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Biomechanical effects of fusion levels on the risk of proximal junctional failure and kyphosis in lumbar spinal fusion surgery

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ABSTRACT

Background: Spinal fusion surgery is a widely used surgical procedure for sagittal realignment. Clinical studies have reported that spinal fusion may cause proximal junctional kyphosis and failure with disc failure, vertebral fracture, and/or failure at the implant-bone interface. However, the biomechanical injury mechanisms of proximal junctional kyphosis and failure remain unclear.
Methods: A finite element model of the thoracolumbar spine was used. Nine fusion models with pedicle screw systems implanted at the L2–L3, L3–L4, L4–L5, L5–S1, L2–L4, L3–L5, L4–S1, L2–L5, and L3–S1 levels were developed based on the respective surgical protocols. The developed models simulated flexion–extension using hybrid testing protocol.
Findings: When spinal fusion was performed at more distal levels, particularly at the L5–S1 level, the following biomechanical properties increased during flexion–extension: range of motion, stress on the annulus fibrosus fibers and vertebra at the adjacent motion segment, and the magnitude of axial forces on the pedicle screw at the uppermost instrumented vertebra.
Interpretations: The results of this study demonstrate that more distal fusion levels, particularly in spinal fusion including the L5–S1 level, lead to greater increases in the risk of proximal junctional kyphosis and failure, as evidenced by larger ranges of motion, higher stresses on fibers of the annulus fibrosus and vertebra at the adjacent segment, and higher axial forces on the screw at the uppermost instrumented vertebra in flexion–extension. Therefore, fusion levels should be carefully selected to avoid proximal junctional kyphosis and failure.

1. Introduction

Degenerative spinal deformities such as sagittal imbalance and lumbar scoliosis are an increasing clinical problem in the aging population (Barrey et al., 2011; de Vries et al., 2010). One surgical procedure for treating these spinal disorders is spinal fusion with pedicle screw systems (Kim et al., 2006b, 2007b). However, spinal fusion may accelerate the degeneration of the adjacent level because the rigidity of the fused level increases segmental motion and intradiscal pressure (Lee and Langrana, 1984; Weinhoffer et al., 1995). In addition, spinal fusion, particularly multi-level fusion, may cause proximal junctional kyphosis (PJK), a postoperative deformity in which the proximal junctional sagittal angle is abnormally greater than the pre-operative angle, and proximal junctional failure (PJF), such as screw pull-out, superior endplate fracture at the uppermost instrumented vertebra (UIV), and vertebra fracture or listhesis at the adjacent level (Glattes et al., 2005; Kim et al., 2005, 2006b, 2007a; Murray, 2012; Swank, 2002; Watanabe et al., 2010; Yagi et al., 2011). The reported prevalence of PJK is 20%–39% at a follow-up of at least 2 years, while the reported prevalence of PJF (Glattes et al., 2005; Kim et al., 2008; Yagi et al., 2011) is 6% within 28 weeks after operation (Hostin et al., 2013).

Clinical studies have reported several factors that contribute to PJK and PJF: (1) patient condition, such as age (Hostin et al., 2013; Kim et al., 2006b, 2008; Watanabe et al., 2010) and pre-operative sagittal balance (Hostin et al., 2013; Watanabe et al., 2010; Yagi et al., 2011); (2) surgical plan and method, such as combined anteroposterior spinal fusion (Kim et al., 2008), fusion to the sacrum and posterior fusion (Yagi et al., 2011), and incorrect selection of the upper end vertebra (Denis et al., 2009); and (3) surgical outcome, such as unsuccessful fusion at the UIV (Denis et al., 2009). Moreover, Cammarata et al. analyzed the effects of surgical method, implant type at the UIV, and rod shape and size of pedicle screw systems on PJK using finite element (FE) models of six adult scoliosis patients (Cammarata et al., 2012, 2014). However, the rationales and mechanisms of proximal joint problems have not been fully elucidated because clinical studies have provided limited biomechanical information and many factors with implications for PJK and PJF remain to be clarified. For example, the potential association of surgical range and fusion level with PJK and PJF has not been considered.

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In this study, we investigated the biomechanical effects of fusion level on proximal junctional problems, including PJK and PJF, using FE analysis. FE models of the thoracolumbar spine from T12 to the sacrum with nine different spinal fusions (four one-level, three two-level, and two three-level) were developed using a validated FE model of a healthy lumbar spine. Several biomechanical properties, including the range of motion (RoM) of individual motion segments, stress on annulus fibrosus fibers and vertebra, and the force on the pedicle screws after various spinal fusions, were predicted and compared. The effect of fusion level on these biomechanical properties was analyzed.

2. Methods

2.1. Development of the FE model for an intact thoracolumbar spine

A three-dimensional FE model of the thoracolumbar spine (T12–S1) was developed based on a previously validated FE model of the lumbar spine (L1–S1) using 1-mm-thick computed tomography (CT) images (Park et al., 2013). The FE model was developed to be symmetric across the mid-sagittal plane and comprised seven vertebrae, six intervertebral discs, and seven types of major ligaments. The material properties used in the model were obtained from previously published literature, and ligament attachment points were determined based on anatomical information (Goel et al., 1995; Guan et al., 2006; Kim et al., 2010; Lu et al., 1996; Natarajan and Andersson, 1999; Natarajan et al., 2000; Park et al., 2009; Rohlmann et al., 2006; Ruberte et al., 2009; Schmidt et al., 2006, 2007a, 2007b, 2009; Shirazi-Adl et al., 1986; Ueno and Liu, 1987; Wagner and Lotz, 2004). The FE models of spinal bones included the cortical (Young's modulus (E) = 12,000 MPa, Poisson's ration (ν) = 0.3), cancellous (E = 100 MPa, $\nu = 0.2$), and post bones (E = 3500 MPa, $\nu = 0.24$) using linear elastic solid elements. Endplates were modeled on the superior and inferior planes of the vertebrae using linear elastic solid elements (E = 23.8 MPa, $\nu = 0.4$). The FE model of the intervertebral disc, which consisted of the nucleus pulposus, annulus ground substance, and annulus fibers, was modeled using an incompressible fluid cavity, hyperelastic solid elements, and strain-dependent tension-only truss elements, respectively. A compressibility of 0.0005 mm^2/N was used for the fluid cavity of the nucleus pulposus, and the Mooney–Rivlin hyper-elastic material property was adopted for the annulus ground substance. Tension-only truss elements with non-linear elastic material properties were used for the seven major ligaments (Rohlmann et al., 2006).

2.2. Development of the FE models of thoracolumbar spines with various spinal fusions

A posterior rigid fixation system with pedicle screws (cylindrical type, diameter = 6.5 mm, length = 43.5 mm, thread depth = 1.0 mm, and each tread pitch = 3.0 mm) and rods (diameter = 6.0 mm) was modeled using tetrahedral elements with the average element size of 0.7 mm. Four one-level fusion models (L2–L3, L3–L4, L4–L5, and L5–S1 fusions), three two-level fusion models (L2–L4, L3–L5, and L4–S1 fusions), and two three-level fusion models (L2–L5 and L3–S1 fusions) were developed based on established surgical protocols and pre-operative planning by a spine surgeon (Fig. 1) (Kabir et al., 2010; Lo et al., 2011; Wilke et al., 2008). Facet joints and capsular and flaval ligaments were resected in the fusion models. Titanium alloy (E = 110 GPa, $\nu = 0.30$) was assumed for the posterior rigid fixation system. The screws were rigidly fixed with the rods and three-dimensional surface-to-surface contact with a friction coefficient of 0.2 was applied to each contact region between the spinal bones and pedicle screws (Zhang et al., 2004, 2006).

2.3. FE analysis of the thoracolumbar spine with various spinal fusions

The biomechanical properties of the thoracolumbar spine fused at various ranges and levels were investigated during flexion–extension motions. The sacrum was fixed in all directions, and bending moments of 7.5 Nm simulating flexion–extension motions were applied to the superior plane of the T12 vertebra in the intact model with a compressive force of 400 N along the follower load direction (Fig. 2). The fusion models were simulated using hybrid testing protocol (Panjabi et al., 2007; Panjabi, 2007; Ruberte et al., 2009). The bending moments that generated flexion–extension rotation angles identical to those observed in the intact model were calculated for each fusion model. The calculated moments were applied to the superior plane of the T12 vertebra with a compressive force of 400 N.

The following biomechanical properties that have been indicated as risk factors for PJK and PJF in previous studies were predicted: RoM, tensile stress on annulus fibrosus fibers, von Mises stress on vertebra of the adjacent segment, and the axial force on the pedicle screw (pull-out force) at the UIV (Kim et al., 2006b, 2007a; Murray, 2012; Swank, 2002; Watanabe et al., 2010; Yagi et al., 2011). RoM was calculated at each motion segment, and the maximum tensile stress on fibers at the upper level adjacent to the fusion was normalized to the values obtained using the intact model. The maximum von Mises stresses on the vertebra at the proximal adjacent level were also normalized to the values obtained using the intact model. The magnitude of the axial force on the pedicle screw at the UIV was calculated during flexion–extension to investigate the risk of failure of the implant-bone interface after spinal fusion surgery. Because spinal fusion affects not only the upper adjacent level but also other healthy levels, changes in biomechanical behavior at the upper adjacent level and UIV were investigated to determine the risk of PJK and PJF. The commercial FE analysis software ABAQUS Standard™ ver. 6.10 (Simulia Corp., USA) was used in this study.

3. Results

3.1. Intersegmental rotation

The ROMs during flexion–extension in the intact model were 7.1°, 7.1°, 7.5°, 7.3°, 9.4°, and 12.6° at the T12–L1, L1–L2, L2–L3, L3–L4, L4–L5, and L5–S1 motion segments, respectively. With spinal fusion, the RoM of non-instrumented levels increased. At the upper adjacent levels, the RoM increased by 11.1%–33.8%, 26.1%–66.3%, and 59.5%–85.4% in the one-, two-, and three-level fusion models, respectively, compared to the intact model (Fig. 3). When fusion was performed at a more distal level and the range of fusion was expanded, a greater increase in the RoM of non-instrumented levels was predicted.

3.2. Stress on the annulus fibrosus at the proximal adjacent level

The maximum tensile stresses on the annulus fibrosus fibers were predicted at the posterior region in flexion and at the anterior region in extension. The values at T12–L1, L1–L2, L2–L3, L3–L4, L4–L5, and L5–S1 levels were 15.2 MPa, 18.0 MPa, 15.1 MPa, 17.5 MPa, 40.7 MPa, and 19.6 MPa, respectively in flexion; 14.4 MPa, 6.0 MPa, 2.7 MPa, 4.4 MPa, 5.0 MPa, and 15.1 MPa, respectively in extension. The stresses during flexion for the one-, two-, and three-level fusion models increased by 21.1%–87.8%, 60.2%–239.4%, and 219.2%–355.9%, respectively, from the intact model values. With the exception of the L3–L4 fusion model, the stresses during extension increased by 11.9%–28.1%, 44.4%–128.3%, and 77.9%–517.1% in the one-, two-, and three-level fusion models (Fig. 4), respectively. One-, two-, and three-level fusion models that included the L5–S1 levels featured greater stress on the annulus fibrosus fibers compared to fusion models that excluded the L5–S1 levels.

3.3. Stress on the vertebra at the proximal adjacent level

The maximum von Mises stresses on the T12, L1, L2, L3, L4, and L5 were 12.2 MPa, 10.9 MPa, 10.3 MPa, 10.7 MPa, 7.8 MPa, and 10.8 MPa, respectively in flexion; 9.9 MPa, 5.0 MPa, 3.7 MPa, 6.7 MPa, 5.5 MPa,