FINITE ELEMENT ANALYSIS OF BIOMECHANICAL STABILITY OF ISTHMIC SPONDYLOLISTHESIS WITH POSTEROLATERAL FUSION WITH AND WITHOUT PHYSIOLOGIC COMPRESSIVE LOADS

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INTRODUCTION: Adult onset isthmic spondylolisthesis, most common at L5-S1 with an incidence rate of 6%, may require surgical intervention. The most common treatment option is posterolateral fusion (PLF) using transpedicular instrumentation. Previous models of spondylolisthesis have included a defect of the pars without including the anterior deformity typical of spondylolisthesis, or have focused on only one motion segment. Thus, to date, no objective biomechanical data exists on the stabilizing effect of PLF in a realistic model of spondylolisthesis using the entire lumbar spine. Therefore the goal of this study was twofold: (i) to create a realistic finite element model of unstable isthmic spondylolisthesis and (ii) to evaluate the biomechanical effectiveness of PLF to stabilize an unstable isthmic spondylolisthesis.

METHODS: A 3-D finite element model created from CT images including six vertebrae with posterior elements, five discs, 6 ligaments per level, and 2 pairs of facet joints at each level was used in the current study. The cortical bone, cancellous bone, posterior elements, endplates, facet cartilage, annulus, and nucleus were modeled as 8-node, three dimensional, isoparametric elements. The left and right superior and inferior articulating surfaces of the facet cartilage were modeled by moving frictionless contact surfaces. The annular fibers of the intervertebral discs were assembled in a criss-cross fashion at an angle approximately 30° to the transverse plane. The ligaments and annular fibers were modeled as 2-node, non-linear truss elements active in tension only. All material properties were taken from the literature.

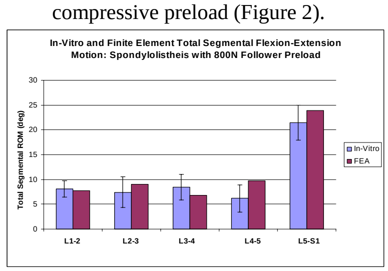
A bilateral defect of the pars interarticularis of L5 was created by removing the appropriate elements in the pars of L5 and the exposed surfaces were then covered with contact surfaces (CONTACT 3, ADINA, Watertown, MA). To create the spondylolisthetic deformity, an anterior displacement was applied to the anterior and posterior aspect of L5 such that a 16.5% anterior slip was produced at L5-S1. Following testing of the spondylolisthetic condition, transpedicular instrumentation was added to L5-S1. Only the screw heads and rods were modeled due to limitations associated with inserting 3-D bodies into the vertebrae which were modeled as volumes. Each screw head was modeled as a cylinder with a radius of 5mm and attached to the posterior elements of the appropriate vertebra using rigid links (RIGIDLINKS). Each rod was also modeled as a cylinder with a radius of 2.75mm and a rigid connection was assumed between the screw head and rod. The PLF construct was assigned material properties for surgical titanium (ASTM F2066).

In the current finite element model a follower load was applied through 2-node thermo-isotropic truss elements attached to the cortical shell of the vertebrae of each motion segment bilaterally and used to compress each of the lumbar spine motion segments.[1]

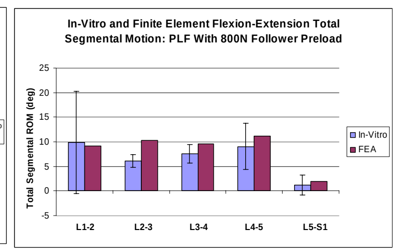
The model has been previously validated against data from 10 cadaveric lumbar spines in compression and flexion-extension range of motion for the intact condition.[1] To further validate the model, the same ten specimens used to validate the intact model was used. A spondylolisthetic deformity was created by performing a partial discectomy and applying a shear load at L5-S1 until a Grade I sip was produced. Following the production of the slip, each spine was subjected to 1200N compressive load using the follower load technique[2] and the amount of compression at each disc was measured using digital fluoroscopy. Each spine was then loaded to 8Nm in flexion and 6Nm in extension without preload and with a follower preload of 800N and the ROM of each vertebra measured. Then, transpedicular instrumentation was inserted in L5-S1 and the compressive and moment loading was repeated. The same compressive and moment loads were then applied to the spondylolisthesis and PLF finite element models via the follower load trusses and vertical forces applied to the top anterior and posterior aspect of L1 for flexion-extension. The respective disc compressions and kinematics compared with the *in-vitro* results.

RESULTS: The application of the compressive follower preload on the spondylolisthesis model produced small segmental rotations of less than 2° in any plane. Disc compression predicted by the spondylolisthesis finite element model closely agreed with the *in-vitro* mean values at all lumbar spine levels except L2-3 and L4-5. The total segmental flexion-extension motion predicted by the spondylolisthesis model with and without preload also closely matched the *in-vitro* results, falling within one standard deviation at each lumbar segment (Figure 1 A). An increase in L5-S1 total flexion-extension ROM was seen with spondylolisthesis as compared to intact with and without a compressive preload, while having less of an affect on the adjacent levels (L1-5) (Figure 2).

The application of a compressive follower load to the PLF finite element model also produced small segmental motions of less than 3° in any plane. As with the spondylolisthesis model, disc compression predicted by the PLF finite element model closely agreed with the *in-vitro* mean values at all lumbar spine levels except L3-4 and L4-5. The total segmental flexion-extension motion predicted by the PLF model with and without preload also closely matched the *in-vitro* results, falling within one standard deviation at each lumbar segment except L2-3 (Figure 1 B). PLF decreased the amount of compression seen at L5-S1 below intact levels without greatly affecting the upper levels (L1-5) PLF greatly reduced the L5-S1 segmental ROM below intact levels while slightly increasing the L4-5 segmental ROM both with and without compressive preload (Figure 2).



A



B

Figure 1: (A) Validation using i*n*-vitro and finite element spondylolisthesis F-E ROM with 800N preload (B) Validation using i*n-vitro* and finite element PLF F-E ROM with 800N preload.

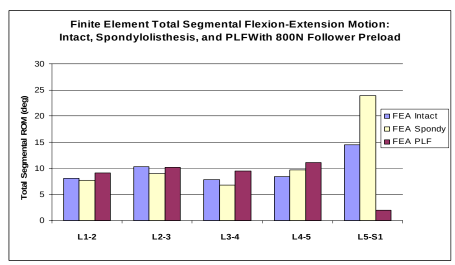


Figure 2: Finite element, intact, spondylolisthesis and PLF F-E ROM with 800N follower preload.

DISCUSSION: With the help of a set of thermo-isotropic truss elements, it was possible to apply a nearly pure large compressive load to the intact, spondylolisthesis, and PLF models with minimal rotational motion at each lumbar disc level. The compressive deformation at each level predicted by the current finite element model compared favorably with those measured using the follower load concept in an *in-vitro* test intact and with spondylolisthesis and PLF.

The increase in compression and flexion-extension motion at L5-S1 with spondylolisthesis seen with the model mimics the instability seen with spondylolisthesis *in vivo*. The decrease in compression and flexion-extension motion seen with the inclusion of transpedicular instrumentation in PLF demonstrates its effectiveness to stabilize full lumbar spine with an unstable isthmic spondylolisthesis with only a small affect on adjacent segment motion.

However, the loading conditions seen with spondylolisthesis *in vivo* are quite complex and some shear load may be present, especially at L5-S1. Therefore, the PLF construct must be tested under conditions that better represent these complex loading conditions seen with spondylolisthesis *in vivo.*

REFERENCES: (1) Renner, S. M. et al. ORS 2005 (2) Patwardhan, A.G., et al. Spine 24(10): 1003-1009, 1999.