pre-load on lumbar sagittal plane kinematics was studied by flexion–extension Patwardhan et al. (2003) and a 15–25% decrease in range of motion with the application of a 1200 N compressive follower load was found.

The application of a physiologic compressive load on the lumbar spine has also been studied using finite element modeling. Shirazi-Adl and Parnianpour (2000) and Shirazi-Adl (2006) developed a method to apply physiologic compression to the lumbar spine through the use of postural changes and “wrapping” elements which wrap around prescribed spatial targets in the center of the endplates of each motion segment such that the compressive load remains perpendicular to the mid-plane of each disc.

Physiologically, motion in the lumbar spine is not limited to the sagittal plane. Many activities associated with manual lifting include large amounts of lateral bending and torsion, which have increasingly been implicated as possible causes for low-back pain and disc degeneration. Thus, the full kinematic response of the entire lumbar spine under physiologic compressive loads is essential. The goal of this study, therefore, was to develop a new finite element model (FEM) by which physiologic compressive loads can be applied to the entire lumbar spine while allowing for the application of bending moments in all three planes, mimicking the loads and motions seen in vivo. It was hypothesized that a compressive follower pre-load would decrease the segmental range of motions (ROM) during flexion–extension, lateral bending, and torsion.

2. Methods

2.1. In vitro model

Testing on cadaveric specimens was done to validate the FEM because there is no data available in the literature using human subjects. Ten human cadaveric lumbar spines from L1 to Sacrum (seven male, three female, mean age = 58 years, range 41–73 years) were compressed to 1200 N in the neutral posture using the follower load technique (Patwardhan et al., 1999). Before testing, plane radiographs (GE OEC 9800 Digital Flouroscopy Machine, GE Medical Systems Inc., Waukeshe, WI) were taken to rule out bony abnormalities and MRI (Hitachi Airis II, Hitachi Medical Systems, Twinsburg, OH) was performed to rule out significant soft-tissue abnormalities and to ensure that each specimen had healthy discs of grade II or better as determined by an orthopedic surgeon. No preconditioning of the in vitro specimens was done, other than that done while optimizing the follower load path as described by Patwardhan et al. (1999). During testing, lateral radiographs were taken before compression (0 N), and with a 1200 N follower load. The heights of each disc under both loads were measured using ScionImage (Scion Corp., Frederick, MY) and taken as the average of the anterior, posterior, and middle disc heights.

In the experimental set up it was possible to apply compressive pre-load only along with flexion–extension moments. The ROM at each of the five levels was measured for moments of 8 N m in flexion and 6 N m in extension without pre-load and with an 800 N follower pre-load. A larger moment was applied in flexion than extension because the ROM of the spine is larger in
flexion than extension. An 800 N pre-load was applied to represent the average load placed upon the lumbar spine while standing and to limit the potential for specimen damage during testing. Lateral bending moments of $\pm 6 \text{ Nm}$ and torsional moments of $\pm 4 \text{ Nm}$ were applied to the cadaver lumbar spines without pre-load. As a consequence of the methods used to apply the moment loads in vitro, a small axial load of approximately 14 N was added to the spine, which is negligible compared to the pre-load used for the analyses. The flexion-extension and lateral bending rotation of each vertebral body due to the moment load was measured using bi-axial inclinometers (Model 902-45, Applied Geomechanics, Santa Cruz, CA). The torsion of each vertebral body due to the moment load was measured using an optoelectronic motion measurement system (Optotrak, NorthernDigital, Waterloo, Ontario). All angular measurements were made to within 0.1°. The total segmental motion of each motion segment, taken as the sum of the two corresponding motions in each plane (i.e. flexion-extension, right + left lateral bend, and right + left torsion) was calculated.

2.2. Non-linear three-dimensional FEM

The geometry of a three-dimensional lumbar finite element model (L1-S1) was created from serial axial computed tomography images of a 38 year-old female. The cortical bone, cancellous bone, posterior elements, endplates, facet cartilage, and nucleus pulposus were modeled as eight-node, three-dimensional, isoparametric elements. The articulating surfaces of the facet cartilage were approximated by flat trapezoidal moving frictionless contact surfaces (CONTACT 3; ADINA, Watertown, MA). In the intervertebral discs, the annulus matrix was assumed as a composite material consisting of fibers embedded in a homogenous matrix material. The matrix was discretized by eight-node, three-dimensional, Mooney–Rivlin hyperelastic elements. The annular fibers were assembled in a criss-cross fashion approximately 30° to the transverse plane and were modeled as two-node, non-linear truss elements that reacted to tension only. The normal nucleus, which is a gelatinous material, was represented by three-dimensional fluid elements. The seven major ligaments were modeled by two-node non-linear cable elements and their attachment points were taken directly from the literature.

The material properties (Table 1) for the intervertebral discs used in the current study were chosen to match those of a healthy spine with approximately grade II discs as seen in vitro. The material properties (Table 1) for the intervertebral discs used in the current study were chosen to match those of a healthy spine with approximately grade II discs as seen in vitro. Details about the other components of the model have been published previously (Natarajan and Andersson, 1999).

A follower load was simulated at each motion segment in the model through a pair of two-node thermo-isotropic truss elements. The follower load trusses were attached bilaterally to the cortical shell of the vertebral body of each motion segment such that each truss spanned the disc, approximately passing through the instantaneous center of rotation of each motion segment, optimizing the follower load path (Fig. 2). Compressive load was applied to each motion segment by inducing contraction in each of these truss elements by decreasing the temperature in each truss. Since the compressive force on each motion segment acts along the axis of the truss, even in the contracted state of the truss, no moment is induced due to compression of the motion segment. The amount of compressive load induced at each of the motion segments for a given temperature change was verified by fixing the bottom surface of each motion segment and calculating the reaction force in the follower load truss.

Two forces of equal and opposite magnitude were applied on the superior surface of the L1 vertebra in the appropriate direction and location to create the bending effect (flexion-extension, lateral bending, and torsion) of the lumbar spine.

| Material properties of the intervertebral discs in the finite element model (Goel et al., 1998; Sharma et al., 1995) |
|---|---|---|---|
| Annulus matrix Mooney–Rivlin material |
| C1, 2 | C3–5 | D1, 2 | $K \text{ (deg}^{-1})$ |
| 8.00E+05 | 50 | 4.50E+06 | 50 |
| Nucleus |
| $E \text{ (Pa)}$ | $\nu$ |
| L1–L4 | 2.50E+06 | 0.45 |
| L4–L5 | 3.00E+06 | 0.4995 |
| L5–S1 | 2.25E+06 | 0.4995 |

Fig. 2. Lateral view of the finite element model showing follower load trusses at each level.
the entire lumbar spine. The loads simulating flexion–extension and lateral bending remained vertical throughout the motion, which affected the actual magnitude of the applied moment. This method is an accepted practice and was used to duplicate the in vitro methods. An 8 Nm flexion moment, 6 Nm extension moment, ±6 Nm lateral bending moment and ±4 Nm torsional moment was applied to the lumbar spine. These moments were applied either with a pre-load (800 or 1200 N) or without pre-load. The ROM of each vertebra in the three planes (sagittal, coronal, and transverse) was determined by taking the sum of the two motions in each plane.

Validation of the FEM was performed by comparing FEM results at each of the five lumbar motion segments with the corresponding results from in vitro specimens for: (1) disc compression with 1200 N follower pre-load, (2) flexion–extension ROM, lateral bending ROM and torsion ROM without pre-load, and (3) flexion–extension ROM with 800 N follower pre-load. If the FEM results fell within one standard deviation of the in vitro results the FEM was considered validated.

The FEM was then used to determine how compressive pre-load affects the flexibility of the entire lumbar spine. Moments in all three planes were applied to the superior aspect of L1 with and without a follower pre-load and the ROM was determined for all loading conditions. A comparison was made between the ROM without pre-load and with follower pre-load in order to determine the effect of pre-load on ROM.

3. Results

Applying compressive forces using thermo-isotropic truss elements, it was possible to mimic the follower load path that a normal lumbar spine encounters in vivo. A 1200 N follower load on the entire lumbar spine produced maximum rotations of 0.9° in the sagittal plane, 0.4° in the transverse plane, and 0.2° in the frontal plane, indicating that the load applied using the current technique corresponded to a follower load path.

3.1. FEM validation

FEM predictions of disc compression at all five levels compared well with in vitro results. In vitro disc compressions under a 1200 N load ranged from 1.2±0.3 to 1.6±0.5 mm. Results obtained from FEM varied from 1.4 to 2.0 mm and fell within one standard deviation of the in vitro results at all levels (Fig. 3).

The response of the FEM to moment loading in all three planes, without pre-load, also compared well with in vitro results. The segmental flexion–extension ROM without pre-load predicted by the FEM fell within one standard deviation of the in vitro results at all levels except L2–L3, L4–L5, and L5–S1 (Fig. 4). The L2–L3 motion, however, fell only 0.6° above one standard deviation of the in vitro results. Mean flexion–extension ROM of L1 with respect to S1 predicted by the FEM (51.1°) closely matched the in vitro results (49.7±9.7°). The torsion ROM predicted by the FEM fell within one standard deviation of the in vitro results at all levels except L1–L2, where a slight under-prediction was seen (Fig. 5). The L1/S1 torsional ROM predicted by the FEM was 9.1°, whereas the in vitro motion was 12.9±3.3°. Thus, the finite element L1/S1 rotation was

![Fig. 3. In vitro and finite element disc compression with 1200 N follower load. Compression predicted by the finite element model compares well with in vitro results at all levels.](image)

![Fig. 4. In vitro and finite element total segmental flexion–extension ROM without follower pre-load. Motion predicted by the finite element model compares well with in vitro results at all levels except L2–L3 and L5–S1 where small over predictions, and at L4–L5 were a slight under prediction was seen, respectively.](image)

![Fig. 5. In vitro and finite element total segmental torsion ROM without follower pre-load. Motion predicted by the finite element model compares well with in vitro results at all levels except L1–L2 where a small under prediction is seen.](image)
only 0.6° from one standard deviation of the in vitro results. The segmental lateral bending ROM predicted by the FEM fell within one standard deviation of the in vitro results at all levels except L4–L5 where, an under-prediction was seen (Fig. 6). The L1/S1 lateral bending predicted by the FEM was 31.7°, whereas the in vitro motion was 48.4 ± 10.5°. Thus, the finite element L1/S1 lateral bending was within 12% of one standard deviation of the in vitro results.

The response of the FEM to sagittal moment loading with compressive pre-load also compared well with the in vitro results. The segmental flexion–extension ROM with an 800 N pre-load predicted by the FEM fell within one standard deviation of the in vitro results at all levels except L2–L3 and L5–S1 where a slight over-prediction in segmental rotation was seen (Fig. 7). The L1/S1 flexion–extension predicted by the FEM with an 800 N follower preload was 49.1°, whereas the in vitro motion was 39.0 ± 7.6°. Thus, the finite element L1/S1 flexion–extension was within 6.7% of one standard deviation of the in vitro results.

3.2. Effect of pre-load on range of motion

Application of a follower pre-load decreased the L1/S1 ROM in flexion–extension, lateral bending and torsion. Follower pre-load of 800 N decreased the L1/S1 ROM by 3.7% in flexion–extension, 4.7% in lateral bending and 3.3% in torsion. The corresponding decrease in ROM under 1200 N pre-load was much higher: 18.4% in flexion–extension, 22.3% in lateral bending and 12.1% in torsion.

Follower compressive pre-load had a stiffening effect on ROM at almost all five segment levels as well. The stiffening effect was more prominent under flexion–extension and lateral bending. With increasing compressive pre-load, flexion–extension ROM monotonically decreased at all segment levels except L4–L5. The decrease in segmental flexion–extension ROM varied from 1.1° to 1.3° with 800 N compressive load and from 0.7 to 3.0° with 1200 N compressive pre-load (Fig. 8).

An 800 N follower pre-load stiffened the lateral bending motion at some levels and made the segments more flexible at other levels. However, these changes in ROM were small (between 0.4° and 0.9°). Larger decreases in segmental lateral bending ROM were seen at all levels, except L4–L5 with a 1200 N compressive pre-load (0.8–2.9°) (Fig. 9). Compressive pre-loads of both magnitudes had the least effect on segmental torsion ROM where the maximum change was only 0.8° (Fig. 10).

4. Discussion

A three-dimensional non-linear lumbar spine FEM has been developed that can be used to evaluate the full kinematic response of the lumbar spine subjected to a large follower pre-load. The current study demonstrates that a set of thermo-isotropic truss elements can successfully apply a large follower pre-load with minimal rotation at each level of the lumbar spine, thus
achieving an optimized follower load. The compressive deformation at each level predicted by the FEM compared favorably with those measured in vitro using the follower load concept, validating the current method of applying a compressive pre-load to the entire lumbar spine. Results of flexion–extension ROM of the lumbar spine under a compressive pre-load compared well with in vitro values. Therefore, the novel method of applying pre-load to the spine can be used to study motions in a spine with compressive pre-load subjected to moment loads.

At some segment levels, the FEM results did not fall within one standard deviation of in vitro results. This discrepancy may be attributed to slight variations in disc degeneration grade from level to level that are common in cadaver specimens, which are not accounted for in the model. Also, because of the viscoelastic nature of intervertebral discs, should the in vitro discs dehydrate slightly during testing, the amount of motion seen at that level will change. This was also not accounted for in the model and may attribute to the differences seen between the in vitro and finite element results.

One of the disadvantages of the current in vitro technique is that it does not allow for any variations in muscle forces at any one of the segments that might arise due to injury or disease. The current finite element modeling method, which applies the compressive pre-load segmentally, can be easily modified to apply varying amounts of compression at different levels in the lumbar spine. Thus, various disruptions in muscle function, which would in turn disrupt the follower load path, can be modeled by the current FEM.

Both the reduction in overall L1/S1 flexion–extension motion with a 1200 N follower pre-load predicted by the FEM and the trend toward increasing reduction in flexion–extension motion with increasing compressive pre-load compared favorably to values previously reported (Patwardhan et al., 2003).

The largest decrease in motion due to a compressive pre-load of 1200 N was seen at L5–S1 for all the three moment loadings. The largest percent decrease in motion due to pre-load was in lateral bending (42%), whereas the percent decrease in torsion was 26% and only 18% in flexion–extension. Therefore the hypothesis that a compressive pre-load would decrease the segmental ROM during flexion–extension, lateral bending, and torsion was supported. The largest decrease in segmental rotation was seen at L5–S1, the level that is often the first to degenerate and cause pain. Increased rotation has been implicated as a key factor in the creation of a pars defect, thus pre-load, which was shown to decrease the L5–S1 rotation, may play a role in the prevention of spondylolysis and spondylolisthesis.

The results of this study demonstrate the ability of a large follower load to stiffen the spine in all three planes. Because the follower load concept acts to mimic optimized muscles forces, this study illustrates the important role of muscles in providing spinal stability. Recently, Shirazi-Adl (2006) has shown that with the help of ‘wrapping’ elements, a physiologic load can be applied to the entire lumbar spine. The methods used here more closely resemble the in vitro technique. Though the current model does not represent the full complexity of the in vivo situation, it is the closest, to date, to the actual in vivo situation in terms of the combination of compressive loading conditions and motions, and can, therefore, more closely replicate actual in vivo conditions than any previous method. This allows for a more thorough study of lumbar spine kinematics under a follower pre-load. This method therefore, can be used to more completely evaluate the effects of various deformities, disease conditions, and surgical constructs and techniques on spinal stability and kinematics under the same type and combination of complex loading conditions seen in vivo.
References


