Effects of rod stiffness and fusion mass on the adjacent segments after floating mono-segmental fusion: a study using finite element analysis

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Abstract

Purpose The aims of the present study were to compare the biomechanical effects on the adjacent segments after mono-segmental floating fusion with posterior semi-rigid or rigid stabilization, and to evaluate the effect of the amount of fusion mass on the biomechanical differences.

Methods A detailed, nonlinear L1–S1 finite element model had been developed and validated. Then five models were reconstructed by different fixation techniques on the L3–L4 level: rigid fixation with an interbody spacer (Ti + IS), rigid fixation with a large interbody spacer (Ti + IS_all), semi-rigid fixation with an interbody spacer (PEEK + IS), semi-rigid fixation with a large interbody spacer (PEEK + IS_all), and semi-rigid fixation only (PEEK). Analyses were conducted for the case of erect standing position, flexion, and extension motion.

Results At L1–L2 and L2–L3, PEEK + IS demonstrated less inter-segmental rotation and nucleus pressure increments from the intact model compared with Ti + IS. The L4–L5 and L5–S1 levels showed slightly higher values with PEEK + IS, but these differences among the

instrumented models were not significant. The motion difference based on the fusion mass at the adjacent levels was at most 3 %. All instrumentation cases generated a 55 % higher facet contact force at the lower adjacent level (L4–L5) compared to that of the intact model during 26° extension and the largest increment was detected at the upper adjacent level (L2–L3) in the Ti + IS. Instrumentation with Ti + IS markedly increased the stress in the intervertebral disk at the upper adjacent level, while the stress with PEEK + IS appeared largest at the lower adjacent level.

Conclusions Posterior instrumentation with semi-rigid rods may lower the incidence of disk and facet degeneration in the upper adjacent segment compared to rigid rods. On the other hand, the possibility of facet degeneration will be similar for all instrumentation devices in the lower adjacent segment in the long-term. The stiffness difference between rigid and semi-rigid rods on the changes in the adjacent motion segments was more crucial than amount of fusion mass.

Keywords Adjacent segment degeneration · Finite element analysis · Rod stiffness · Amount of fusion mass

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Introduction

Adjacent segment degeneration (ASD) after spine fusion is a troubling problem because it can lead to increased pain and functional loss, as well as possible revision surgery. Thus, it has been one of the most important complications and problems of spinal fusion. It is not fully understood whether these changes occur as a direct result of the fusion or as a natural progression of the initial spinal degeneration. There has also been debate about the exact incidence

and risk factors of ASD. This is because ASD diagnosed radiographically does not match symptomatic ASD. Recently, the annual incidence of symptomatic ASD was 3.9 % as reported by Ghiselli et al. [1] and 2.5 % according to Sears et al. [2]. Sears et al. also reported that the mean annual incidences in the first 10 years after a lumbar fusion were 1.7 % for a single level, 3.6 % for two levels, and 5.0 % for three and four levels. The 5- and 10-year prevalence was 9 and 16 % for a single level, 17 and 31 % for two levels, and 29 and 40 % for three and four levels, respectively.

Known risk factors of ASD at the upper adjacent segment include facet joint violation during the placement of the cephalad-most screws [1], fusion above L5 [2], pre-existing facet degeneration [3], coexistence of laminar horizontalization and facet tropism [4], length of fusion [5], posterior interbody fusion rather than anterior interbody fusion [6], and performing an additional laminectomy adjacent to a fusion [7]. ASD at the lower adjacent segment may also occur due to failure of the most distal fixation, positive sagittal balance, younger age, or preoperative radiographic degeneration [8, 9].

It has been suspected that the stiffness of the instrumented levels relates directly to increased stress on the cranial adjacent segment and facet joints [10, 11]. The increased load over time leads to segmental hypermobility, facet joint hypertrophy, osteophyte formation, and spinal stenosis [12]. The stiffness of a titanium (Ti) rod is up to 20 times greater than bone and enough to eliminate motion, which provides intermediate stability until biological fusion is accomplished. However, a rigid rod may grossly alter the physiologic load transmission at the instrumented level.

Semi-rigid rods, such as AccuFlex (GlobusMedical), PEEK rods (Medtronic Sofamor Danek), Isobar (Scient'X), and more recently, carbon fiber reinforced PEEK rods (Depuy Spine), have been introduced as an alternative to rigid fusion devices. The primary concept of semi-rigid fixation is to achieve a careful balance between the amount of motion and loading of the spinal segment, but also to modify the physiology of the bone to achieve a solid fusion. Additionally, it is anticipated that semi-rigid rods will minimize the risk of ASD in the long-term. Recently, some biomechanical analyses have been carried out to compare the effectiveness of semi-rigid fixation devices. Using a posterior fixation device for dynamic, non-fusion stabilization was studied by Rohmann et al. [13] for its effects on the loads in the lumbar spine. Using a lumbar spinal column finite element model with a paired implant at the L3–L4 level, they found that the stiffness of a posterior implant has only a minor effect on intradiscal pressure (IDP) at the adjacent L2–L3 level despite an increase in intersegmental rotation (ISR) and facet joint forces for axial rotation and extension. A similar analysis was also carried out by

Galbusera et al. [14]. The effect of stiffness of different rods on the biomechanics of an L2–L5 FE model was studied with the application of a non-fusion stabilization system to the L4–L5. They suggested distinguishing “flexible” devices, which were able to preserve only a minor fraction of the physiological intersegmental range of motion (ROM), from “dynamic” devices, which induced a smaller ROM restriction. A biomechanical study by Turner et al. [15] using PEEK rods as a kind of fusion technology demonstrated that planar motion showed no differences between the groups instrumented with Ti and PEEK rods.

None of the previous studies have dealt with the effect of semi-rigid rods after mono-segmental fusion and posterior semi-rigid stabilization on the mechanical behavior of the lower and upper adjacent segments quantitatively. There is a need to compare the effects on each of the adjacent segments. The aims of the present study were to compare semi-rigid rods and rigid rods for their effects on the stability of the instrumented level and the accompanying effect on the adjacent segments under the conditions of L3–L4 floating fusion. An additional aim of the study was to evaluate the impact of the amount of fusion mass on the biomechanical differences between the two groups.

Materials and methods

A three-dimensional whole lumbar spinal column model was constructed as an extension of our validated one motion segment models [16–18]. Those one motion segment models consisted of L4 and L5 vertebrae composed of hexahedron elements. For practical purposes, the transverse cross-sectional shape of an intact specimen was assumed to be symmetrical about the mid-sagittal plane. The vertebral body was composed of two kinds of bones: cortical and cancellous ones. The cortical shell (Young's modulus $E = 12$ GPa; Poisson's ratio $\nu = 0.3$), which has a 1 mm thickness and the cancellous bone core ($E = 100$ MPa; $\nu = 0.3$) were modeled separately in order to be able to assign different material properties to each. All nodes between the cortical and cancellous bones were shared. To the posterior bony elements, the same material property values ($E = 3.5$ GPa, $\nu = 0.25$) were assigned because the amount of cancellous bone is difficult to estimate. The thickness of the endplate was 0.5 mm. The annulus was modeled as a composite of fiber layers superimposed on the ground substance and assumed to have four layers to match the corresponding bony elements. The average angle of the modeled fibers of the annulus, in a crisscross pattern, was $34^\circ \pm 7.4^\circ$ with respect to the horizontal plane. For the ground substance, a three-dimensional isotropic solid element was applied. The volume fraction for the fiber in the annulus was assumed to be

16 % [19]. The nucleus, which occupies 48 % of the disk cross-sectional area, was simulated with a three-dimensional incompressible fluid element. The reported tensile behavior of the multi-layer annulus indicated that the anterior annulus had larger values for tensile moduli than the posterior lateral annulus. Also, the outer regions of the annulus had greater moduli than the inner regions [20]. According to this observation, the material properties of the annulus fibers in our previous model were modified based on this observation. Twelve different groups of nonlinear material properties were assigned to the tension-only truss element in the annulus fiber model to incorporate those regional variations (Table 1). The simulated model of the facet joint articulation was a three-dimensional gap element model, which supports only compression in the direction normal to the surface. The seven spinal ligaments that connect the vertebrae were modeled as three-dimensional, tension-only truss elements. The elements were oriented along the direction of the corresponding ligaments. The same material properties adopted in our previous studies [16–18] were assigned to each component to complete the model. Responses in the modified motion segment model (i.e., force–displacement relation, nucleus pressure, facet contact force, and ligament strain) also matched well with our previous reports. With this modified one motion segment model, a mesh sensitivity analysis was performed and it was determined that an acceptable compromise between solution accuracy and solution time was achieved. (The nucleus pressure and axial displacement differences were only 3.3 and 5.7 % under 400 N of compressive load, and the angular rotation differences were 5.73 and 4.78 % under 6 Nm flexion and extension moment, respectively, when the mesh density was increased eightfold. The calculation time was increased 27 times despite this small change, and therefore less refined model was selected for this analysis.)

To develop the whole lumbar spine model, the shape of the vertebrae was modified on the basis of the human anatomy surface data [21, 22] to create that of the different levels. The same element types, modeling procedure, and material properties used in our one motion segment model were applied to the spinal column model generation. The pelvis and sacrum were modeled as a rigid body. The model that was developed shows an erect standing posture and the initial lordosis angle between the superior endplates of L2 and S1 was 43° (Fig. 1) [23]. The constructed model contained 26,364 nodes and 38,173 elements.

A rigid plate model was placed at the upper endplate of the L1 vertebral body to provide a loading frame and 200 mm of lever arm was additionally attached to the rigid plate. The mid-center of the lever arm was in sagittal plane, 2.5 mm cranial to the upper endplate of L1 and, for standing, 3.8 mm posterior to the center of rotation of the L1/L2 disk. The spinal column model was validated on the basis of the results from in vitro measurement [24, 25] and FE analysis [26]. Good agreement was found for a 10 Nm flexion–extension bending moment with the intersegmental ROM predicted by our FE model falling within the experimental standard deviation interval (Fig. 2).

The implanted models were constructed to simulate postoperative changes with two kinds of fixation systems: a conventional Ti rigid rod (circular cross-section, diameter = 6.0 mm; $E = 110$ GPa; $\nu = 0.3$) and a PEEK rod (ellipsoidal cross-section, major diameter = 7.0 mm, minor diameter = 6.2 mm, rod radius of curvature = 120.0 mm; $E = 3.6$ GPa; $\nu = 0.3$) with pedicle screws. Among the semi-rigid fixation devices, the PEEK rod was chosen for this study because of its unique design characteristics, in which its stiffness in extension is larger than in flexion. The L3–L4 motion segment was selected for application of the instrumentation system. The implant was fixed to the vertebrae by two transpedicular screws on each

Table 1 Material properties applied to different annular layers

Annulus fiber	Young's modulus (MPa)
Innermost anterior layer	34 ($\epsilon < 0.031$), 28 ($\epsilon > 0.031$)
Second layer of the anterior section	42.67 ($\epsilon < 0.04$), 75.33 ($\epsilon > 0.04$)
Third layer of the anterior section	51.33 ($\epsilon < 0.03$), 122.67 ($0.03 < \epsilon < 0.049$), 422.67 ($\epsilon > 0.049$)
Outermost anterior section	60 ($\epsilon < 0.037$), 170 ($0.037 < \epsilon < 0.058$), 620 ($\epsilon > 0.058$)
Innermost lateral section	17.25 ($\epsilon < 0.039$), 14.05 ($\epsilon > 0.039$)
Second layer of the lateral section	33.17 ($\epsilon < 0.05$), 61.03 ($\epsilon > 0.05$)
Third layer of the lateral section	49.08 ($\epsilon < 0.035$), 108.02 ($0.035 < \epsilon < 0.06$), 374.68 ($\epsilon > 0.06$)
Outermost lateral section	65 ($\epsilon < 0.044$), 155 ($0.044 < \epsilon < 0.071$), 555 ($\epsilon > 0.071$)
Innermost posterior section	0.5 ($\epsilon < 0.047$), 0.1 ($\epsilon > 0.047$)
Second layer of the posterior section	23.67 ($\epsilon < 0.059$), 46.73 ($\epsilon > 0.059$)
Third layer of the posterior section	46.83 ($\epsilon < 0.041$), 93.37 ($0.041 < \epsilon < 0.072$), 326.70 ($\epsilon > 0.072$)
Outermost posterior section	70 ($\epsilon < 0.05$), 140 ($0.05 < \epsilon < 0.084$), 490 ($\epsilon > 0.084$)

Fig. 1 The detailed lumbar spinal column model developed in the present study. View of segments with vertebral bodies, disks, and ligaments. Part of annulus at the anterior region is removed to show its layers

side. The screw section in the pedicle screw was simplified as a beam element that had the same bending stiffness as the actual screw, and the screw head section was reconstructed through FE modeling. Since our models were designed to simulate the biomechanical behavior of long-term status after instrumentation, the bone–screw interface was assumed to be completely bonded via the node sharing. The connection between the rods and screws was simulated to be firmly connected. The same material property of Ti was assigned to the screw head section. For the simulation of interbody fusion, interbody spacer (IS) models ($E = 6$ GPa), which were placed parallel to the mid-sagittal plane maintaining the initial L3–L4 disk angle, could be generated by replacing parts of the nucleus elements. The nodes between the endplate and interbody spacer model were shared to simulate complete interbody union. This IS model occupied 38 % of the nucleus volume. The largest interbody spacer model (IS_all) was also designed with the whole nucleus replaced by the interbody spacer to identify the effect of the fusion mass. The possible disk degeneration at the instrumented level and the alteration of the annulus during the surgical procedure were not considered because of the large stiffness difference between the implants and intervertebral disk. The four formulated models, that is, the intact, instrumented with rigid fixation and an interbody spacer (Ti + IS), instrumented with semi-rigid fixation and an interbody spacer (PEEK + IS), and instrumented with semi-rigid fixation only (PEEK) models (Fig. 3), were compared under each set of sagittal plane loading conditions.

During the trunk bending motion, the arc motion will appear with straightening (in flexion) or bending (in extension) motion in the lumbar spinal column. This

Fig. 2 FE models of **a** intact, **b** instrumented with PEEK semi-rigid rod, and **c** Ti rigid rod systems. Parts of the bony elements in the instrumented segment (in PEEK + IS) were removed for a clear view of the pedicle screw model. IS in the intervertebral disk model indicates the interbody spacer. Loading frames and follower load paths are shown in the intact model

Fig. 3 Comparison between the relative rotation angle under a 10 Nm flexion–extension moment predicted by FE studies and in vitro measurements

motion simulation was conducted through the following loading procedures.

Step 1: The desired position of vertebra L1 during the arc motion of the lumbar spinal column (up to 30° for flexed posture and -20° for extended posture at 10° intervals) was first calculated. To calculate its position, the imaginary line, which connected the center of the L1 vertebral body and the center of the upper endplate of S1, was assumed. The desired position of vertebra L1 of the lumbar spinal column was calculated by rotating this imaginary line with respect to the center of the upper endplate of S1.

Step 2: Rotation of the spinal column was performed by moving the rigid plate placed at the upper endplate of the L1 vertebral body to the predetermined position in step 1 by displacement loading condition. A fixed boundary condition was applied to the sacral body.

Step 3: 400 N of upper body weight (UBW) was imposed in the gravitational direction onto the rigid plate along with a compressive follower load (FL).

Step 4: A proper bending moment was additionally applied so as to sustain the position of the L1 vertebral body center on the rotated imaginary line developed in step 1. The bending moment was generated through the force imposed on the lever arms (each 100 mm in length) used in the verification procedure.

During the loading procedure in steps 2, 3, and 4, motion of every vertebra was not fixed and therefore the forces and moments were balanced within the spinal column. For the erect standing simulation, only steps 3 and 4 were required.

In order to impose the compressive FL, connector elements were applied between each of the vertebral body centers and then local coordinates were assigned to each connector element, which were to provide the acting

direction of the FL in accordance with the changed curvature of the spinal column. For the proper selection of the FL, the calculated result of the L4–L5 nucleus pressure of the intact model was compared with reported in vivo measurements [27, 28]. The optimal value of the FL (100 N) could be derived by trial and error so as to generate a nucleus pressure close to that of the experimental data (Fig. 4). The magnitude of the FL was maintained regardless of lumbar motion. The angle difference between the vertebra L1 angle after step 4 and that in erect standing was processed as the final flexion or extension angle of the spinal column.

ABAQUS (ver. 6.10, Hibbitt Karlsson & Sorresen Inc., Pawckur, RI, USA) was used to carry out a nonlinear structural analysis on the detailed lumbar spinal column model.

Results

The angle of vertebra L1 with respect to the horizontal plane in the unloaded model was -3° (extended) and the resulting angles of the four different models showed slightly more extended postures during erect standing. Instrumentation at L3–L4 level affected the disk angles of the other levels and accordingly, the angle of vertebra L1 of the Ti + IS model revealed a 0.18° difference compared with that of the intact model, and -0.03° and -0.21° differences were calculated in the PEEK + IS and PEEK models, respectively. To maintain the L1 vertebra center in loading step 4, 7.09, 8.92, 9.26, and 10.0 Nm of extension moment were required for the intact, Ti + IS, PEEK + IS, and PEEK models, respectively.

During the sagittal plane motion, the 30° flexional arc motion in loading step 2 eventually became a 50° flexion

Fig. 4 Nucleus pressure at the L4–L5 level predicted in the present study compared to in vivo measurement [27]

motion after loading step 4 because of the accompanying straightening motion in the spinal column. In this case, the intact model required 2.59 Nm of extension moment to accomplish loading step 4. In contrast to the intact case, 7.70, 2.36, and 1.57 Nm of flexion moment was required for the Ti + IS, PEEK + IS, and PEEK models, respectively. For the 26° extended posture, 16.87, 32.32, 26.50, and 26.0 Nm of extension moment were needed for the intact, Ti + IS, PEEK + IS, and PEEK models, respectively. The final angles of vertebra L1 for the different postures are summarized in Table 2.

The motion changes of the adjacent segments were maintained to almost the same degree for the Ti and PEEK models regardless of the amount of fusion mass. The largest relative intervertebral disk angle change between the IS and IS_all was generated at the instrumented level with PEEK rods during 50° flexion, and only a difference of less than 3 % was detected at the other levels (Fig. 5). Therefore, the effects of the instrumentation devices were analyzed only for the IS case.

The ISRs for the flexed/extended posture were calculated from the erect standing posture of each model and

then the ISR difference at each level was further derived with respect to those of the intact model. The results indicated that the largest change was generated at the instrumented level as expected, but the ISR difference between the Ti + IS and PEEK + IS was not noticeable (less than 0.14° during 50° flexion) at this level (Fig. 6). Meaningful ISR differences were produced at the upper segments (L1–L2 and L2–L3). The PEEK + IS demonstrated a smaller ISR increment than did the Ti + IS. The increment of the ISR at the upper adjacent level (L2–L3) was more than double in the case of the Ti + IS than that in the PEEK + IS. In contrast to the upper level, the PEEK + IS showed slightly increased ISRs at the lower level (L4–L5 and L5–S1), even though these differences were not significant.

Flexion/extension motion increased the IDP at all levels except the instrumented level, while noticeable differences were not generated with erect posture (Fig. 7). At the upper segment (L1–L2 and L2–L3), a large pressure increment was calculated in the instrumented models. The difference between the intact and Ti + IS models in flexion motion increased with the magnitude of motion. Under a 50°

Table 2 Final angles of vertebra L1 with respect to the horizontal plane for different postures

Arc motion ^a (step 2)	FLX/EXT motion ^b	Intact	PEEK	PEEK + IS	Ti + IS
20° extensional	EXT26	−35.09	−35.32	−35.26	−36.53
10° extensional	EXT12	−20.90	−21.12	−21.19	−22.11
0°	ERECT	−8.90	−9.12	−8.94	−8.72
10° flexional	FLX14	5.14	4.64	5.19	7.69
20° flexional	FLX30	22.91	21.14	22.63	26.96
30° flexional	FLX50	40.98	40.82	40.57	42.90

Positive values indicate flexed angle and negative values indicate extended angle from the horizontal plane

^a Rotation of spinal column to the predetermined position in step 1

^b The angle difference between the final angle of vertebra L1 after step 4 and that in erect standing. FLX50 indicates 50° flexion motion

Fig. 5 Relative motion change according to the fusion mass in rigid and semi-rigid fixation system

Fig. 6 Intersegmental disk angle changes with respect to the intact model for different fixation system

flexed posture, the pressure in the intact L2–L3 level was 776.8 kPa, while it was 1081.9, 909.0, and 879.0 kPa in the Ti + IS, PEEK + IS, and PEEK models, respectively. The

uppermost level (L1–L2) also generated a similar difference. In contrast to the results of the upper adjacent segment, the IDP at the lower adjacent segment (L4–L5) in

Fig. 7 Computed intradiscal pressure values at each level under different postures and fixation system

PEEK + IS was larger than that of Ti + IS by 2.0 %. On the other hand, extension motion produced the opposite changes (2.5 % decreased under 26° extended posture). At the L5–S1 level, flexion/extension motion showed similar changes to those of the L4–L5 level although relatively higher pressure was calculated.

Instrumentation also increased the facet joint contact force at all levels except the instrumented level, and the largest increment was detected at the upper adjacent level (L2–L3) in the Ti + IS during 26° extension (Fig. 8). At the lower adjacent level (L4–L5), instrumentation itself generated a 55 % increased facet contact force compared to the intact model.

Von Mises stress distributions in the mid-plane of the intervertebral disk at the adjacent levels were predicted for 50° flexion and 26° extension (Fig. 9). Instrumentation with Ti + IS markedly increased the stress in the intervertebral disk at the upper adjacent level, while relatively small stress increments were found with PEEK + IS and PEEK. Similar to the IDP change, PEEK + IS generated slightly increased stress distribution at the lower adjacent level.

Discussion

Adjacent segment degeneration after interbody fusion is one of the most important sequelae affecting long-term results. To understand the motion increase of the disk adjacent to the interbody fusion in a semi-rigid fixation system compared to a rigid system, a whole lumbar spinal column FE model was established and biomechanical analysis was carried out with this model.

Proper loading conditions are essential for the accurate outcome of FE analysis. The UBW is actually located anterior at the L1 vertebra in the upright standing posture and it moves further in the anterior/posterior direction depends on flexion/extension motion, but in this study the UBW was applied onto the rigid plate placed on the L1 vertebral body for all loading cases. Due to this loading condition, an additional flexion moment was required for the instrumented models as listed in the results. Even though the loading condition used in this analysis did not exactly match with that of in vivo, the results of disk pressure were in good agreement with in vivo measurement as shown in Fig. 4. The concept of FL was effectively

Fig. 8 Facet contact force during erect standing and 26° extended posture

applied to in vitro experiments and the path of FL was known to be an important factor to avoid the induced bending by non-optimized compressive FL [29]. According to Rohlmann et al. [30], applying 500 N of FL acting in the center of each vertebral body and directed to the centers of the adjacent vertebral bodies delivered the most probable ISRs among six different loading modes for simulating standing. Our model also produced similar ISRs by applying 500 N of FL only instead of a loading method used in this study, but small magnitudes of nucleus pressure were calculated especially in the lower levels (399.7, 400.5, and 452.7 MPa at the L3–L4, L4–L5, and L5–S1 level, respectively). Moreover, this loading method produced a posterior movement at vertebra L1. The recently proposed hybrid method [31] or hybrid method together with FL [32] also produced anterior/posterior movement at the ending vertebra (vertebra L1). It seems more reasonable that anterior/posterior movement of vertebra L1 should be restricted to consider the conjunction of vertebra L1 with the upper elements (vertebra T12 and the trunk) in the in vivo state. However, restriction in anterior/posterior movement generated different angles at the ending vertebra for instrumented models although the difference was small. A stiffness change at the instrumented segment may cause such a difference.

The effect of the fusion mass (IS vs. IS_all) on the ISR at the instrumented level as seen in Fig. 5 showed a maximum of 32.25 %; however, its angle change was only 0.6°. The motion difference based on the fusion mass at the adjacent levels was at most 3 %. This finding suggests that the stiffness difference between the Ti and PEEK rods in

Fig. 9 Predicted von Mises stress contour in the center plane of the intervertebral disk

the ISR at the adjacent motion segments was more crucial than the amount of fusion mass.

The effect of the Ti and PEEK rods in reducing the ISR was found to be critical in the instrumented level (71.4 % in PEEK with respect to the motion value obtained from the model of the intact spine, 81.2 % in PEEK + IS, and 85.4 % in Ti + IS during 50° flexion). These findings are similar to the results of a cadaveric study (80 % in a PEEK rod only model and 85 % in a Ti rod only model) [33] and another simulation (72 % motion decrement in a PEEK rod model and 80 % in a Ti rod model with an intact disk) [14].

Subsequently, determining whether a PEEK rod actually has a beneficial effect on the adjacent segment compared with a rigid rod is important. The data obtained from this detailed biomechanical study would be helpful for the selection of the optimal instrument and prediction of long-term postoperative complications. The ISR change in the instrumented level induced the motional change in the adjacent segments. A noticeable difference between the Ti and PEEK rod was produced at the upper adjacent segment. Compared to the intact model, Ti + IS and PEEK + IS increased the ISR by 48.9 and 26.4 %, respectively, at the upper adjacent segment and the same degree of increment was also found at the uppermost level (L1–L2). A similar change was also detected in the facet joint contact force. Under 26° extension, 50.2 and 32.3 % increments were generated in Ti + IS and PEEK + IS, respectively, while the difference between the two models was not noticeable at the L4–L5 level. These findings indicated that PEEK + IS may be advantageous to the upper adjacent segments compared with Ti + IS. However, the effect of PEEK + IS on the lower adjacent segments was slightly negative. That would be due to the compensatory effect of motion. Compared with Ti + IS, the decrease in the rotational motion in the upper level should be compensated for at the lower level for the generation of a similar amount of flexion motion. The increase in the motion produced 25 kPa higher IDP at the L4–L5 level under 50° flexion. This finding was also confirmed by the stress distribution as shown in Fig. 9. Nevertheless, changes in the lower adjacent segment were much smaller than those in the upper adjacent segment.

Some of the clinical follow-up studies support our results. Radiographic findings suggested that fusion imposed additional stress on the motion segment above the fusion site [34] and the incidence of distal ASD was much lower than that of proximal ASD in floating fusion [35]. Most of the previous biomechanical studies using in vitro and FE analysis could not interpret this level difference. In the cadaveric study of Turner et al. [14], the IDPs in the erect position and during flexion at the lower adjacent segment were in disagreement with their results. They reported a higher IDP during flexion and a lower pressure

during compression in the Ti + IS than in the PEEK + IS. The discrepancy is likely to be attributed to several factors such as short segment specimens, different loading methods, the effect of lordosis, and deep muscle activation simulated as a follower load in our study, which was not considered in their in vitro study.

Even though PEEK rods were shown to be protective of facet joint degeneration in the upper adjacent level compared with Ti rods, the lower adjacent facet joints of all instrumented models produced 55 % increased contact force over that of the intact model. This finding implies that PEEK rods may not be appropriate for a protection of facet joint degeneration. The stress distribution in the disk at the lower adjacent level also implies that the possibility of disk degeneration is higher in the PEEK + IS than the Ti + IS. All these findings indicate that PEEK rods may not be as protective from facet joint degeneration and disk degeneration at the lower adjacent segment.

This computational study has some limitations. Several assumptions and simplifications were necessary for creating the finite element model. Solid interbody fusion and bone–screw fusion were assumed in the instrumented level and therefore relative motion between the bone–screw and bone interface in the early stage of instrumentation could not be simulated. Surgical defects in the posterior element such as laminectomy, posterior annulectomy, and medial facetectomy were not incorporated in the simulation. During the modeling, the vertebra L1 model was located somewhat anteriorly compared to the averaged neutral erect posture and the pelvic tilting was not considered during flexion/extension motion. As the proposed loading method attempts to simulate the long-term in vivo spinal behavior of a patient, presently it is not known if an angle difference at vertebra L1 will be produced in in vivo behavior. Clinical studies documenting the long-term changes in intervertebral and total motion due to the adaptive response to surgery are still lacking. Further clinical observation is needed in order to clarify this difference. The effect of the paraspinal deep muscles was treated as a FL, but the major lumbar extensor muscles provided evidence documenting the compromise in low back shear force support [36]. This finding indicated that FL is not enough for the simulation of realistic paraspinal deep muscle activation in vivo. A more sophisticated loading method will be further required for a more accurate outcome. These deficiencies could offer a range of possibilities for exaggerated ISR and IDP in the lower lumbar spine; thus, we should be cautious about interpretation and our findings should be taken as trends rather than absolute values.

In conclusion, mono-segmental fusion and posterior instrumentation with PEEK rods may lower the incidence of disk and facet degeneration in the upper adjacent

segment compared to Ti rods. On the other hand, the possibility of disk and facet degeneration will be similar in the lower adjacent segment in the long-term. All these changes at the adjacent segments may be derived from the change of load transmission by the difference in rod stiffness. In addition, simulation in other settings such as with a different number of fusion levels, instrumentation levels, and degree of lumbar lordosis would be necessary to compare the impact of those parameters on the biomechanical changes of adjacent segments.

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Conflict of interest None.

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