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ORIGINAL ARTICLE

Effect of the cord pretension of the Dynesys dynamic stabilisation system on the biomechanics of the lumbar spine: a finite element analysis

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Abstract The Dynesys dynamics stabilisation system was developed to maintain the mobility of motion segment of the lumbar spine in order to reduce the incidence of negative effects at the adjacent segments. However, the magnitude of cord pretension may change the stiffness of the Dynesys system and result in a diverse clinical outcome, and the effects of Dynesys cord pretension remain unclear. Displacement-controlled finite element analysis was used to evaluate the biomechanical behaviour of the lumbar spine after insertion of Dynesys with three different cord pretensions. For the implanted level, increasing the cord pretension from 100 to 300 N resulted in an increase in flexion stiffness from 19.0 to 64.5 Nm/deg, a marked increase in facet contact force (FCF) of 35% in extension and 32% in torsion, a 40% increase of the annulus stress in torsion, and an increase in the high-stress region of the pedicle screw in flexion and lateral bending. For the adjacent levels, varying the cord pretension from 100 to 300 N only had a minor influence on range of motion (ROM), FCF, and annulus stress, with changes of 6, 12, and 9%, respectively. This study found that alteration of cord pretension affects the ROM and FCF, and annulus stress within the construct but not the adjacent segment. In addition, use of a 300 N cord pretension causes a much higher stiffness at the implanted level when compared with the intact lumbar spine.

Keywords Lumbar spine · Biomechanics · Dynesys dynamic stabilisation system · Finite element method · Cord pretension

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Introduction

In the recent years, the approach to the design of spinal implants has changed from stable fusion to mobile non-fusion that attempts to reduce the incidence of the adjacent segment degeneration (ASD) associated with rigid fusion [1]. Non-fusion technology can be classified into several types: pedicle-based dynamic stabilisation systems, artificial discs, nucleus prostheses, total facet arthroplasty systems, and interspinous process spacers [1, 2]. The Dynesys (Zimmer GmbH, Winterthur, Switzerland) is a pedicle-based dynamic stabilisation system that consists of a polyethylene terephthalate (PET) cord, a polycarbonate urethane (PCU) spacer, and conical titanium alloy pedicle screws. The PET cord resists tensile forces to limit flexion, and the PCU spacer sustains compressive forces to prevent excessive extension. Clinical reports have indicated that Dynesys provides a safe and effective alternative in the treatment of unstable lumbar diseases, with a reported satisfaction rate ranging from 60 to 90% [3–7]. However, complications including ASD, screw loosening, and screw breakage have occurred in the postoperative period [4–7].

The biomechanical characteristics of Dynesys have been evaluated by several laboratories using in vitro experimental tests or finite element (FE) analysis. Freudiger et al. [8] indicated that Dynesys increases stability under flexion, extension, and shear load, but reduces stability under compressive load when compared to the intact lumbar spine. The device also reduced the bulging of the posterior annulus. Schmoelz et al. [9] found that the Dynesys provided stability for the unstable segment, but was more flexible than was rigid fixation system. No differences were found in motion and intradiscal pressure at the adjacent segments between the Dynesys and rigid fixation systems [9, 10]. In 2007, Rohlmann et al. [11] indicated that, other than after distraction, the mechanical effects of the Dynesys are similar to those of a rigid fixation system. Zander et al. [12] evaluated the manner in which a Dynesys adjacent to a rigid fixation system affects the mechanical behaviour of the lumbar spine. They found that whether the Dynesys was positioned superior or inferior to the rigid fixation system had only a minor influence on biomechanical results. However, the implant forces strongly depend on the stiffness of the Dynesys. Niosi et al. [13, 14] investigated the effects of the spacer length of the Dynesys. The results of this study indicated that a Dynesys with a long spacer typically caused an increase in range of motion (ROM) and a decrease in facet loads compared with those with a short spacer. Our earlier study evaluated whether various depths of Dynesys screw placement would affect the biomechanical characteristics of the lumbar spine. We demonstrated that the profile of screw placement only had a minor influence on the ROM, annulus stress, and facet joint force, but that the screw stress was noticeably increased as the screw was moved further out of the pedicle [15].

The Dynesys implantation guide recommends that a 300 N preload should be applied on the PET cords to distract the disc during the implantation procedure. From a biomechanical point of view, different magnitudes of cord pretension may change the stiffness of the Dynesys system and result in diverse clinical outcomes. However, the effects of Dynesys cord pretension have not yet been studied in detail. Therefore, the purpose of this study was to investigate the influence of Dynesys cord pretension on the ROM, facet contact force (FCF), annulus stress, and screw stress distribution.

Materials and methods

FE model of the intact lumbar spine (INT model)

A three-dimensional L1–L5 intact lumbar spine (INT) FE model was built using the FE analysis software ANSYS 11.0 (ANSYS Inc., Canonsburg, PA, USA). The INT model included the vertebrae, intervertebral discs, endplates, posterior bony elements, and all seven ligaments (anterior and posterior longitudinal ligaments, flaval ligament, facet capsules, intertransverse, interspinous, and supraspinous ligaments). The 8-node solid element was used to model the cortical bone, cancellous bone, endplate, posterior bony element, and annulus ground substance. For the disc, 12 double cross-linked fibre layers were embedded in the ground substance, and fibre stiffness was increased proportionally from the outermost layer to the innermost layer [16]. The 43% of the cross-sectional area in the disc was defined as the nucleus, within the range reported by Panagiotacopulos [17] (30–50%). The nucleus pulposus was modelled as an incompressible fluid by an 8-node fluid element. The nonlinear annulus ground substance was simulated using a hyper-elastic Mooney-Rivlin formulation. All seven ligaments were arranged according to their anatomic direction and represented by the 2-node tension-only link element. The facet joint was treated as having a sliding contact behaviour using three-dimensional 8-node surface-to-surface contact elements (CONTA174), which were allowed to slide between three-dimensional target elements (TARGE170). The coefficient of friction was set at 0.1. The initial gap between a pair of facet surfaces was kept within 0.5 mm. The stiffness of the spinal structure changes depending on the contact status, so the standard contact option in ANSYS was adopted to account for the changing-state nonlinear problem in this study. A more detailed description of this intact lumbar spine FE model has been presented in our earlier studies [18, 19]. The FE model of the intact lumbar spine consisted of 112,174 elements and 94,162 nodes.

Convergence test and model validation

For the convergence test, three mesh densities (coarse model: 4,750 elements/4,960 nodes; normal model: 27,244 elements/30,630 nodes; finest model: 112,174 elements/94,162 nodes) were selected to test the ROM changes in the INT model, and the finest mesh density was selected because the changes between the normal model and finest model were within 1.03% in flexion (<0.2°), 4.39% in extension (<0.5°), 0.01% in torsion (<0.2°), and 0.001% in lateral bending (<0.1°), respectively [20].

For the model validation, ROM and FCF were chosen for comparison with the previous literature. First, ROMs in five levels of the INT model under different moments of 3.75, 7.5, and 10 Nm were validated with the results of the previous study [18]. Good agreement of ROMs was achieved under most of the loading cases. Under a 10-Nm moment with a 150-N preload, the current INT model showed some stiffer behaviour in flexion when compared with mean value of in vitro tests. Second, FCF in torsion of each level was compared with Chen's [19] and Shirazi-Adl's [16] FE studies. A similar trend was achieved and demonstrated in our previous study. Thus, this INT model could be used for further simulation analysis.

FE model of the Dynesys dynamic stabilisation system

This study modelled bilateral insertion of the Dynesys system into the L3/L4 segment of the INT model according to standard surgical procedures (Fig. 1a). A Dynesys system includes two conical screws (diameter = 6.4 mm, length = 45 mm), a PCU spacer (diameter = 12 mm, length = 30 mm), and a PET cord as shown in Fig. 1b. The conical titanium alloy screws and the PCU spacer were modelled using an 8-node solid element. According to the implantation guide, the conical titanium alloy screws were inserted through a medialised screw trajectory that can preserve the facet joint as shown in Fig. 1c. Both sides of the PET cord were modelled as a connection with screw heads at L3 and L4 using the 2-node tension-only link element. The Young's modulus and Poisson's ratio of all the Dynesys components were, respectively, 110 GPa and 0.3 for the titanium alloy screw, 68.4 MPa and 0.4 for the PCU spacer, and 1,500 MPa and 0.4 for the PET cord.

In order to investigate the effects of cord pretension, loads of 100, 200, and 300 N were bilaterally applied on the PET cords and defined as the PT100, PT200, and PT300 models, respectively.

Loading and boundary conditions

The loading condition of a constant ROM has been shown to be more clinically relevant in predicting the adjacent level effects following the insertion of a spinal implant [18, 20]; for this reason, this study used the displacement control method. For all Dynesys cases, three load steps were included in the whole simulation process. The first load step included three different magnitudes (100, 200, and 300 N) of cord pretension, which were bilaterally applied on the PET cords by controlling the initial strain of the link element. The second load step consisted of application of a 150-N compressive preload perpendicular to the superior surface of the L1 vertebra. In the third load step, a higher pure moment was applied incrementally to ensure that the

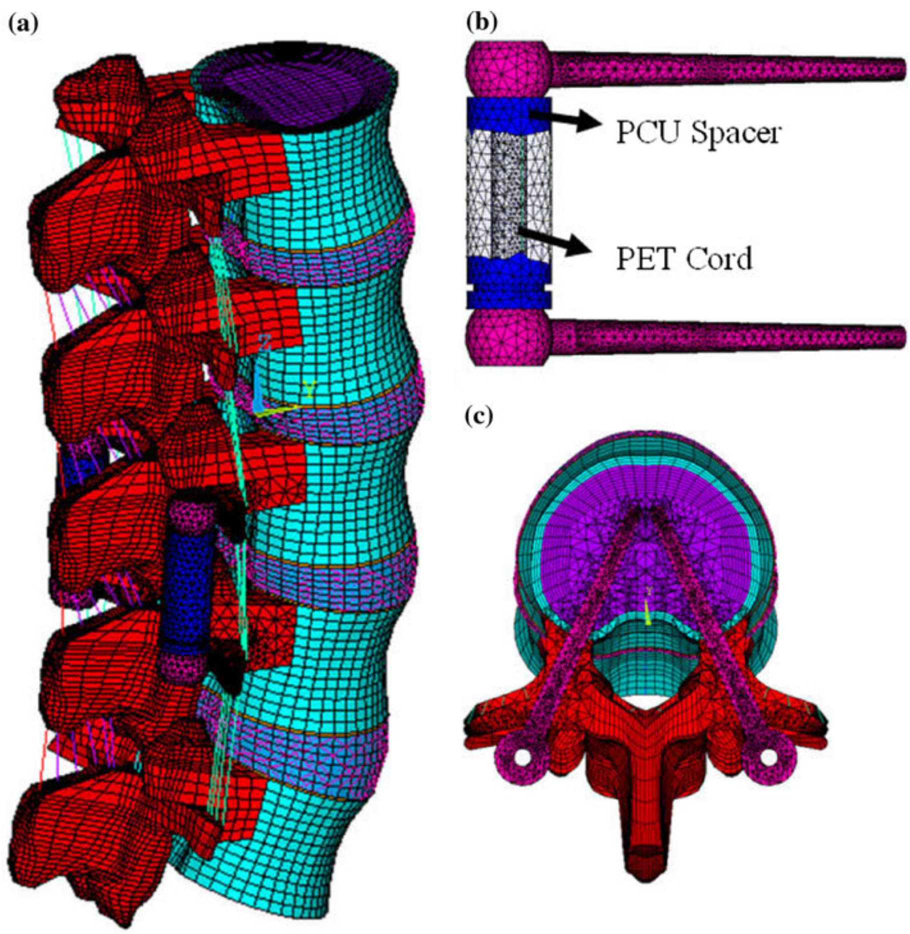


Fig. 1 Finite element model of the lumbar spine paired with a Dynesys system. **a** The Dynesys was implanted into the standard position within L3/4. **b** The Dynesys system consists of a polyethylene terephthalate (PET) cord, a polycarbonate urethane (PCU) spacer, and conical titanium alloy pedicle screws. **c** The Dynesys screws were inserted through a medialised screw trajectory

resultant ROMs (L1–L5) of all FE models would match the values of 20° in flexion, 15° in extension, 8° in torsion, and 20° in lateral bending under various pure moments. The load corresponding to the ROM was 16 Nm in flexion, 14.9 Nm in extension, 9.2 Nm in torsion, and 14.9 Nm in lateral bending for the intact lumbar spine model. The resulting difference of ROMs among these FE models was controlled within 0.09° in flexion, 0.07° in extension, 0.07° in torsion, and 0.11° in lateral bending (Table 1). All the degrees of freedom were constrained at the bottom of the fifth vertebra for all FE models.

The results included ROM, FCF, annulus stress, and the stress distribution of pedicle screws on L3. All the data were normalised to the INT model as percentage values under each loading mode.

Results

The effects of cord pretension

Without any external applied moment, the lordosis angle and FCF of the L3–L4 motion segment were increased following with increasing cord pretension listed in Table 2. The lordosis angle in the PT300 model maximally increased by 14.6% when compared with the INT model. The contact force of the facet joint in the PT300 model also increased to 33 N. However, the alteration of spacer length with various cord pretensions remained within 1%.

Range of motion and stiffness at the implanted level

The ROM at the implanted level was reduced dramatically in flexion (at least 78.4%), extension (at least 28.6%), and lateral bending (at least 53.9%), but less so in torsion (at least 12.8%) when compared with the INT model. Increasing the cord pretension from 100 to 300 N resulted in a further ROM decrease by 15% in flexion and ROM increase by 17% in extension (Fig. 2). Increasing the cord pretension resulted in a marked increase in stiffness under flexion and a slight reduction in stiffness under extension for the implanted level. These results are listed in Table 2. There were no prominent differences in ROM and stiffness between different cord pretensions under torsion and lateral bending.

Range of motion at the adjacent levels

The ROM at both adjacent levels increased obviously in flexion (by at least 22.4%) and lateral bending (by at least 15.6%), but less so in extension (by at least 7.2%) and torsion (by at least 4.6%) when compared with the INT

Table 1 Intervertebral range of motion among the INT, PT100, PT200, and PT300 models

Motion	Model	ROM (deg)				Five-level total lumbar ROM (deg)	Stiffness at the implanted level (Nm/deg)
		L1–L2	L2–L3	L3–L4	L4–L5		
Flexion							
	INT	4.55	4.67	4.72	6.15	20.09	3.390
	PT100	5.54	5.84	1.02	7.53	19.93	19.004
	PT200	5.67	5.99	0.67	7.70	20.03	31.194
	PT300	5.75	6.08	0.33	7.81	19.97	64.545
Extension							
	INT	3.56	3.48	3.39	4.62	15.05	4.395
	PT100	4.03	3.90	1.85	5.22	15.00	9.622
	PT200	3.97	3.83	2.13	5.14	15.07	8.169
	PT300	3.86	3.73	2.42	5.00	15.01	6.983
Torsion							
	INT	1.67	1.79	2.11	2.38	7.95	4.360
	PT100	1.79	1.91	1.84	2.49	8.03	5.598
	PT200	1.80	1.93	1.78	2.52	8.03	5.787
	PT300	1.78	1.91	1.73	2.51	7.93	5.954
Lateral bending							
	INT	4.84	4.74	4.90	5.60	20.08	3.041
	PT100	5.63	5.48	2.26	6.61	19.98	7.699
	PT200	5.63	5.50	2.15	6.62	19.90	8.093
	PT300	5.63	5.51	2.14	6.61	19.89	8.131

Table 2 Comparison of lordosis angle, spacer length and FCF of the L3–L4 motion segment

	INT	PT100	PT200	PT300
Lordosis angle (deg)	8.00	8.44	8.82	9.17
Facet contact force (N)	0	1.46	15.42	33.26
Spacer length (mm)	–	29.59	29.21	28.84

The original spacer length in Dynesys system was 30 mm

model. When cord pretension was increased, ROM was slightly increased in flexion and slightly decreased in extension, but the differences between these Dynesys models were still within 6% for all motions. Both adjacent levels exhibited similar trends (Fig. 2).

Facet contact force at the implanted level

Insertion of Dynesys resulted in an obvious decrease in FCF at the implanted level in extension (at least 49.9%), but not in torsion (between −9.1 and +22.6%) when compared with the INT model. When the cord pretension was increased from 100 to 300 N, the FCF of the PT300 model was markedly increased by 35% in extension and 32% in torsion when compared with that of the PT100 model (Fig. 3).

Facet contact force at the adjacent levels

The FCFs at both adjacent levels were increased in extension (by at least 17.2%) and torsion (by at least 11.1%) when compared with the INT model. When the cord pretension was increased from 100 to 300 N, the FCF at the L2/3 level was decreased by 12% in extension, whereas no prominent differences were found in torsion. Both adjacent levels exhibited similar trends (Fig. 3).

Maximum von Mises stress in the disc annulus and pedicle screw at the implanted level

The annulus stress at the implanted level was strongly reduced with respect to that of the INT model in flexion (at least 39.3%), extension (at least 14.1%), and lateral bending (at least 48.4%), but not in torsion (between −2.8 and +37.4%). When the cord pretension was increased from 100 to 300 N, annulus stress increased by 40% in torsion, while the differences were within 7% for flexion, extension, and lateral bending (Fig. 4).

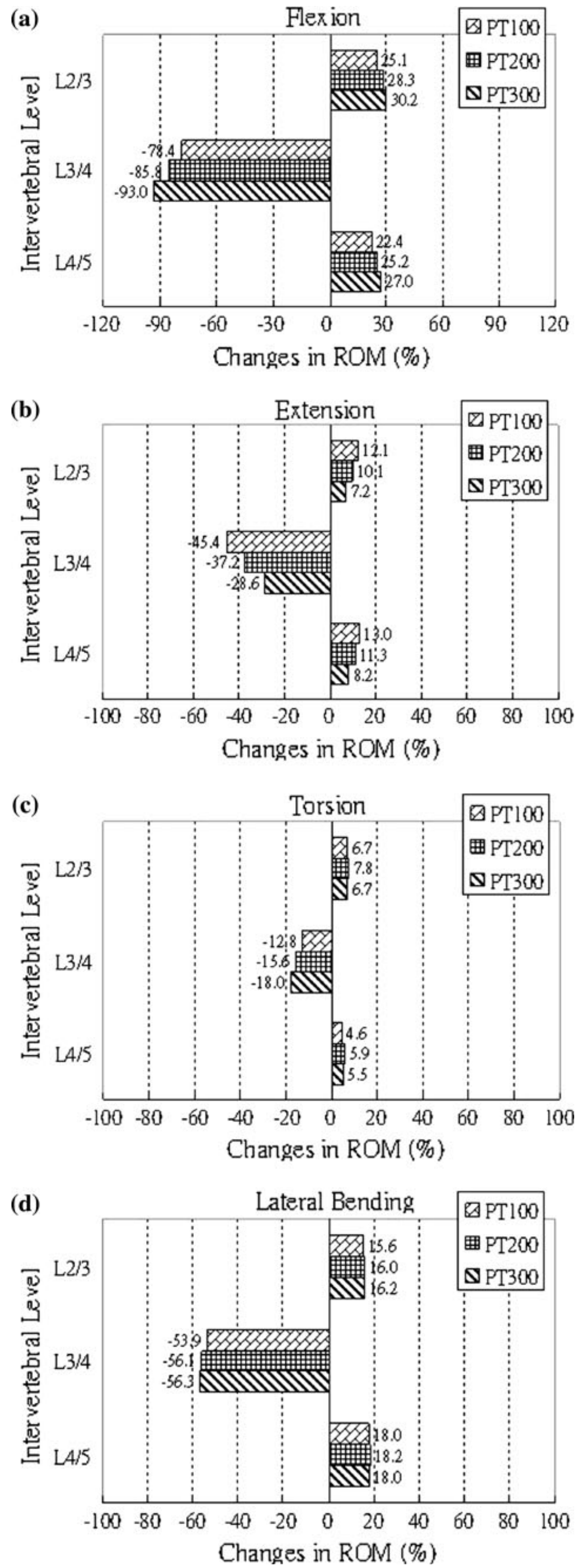


Fig. 2 Changes in the ROM (% of intact) at the implanted and adjacent levels under **a** flexion, **b** extension, **c** torsion, and **d** lateral bending ▶

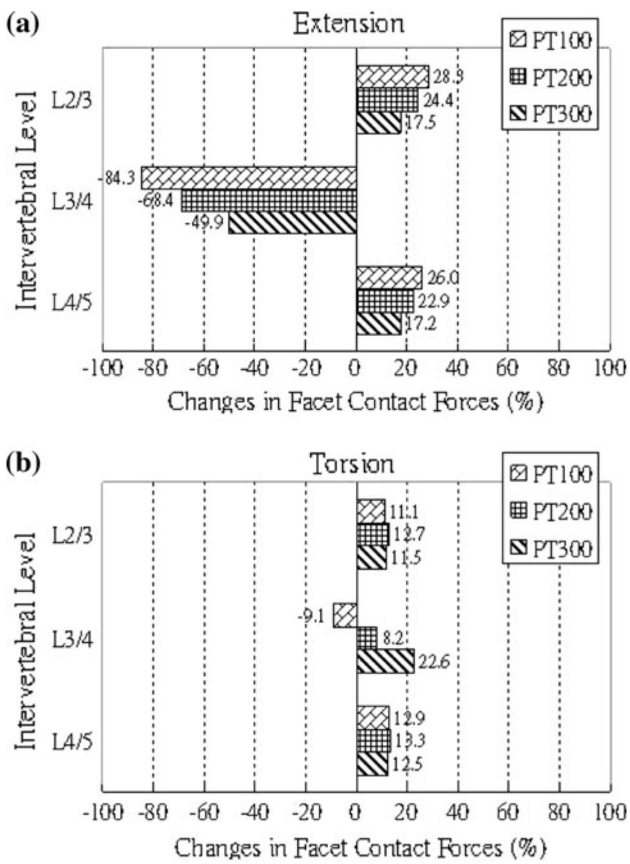


Fig. 3 Changes in the FCF (% of intact) at the implanted and adjacent levels under **a** extension and **b** torsion

The maximum screw stress occurred in the PT300 model in lateral bending and reached 122 MPa. The maximum screw stress in the PT100 model was lower than that in the PT300 model, as shown in Table 3. The high-stress region of pedicle screws increased gradually as the cord pretension was increased from 100 to 300 N, particularly in flexion and rotation. These high-stress regions are located in the pedicle region.

Maximum von Mises stress in the disc annulus at the adjacent levels

The maximum von Mises stress was calculated at ground substance of disc. The annulus stress at both adjacent levels increased obviously in flexion (at least 28.8%) and lateral bending (at least 16.3%), but less so in extension (at least 5.8%) and torsion (at least 8.9%) when compared with the INT model. When cord pretension was increased, annulus stress was slightly increased in flexion and decreased in extension. Despite this, the differences between these Dynesys models were still within 9% for all motions. Both adjacent levels exhibited similar trends, as shown in Fig. 4.

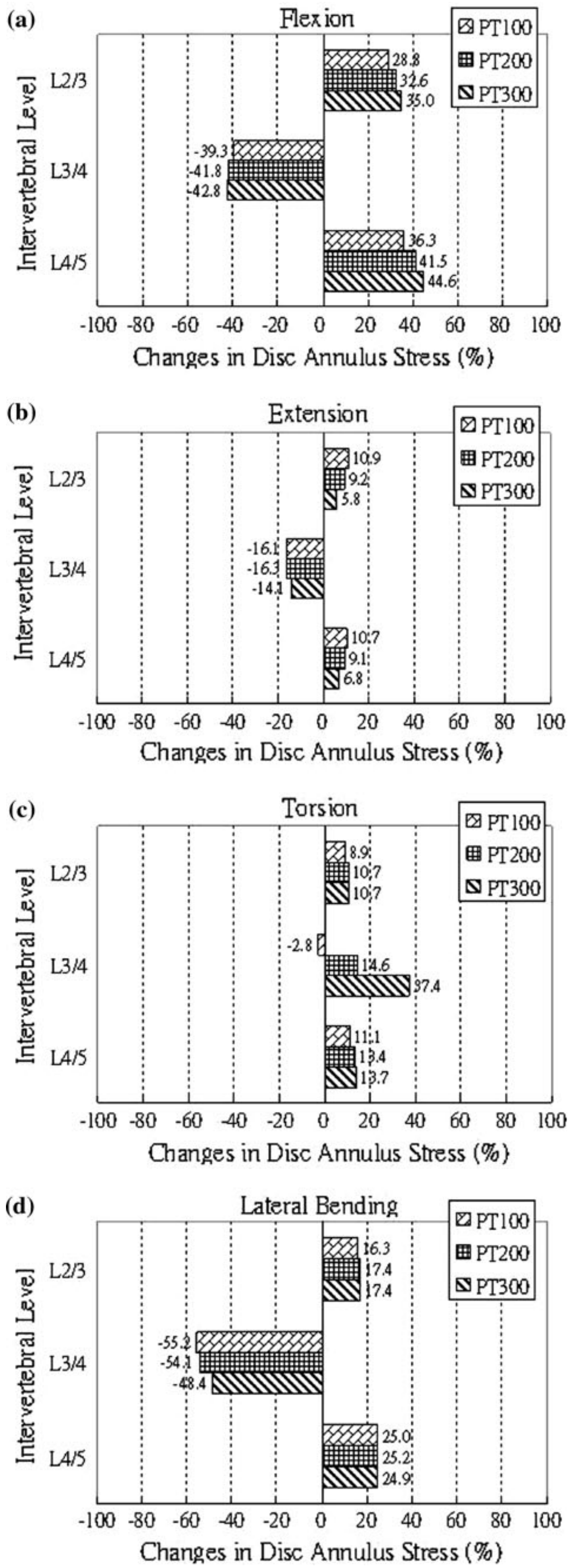
Table 3 The maximum screw stress among the PT100, PT200, and PT300 models (units: MPa)

Finite element model	Physiological motions			
	Flexion	Extension	Torsion	Lateral bending
PT100	95.4	104.0	36.6	121.0
PT200	96.6	101.0	39.5	120.0
PT300	100.0	96.1	42.9	122.0

Discussion

The basic intent of the Dynesys dynamic stabilisation system is to reduce the stiffness of the rigid fixation system to allow limited motion at the implanted level and to prevent stress concentrated at adjacent levels [1]. This FE analysis showed that the insertion of Dynesys reduced the ROM of the implanted level in terms of flexion and lateral bending, but less so in extension. The results are in good agreement with most previous biomechanical studies in which the Dynesys reduced the ROM below the magnitude of the intact spine for the implanted level under flexion and lateral bending [8, 9, 11, 13]. For the adjacent levels, the present study revealed that the ROM increased by at least 22.4% in flexion and 15.6% in lateral bending; these trends conflict with the findings of some in vitro tests under the load control method [9, 21]. Our earlier study demonstrated that the effects on adjacent levels are more prominent under the displacement control method after insertion of a spinal implant when compared with the load control method [18]. This conflicting result can be explained by the use of different testing protocols in the present FE simulation and previous in vitro tests.

The present study found that the cord pretension has a strong influence on the ROM at the implanted level in flexion–extension and a minor influence on the ROM at the adjacent levels. Particularly in flexion, the stiffness of the implanted level showed a great increase, from 100 to 300 N of core pretension. However, in extension, the stiffness of the implanted level exhibited a decrease when the core pretension was increased. The reason should be attributed to a closure of the facet joint and a greater FCF in PT300 model. Additionally, the lordosis angle of the L3–L4 motion segment increased 1.17° after giving a pretension of 300 N but only increased 0.44° after giving a pretension of 100 N. Because a pretension of 100 N resulted in less alteration of lordosis angle and FCF, the extension moment was more easily able to generate a greater ROM, and thus, a lower stiffness was calculated in extension. There was only a slight change in torsion, because the core pretension did not act directly on a transverse plane. These results indicate that the stability of the lumbar spine under flexion–extension can be modified



◀ **Fig. 4** Changes in annulus stress (% of intact) at the implanted and adjacent levels under **a** flexion, **b** extension, **c** torsion, and **d** lateral bending

by applying different cord pretensions during the implantation of Dynesys. The results also imply that a 300-N pretension causes a much higher stiffness at the implanted level when compared with the intact lumbar spine. Additionally, increasing the cord pretension from 100 to 300 N resulted in marked increases of 35 and 32% in the FCF of the implanted level in extension and torsion, respectively. The values were always below the magnitude of the intact spine for the three cord pretension cases under extension; however, the FCF was 22.6% higher than the intact spine for the case of 300-N cord pretension under torsion. These results indicate that Dynesys can alleviate facet loading on the implanted level under extension, and they follow a similar trend to the results presented by Rohlmann [11]. Although FCF was noticeably increased at the adjacent levels, it remains much lower than the values of the fusion model (+169% under extension; +28.9% under torsion) [20]. At present, several clinical reports have demonstrated that Dynesys implantation does not induce a problem in the adjacent facet joint [2–7]. Therefore, the present study deems that Dynesys resulted in a lower incidence of adjacent facet degeneration than did the rigid fixation system. In addition, although a higher cord pretension can slightly reduce the FCF of the adjacent levels, it only had a minor influence on the adjacent facet joint due to the considerably lower FCF when compared with the fusion case.

Some clinical studies have reported that a disc degeneration of between 7.5 and 47% was noticed at the adjacent level after implantation of the Dynesys system [4–7]. The long-term complications of ASD may be explained by a prominently increased disc annulus stress of the adjacent levels under flexion and lateral bending following the implantation of Dynesys when using a 300-N cord pretension. A lower cord pretension of 100 N can reduce the adjacent disc annulus stress for flexion by approximately 9%. This implies that a lower cord pretension, such as 100 N, could reduce the possibility of developing ASD disease in the disc. However, more evidences are needed to support this assumption.

Several clinical reports also noted the complication of a broken screw [4, 6, 7]. The location of high screw stress in FE analysis was different from the location of screw breakage in a previous clinical report [7]. One possible reason may be the modelling of the bone–screw interface and consideration of the screw thread. This study did not model the screw thread and assumed a good connection between the bone and screw. However, screw breakage was noticed in the clinic at the screw thread. In reality, a repetitive loading destroyed the interface between the bone and screw. Thus, a greater stress was gradually transferred to screw tip and eventually fractured inside the vertebral body. However, in FE analysis, the effective bond at bone–screw interface caused that stress to be primarily concentrated at the junction between the bone and screw. Therefore, high screw stress was unable to transfer to the screw tip in FE analysis. Despite to that, the FE analysis calculated that the high stress region of the screw was markedly reduced when using a lower cord pretension, particularly in the case of flexion. Because the Dynesys system is intended to maintain spinal stability for a long time rather than to assist bony fusion as a conventional rigid spinal fixator, the screw stress in the Dynesys system should be as low as possible while the spinal segment is stabilised.

With respect to the model limitations, this study did not consider various grades of disc degeneration and only considered a single geometry of the lumbar spine. Regarding the simplification of the spinal implant, visco-elastic behaviour of PCU and PET was not considered. In addition, there was only one length of the spacer, and the thread of the pedicle screw was neglected. Meanwhile, the screw stress in the FE analysis may be underestimated because of the lack of a screw thread. The loading condition of the FE model was only applied to the displacement control method. The previous study [18] compared the biomechanical effect between the load control method and displacement control method. If the load control method is applied to the evaluation of non-fusion spinal implants, the variation of ROM for the stabilised motion segment is more remarkable. Conversely, in order to realise the effect of the adjacent segment, the FE model applied to the displacement control method is more adequate, because the ROM of the adjacent segment is more remarkable. However, the results for the adjacent levels depend on the number of vertebra included in the model. Regarding the consideration of the loading condition, a realistic load simulating the upper body [22], rather than the preload of 150 N, should be used to conduct FE analysis in the future studies. Finally, the presented FE model did not consider the nuclectomy and laminectomy surgeries. The ROM, FCF, and disc annulus stress at the implanted level could increase after removing these spinal structures.

Although the FE analysis was conducted with some assumptions and limitations, the FE model was able to quantify biomechanical alteration for a Dynesys system with various core pretensions, such as in ROM, FCF, annulus stress, and screw stress. The FE analysis also indicated that adjusting core pretension might affect ROM, FCF, and annulus stress within the construct but not in the adjacent segment, particularly in flexion–extension. Therefore, the biomechanical behaviour of the lumbar spine under flexion–extension can be modified by applying different cord pretensions during the implantation of Dynesys.

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Conflict of interest None.

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