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Effect of a posterior dynamic implant adjacent to a rigid spinal fixator

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Abstract

Background. A slightly degenerated disc adjacent to a segment that has to be fused is sometimes instrumented with a dynamic fixator. The dynamic implant is assumed to reduce disc loads at that level and to preserve disc function, thus inhibiting the progression of degeneration.

Methods. A three-dimensional finite element model of the lumbar spine was used to study the effect of a dynamic implant on the mechanical behavior at the corresponding level. After studying a healthy lumbar spine for comparison, a rigid fixator and a bone graft were inserted at L2/L3. Healthy and degenerated discs were assumed at the adjacent level, i.e. L3/L4. An additional paired dynamic posterior fixator was then implemented at level L3/L4. Finally, the segment with the dynamic fixator was distracted to the height of a healthy disc. The loading cases of walking, extension, flexion and axial rotation were simulated.

Findings. A dynamic implant reduces intersegmental rotation for walking, extension and flexion as well as facet joint forces for axial rotation at its insertion level. Intradiscal pressure is not markedly reduced by a dynamic implant. Moreover, there are no substantial differences between the mechanical behavior of rigid and dynamic fixators.

Interpretation. Our model does not predict major differences in the mechanical effects between rigid and dynamic fixators despite the extreme assumption that a dynamic implant does not transfer moments. The results do not support the assumption that disc loads are significantly reduced by a dynamic implant. For axial rotation, however, dynamic fixation devices do reduce the force in the facet joint.
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1. Introduction

A spinal segment adjacent to one that has to be fused is sometimes slightly degenerated. This raises the question of what to do with this segment. Some surgeons recommend bridging with a so-called 'dynamic' implant that is semi-rigid. A monosegmental rigid spinal fixation device combined with anterior interbody fusion drastically reduces motion in the treated segment. This leads to greater deformation in the adjacent segments if the overall deformation of the spine is predetermined. This increased motion causes higher stresses and is believed to accelerate the degeneration process. A dynamic posterior implant is assumed to reduce disc loads at implant level while preserving its elastic function, thus inhibiting the progression of degeneration in the affected segment (Putzier et al., 2005).

Several dynamic posterior spinal fixation systems are clinically applied in addition to various rigid fixators. These implants are supposed to reduce the load on a slightly degenerated disc and on the facet joints. Contrary to solid fusion, the non-fusion systems are intended to maintain intersegmental motion or restore it to the degree found in healthy spines. The Graf™ ligament system (SEM Co., Montrouge, France) consists of a posterior inductile band that serves as a ligament between two pedicle-based screws (Nockels, 2005). The dorsal transpedicularly fixed dynamic neutralization system (Dynesys®, Zimmer Spine Inc., Minneapolis, MN, USA) has been in clinical use for more than a decade. It comprises pedicle screws, spacers and cords (Stoll et al., 2002). Graf™ ligamentoplasty and the Dynesys®

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system appear to yield similar clinical results (Grob et al., 2005).

Schmoelz et al. (2003) experimentally determined the degree of stabilization achieved by Dynesys®. For the bridged segment, they found that it stabilized the spine and was more flexible than an internal fixator, especially in extension. The measured changes in intradiscal pressure during loading in flexion, extension and lateral bending were similar for both rigid and dynamic stabilization of a bridged segment (Schmoelz et al., 2006).

Eberlein et al. (2002) created a finite element model of an L2–L3 motion segment with Dynesys® and determined the effect of the implant on the overall stiffness. They found it to cause considerable stiffening in flexion and extension and extreme stiffening in torsion.

The effects of mono- and bisegmental fixators on the mechanical behavior of the lumbar spine have been investigated by Zander et al. (2002a,b). Their finite element model did not predict clear differences between mono- and bisegmental fixation at implant level. They showed that pretension at the bridged level strongly affects spinal loads. A paired monosegmental posterior dynamic implant has been studied for its influence on lumbar spine loads by Rohlmann et al. (submitted for publication). They calculated reduced intersegmental rotation at implant level for flexion and extension, lower intradiscal pressure for extension, and decreased facet joint forces at implant level. A dynamic implant had only slightly less pronounced effects than a rigid fixator, although implant stiffnesses differed by more than the factor 200.

Disc degeneration is normally accompanied by a reduction of disc height. During surgery the affected level is often distracted by the implant to the height of a healthy disc. It is useful to know how distraction combined with a dynamic implant affects the mechanical behavior of the lumbar spine.

The aim of this study was to determine how a dynamic implant adjacent to a rigid fixator affects the mechanical behavior of the lumbar spine. We hypothesized that a dynamic implant adjacent to a rigid one reduces intradiscal pressure and lowers forces in the facet joints at its insertion level.

2. Methods

A three-dimensional nonlinear finite element model of the osseoligamentous lumbar spine was created using eight-node hexahedral elements for the five vertebrae (Fig. 1). The annuli fibrosi of the discs were modeled with eight-node volume elements representing the ground substance and tension-only spring elements representing the fibers. The fibers were arranged in two times four layers in radial directions, their orientation alternating between about 30° and 150° to the mid-cross-sectional area of the disc. Fiber stiffness increased from the center to the outer shell (Shirazi-Adl et al., 1986). The element mesh was eight times finer at L3/L4 than at the other levels. The displacement of the additional nodes at the border between the coarse and fine mesh was linear interpolated to the adjacent nodes by MPC (multi-point constraints) elements. Fibers at that level were arranged in two times seven layers. The fiber stiffness at that level was adopted so that the refinement of the element mesh did not change the stiffness of the disc. The finer mesh results in a more precise stress distribution within the disc. The nuclei pulposi were modeled as incompressible fluid-filled cavities. The facet joints had a gap of 0.5 mm and could transmit only compressive forces. All seven ligaments of the lumbar spine were included. Material properties of the different tissues were taken from the literature (Table 1), (Goel et al., 1995; Rohlmann et al., 1980, 2005a; Rohlmann et al., in press; Shirazi-Adl et al., 1984, 1986; Zander et al., 2001); the fibers and ligaments have been described in detail elsewhere (Rohlmann et al., 2005a; Zander et al., 2001). A relatively low elastic modulus was chosen for cortical bone to compensate for the higher thickness of the corresponding elements.

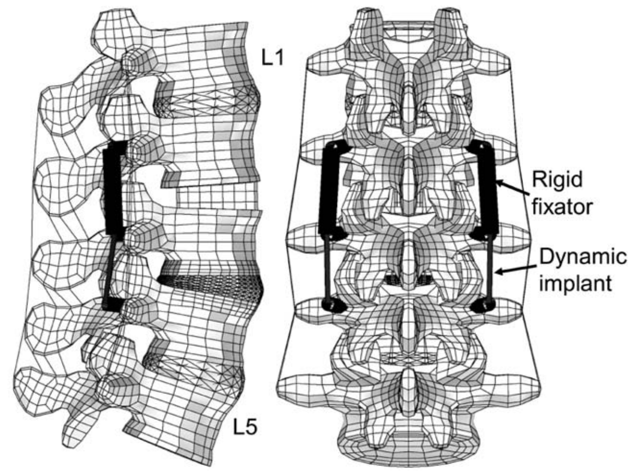

Fig. 1. Finite element model of the lumbar spine with a rigid fixator and a bone graft at L2/L3 plus a degenerated disc and a posterior dynamic implant at L3/L4.

The inferior endplate of the L5 vertebra was rigidly fixed. The model was loaded with the upper body weight and muscle forces to simulate the four loading cases: stance phase of walking, 50° flexion (30° in the hip joints and 20° in the lumbar spine), 25° extension (10° in the hips and 15° in the lumbar spine) and 10° left axial rotation. Flexion and extension of the hip joint was simulated by tilting the model which changed the center of gravity relative to the lumbar spine. The muscle forces for standing, flexion and extension have been estimated in previous studies (Rohlmann et al., 2006; Wilke et al., 2003; Zander et al., 2001). Spinal loads were assumed to be 25% higher for walking than for standing (Rohlmann et al., 2001a). For standing, flexion, and extension, an upper body weight of 260 N and a follower load of 200 N were assumed. The upper body weight acted in vertical direction 30 mm anterior of the center of the L1 vertebra (Rohlmann et al., 2006). The follower load (Patwardhan et al., 1999; Rohlmann et al., 2001b) with a force direction following

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