# Combined effect of artificial cervical disc replacement and facet tropism on the index-level facet joints: a finite element study

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**Research Article**

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# Abstract

Objective

The purpose of this study was to assess the impact of facet tropism on the facet contact force and facet capsule strain after artificial cervical disc replacement (ACDR).

Methods

A finite element model was constructed from computed tomography (CT) scans of a 28-year-old male volunteer. Symmetrical, moderate asymmetrical (7 degrees tropism), and severe asymmetrical (14 degrees tropism) models were created at the C5/C6 level. C5/C6 ACDR was simulated in all models. A 75 N follower load and 1 N·m moment was applied to the odontoid process to initiate flexion, extension, lateral bending, and axial rotation, and the range of motions, facet contact forces, and facet capsule strains were recorded.

Results

In the severe asymmetrical model, the right-side FCF increased considerably under extension, right bending, and left rotation compared with the symmetrical model after C5/C6 ACDR. The ride-side FCFs of the severe asymmetrical model under extension, right bending, and left rotation were about 1.7, 3.1, and 1.8 times of those of the symmetrical model, respectively. The facet capsule strains of both the moderate and severe asymmetrical models increased significantly compared with those of the symmetrical model after C5/C6 ACDR. The left-side capsule strains of the severe asymmetrical model were 2.1, 2.4, 1.6, and 8.5 times of those of the symmetrical model under left bending, right bending, left rotation, and right rotation, respectively. The right-side capsule strains of the severe asymmetrical model were 6.3, 1.6, 3.7, and 2.2 times of those of the symmetrical model under left bending, right bending, left rotation, and right rotation, respectively. The stress distribution on the facet surface in the asymmetrical models was different from that in the symmetrical model.

Conclusions

The existence of facet tropism could considerably increase facet contact force and facet capsule strain after ACDR, especially under extension, lateral bending, and rotation. Facet tropism also could result in abnormal stress distribution on the facet joint surface and facet joint capsule. Such abnormality might be a risk factor for post-operative facet joint degeneration progression after ACDR, making facet tropism noteworthy when ACDR was considered as the surgical option.


# Introduction

Long-term outcomes proved the artificial cervical disc replacement (ACDR) to be an efficient treatment for cervical degenerative disc disease. Long-term results from several randomized controlled trials showed that ACDR had superior overall success, fewer secondary surgeries, and lower adjacent level degeneration rate than anterior cervical discectomy and fusion (ACDR)[1−6]. By preserving the surgical level range of motion (ROM), ACDR seemed able to protect the adjacent level discs. However, clinical information regarding the facet joint alterations after ACDR was limited and contradictory. At two-year follow-up after ACDR, Ryu et al.[7] observed progression of facet arthrodesis in approximately 20% of their cohort. In contrast, Meisel et al.[8] found no radiographic progression of facet degeneration during a four-year observation after ACDR.

The intervertebral disc and the facet joints formed the three-joint complex. Change of any part could affect the others. Besides, the cervical facet joints played an important role in guiding the motion and transferring much of the load. In such case, in contrast to limited clinical evidence, both cadaveric studies and finite element (FE) analyses assessed the effect of ACDR on the facet joints. In short, ACDR could alter the facet contact force, the stress distribution, and facet capsular ligament strain, to various degrees based on the kinematic design of the prosthesis. And it was speculated that abnormal loading could be the risk factor for joint degeneration. Facet tropism, angular difference between left- and right-side facet joint orientation, is common in the sub-axial cervical spine and related to facet joint degeneration[9]. We previously demonstrated that facet tropism could result in increased cervical facet pressure under flexion, extension, lateral bending, and axial rotation[10].

What is the combined effect of ACDR and facet tropism on the index level facet joint? We conducted this FE study to assess the influence of facet tropism on the facet contact force and facet capsule strain after ACDR with a semi-constrained artificial disc.

# Materials and methods

A 28-year-old male healthy volunteer (165 cm, 65 kg) signed the informed consent to participate this study. The CT scan of his cervical spine was acquired using a CT scanner (SOMATOM Definition AS+, Siemens, Germany). The slice thickness was 0.75 mm, and the slice increment was 0.69 mm. Data in DICOM format was used for the cervical spine model reconstruction.

The finite element models were developed to study the effect of facet tropism on the facet contact force and facet capsule strain both before and after the C5/C6 arthroplasty by the Prestige LP (Medtronic, Minnesota, US) prosthesis. The software used in the study included Mimics 21.0 (Materialize Inc, Leuven, Belgium), Geomagic Studio 15.0 (3D System Corporation, Rock Hill, South Carolina, USA), Pro/E 5.0 (Parametric Technology Corporation, Massachusetts, USA), and ABAQUS 6.13 (SIMULIA, Rhode Island, US).

# The symmetrical intact model

The CT scans were loaded into Mimics 21.0 software to reconstruct the C2–T1 vertebrae by thresholding and dynamic region growing. Afterwards, the mid-sagittal plane was used to bisect the bony model. The symmetrical model was then created by mirroring the left half part. The bony structures were then imported into the Geomagic studio 15.0 for subdividing, noise reduction, smoothing, and surface fitting. The entity of each intervertebral disc was created by extending the lower and upper endplates of the adjacent vertebrae. Then, the internal disc structures were constructed, including the cartilage endplates, nucleus pulposus, and annulus fibrosus. The nucleus pulposus consist of 44% of the disc space and the substance of the annulus fibrosus consist of 56%. Within the substance of the annulus fibrosus, eight layers of circumferential fibers with 30° to 45° inclination were generated, each layer going in the opposite direction. For each facet joint, two layers of cartilages were enclosed by the simulated facet capsule. Finally, the anterior longitudinal ligament, posterior longitudinal ligament, ligamentum flavum, capsular ligament, intertransverse ligament, interspinous ligament, and supraspinal ligament were added. The modulus of elasticity, Poisson ratio, cross-sectional area, and element type are shown in Table 1. (Fig. 1A)

Table 1
Material properties and mesh types of the cervical finite element model

| | Modulus of elasticity(MPa) | Poisson ratio | Cross-sectional area (mm$^2$) | Element type |
|---|---|---|---|---|
| Cortical bone | 12000 | 0.30 | / | C3D4 |
| Trabecular bone | 100 | 0.20 | / | C3D4 |
| Pedicles | 3500 | 0.25 | / | C3D4 |
| Facet joint cartilage | 15 | 0.45 | | C3D8 |
| Cartilage endplate | 24 | 0.25 | / | C3D8 |
| Nucleus pulposus | 1 | 0.50 | / | C3D8 |
| Annulus fibrosus | 4.2 | 0.45 | / | C3D8 |
| Annulus fibrosus fibers | 175 | / | 0.76 | T3D2 |
| ALL | 7.8 | / | 63.70 | T3D2 |
| PLL | 1 | / | 20 | T3D2 |
| LF | 1.5 | / | 40 | T3D2 |
| CL | 7.5 | / | 30 | T3D2 |
| ITL | 10 | / | 1.80 | T3D2 |
| IL | 1 | / | 40 | T3D2 |
| SL | 3 | / | 30 | T3D2 |
| Titanium alloy | 110000 | 0.3 | / | / |

ALL, anterior longitudinal ligament. PLL, posterior longitudinal ligament. LF, ligamentum flavum. CL, capsular ligament. ITL, intertransverse ligament. IL, interspinous ligament. SL, supraspinous ligament.

## The facet tropism models

Based on the symmetrical model, the moderate and the severe asymmetrical model were developed, as we previously reported. The only difference was the left and right facet orientation at the C5-C6 level. The moderate asymmetrical model was set as 7 degrees tropism whereas the severe asymmetrical model was set as 14 degrees tropism. (Fig. 1B, 1C)

## The ACDR models

The C5/C6 ACDR was simulated in the intact, moderate asymmetrical and severe asymmetrical models. Briefly, the anterior longitudinal ligament, the intervertebral disc, and the posterior longitudinal ligament at C5/C6 level were removed. Then the Prestige LP prosthesis (6 mmĺ16 mm) was inserted into the disc space. The interface between the vertebral endplates and prosthesis footprints were set as tie. The surface-to-surface coefficient of friction was set at 0.1 between the two metal pieces of the Prestige LP prosthesis.

## Material properties assignment and meshing

ABAQUS was used to complete material properties assignment and meshing, as shown in Table 1. The material property of the Prestige LP artificial disc was assigned as titanium alloy (Ti6Al4V). The ligaments were assigned as truss element to withstand tension, not compression.

## Boundary condition

The connections between vertebrae and intervertebral discs, vertebrae and facet cartilages, and ligament insertions to bone were designated as tie. The connection between the facet cartilages was simulated as a sliding contact without friction. The connection between reference point and vertebrae was designated as coupling.

## Experimental condition

The cervical model was fixed with six degrees of freedom at the inferior endplate of the T1 vertebra. A 75 N follower loading was used to represent the head's weight. On the odontoid process of the C2 the 1 N•m moment was given to initiate flexion, extension, lateral bending, and axial rotation. The range of motions (ROMs) under all moments, as well as facet contact forces (FCFs) and facet capsule strains were tested.

## Results

# Validation of the symmetrical model

The symmetrical intact model was validated by comparing the ROMs with previously reported data. The ROMs of flexion, extension, lateral bending, and axial rotation are listed in Table 2. The ROMs of the symmetrical intact model were within the ranges of the ROMs reported in previous studies[11–13].

Table 2
Comparison of the predicted range of motion of current symmetrical intact model and previous reported data.

| | **Present study** | | | **Lee *et al.* (2016)** | | | **Liu *et al.* (2016)** | | | **Panjabi *et al.* (2001)** | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | C4-C5 | C5-C6 | C6-C7 | C4-C5 | C5-C6 | C6-C7 | C4-C5 | C5-C6 | C6-C7 | C4-C5 | C5-C6 | C6-C7 |
| Flexion (°) | 4.66 | 4.52 | 3.41 | 4.21 | 3.61 | 4.32 | 5.84 ± 1.19 | 5.84 ± 0.97 | 4.45 ± 1.62 | 5.3 ± 3.0 | 5.5 ± 2.6 | 3.7 ± 2.1 |
| Extension (°) | 5.02 | 4.70 | 4.42 | 3.47 | 4.49 | 5.43 | 4.89 ± 1.18 | 4.80 ± 0.99 | 3.80 ± 1.50 | 4.8 ± 1.9 | 4.4 ± 2.8 | 3.4 ± 1.9 |
| Axial rotation (°) | 6.89 | 6.25 | 4.98 | 5.17 | 4.11 | 4.36 | 3.58 ± 0.37 | 2.98 ± 0.73 | 1.95 ± 0.82 | 6.8 ± 1.3 | 5.0 ± 1.0 | 2.9 ± 0.8 |
| Lateral bending (°) | 6.75 | 6.21 | 4.77 | 2.31 | 2.00 | 2.28 | 5.06 ± 1.22 | 3.70 ± 0.89 | 2.40 ± 0.60 | 9.3 ± 1.7 | 6.5 ± 1.5 | 5.4 ± 1.5 |

# The segmental ROMs after C5/C6 ACDR

The segmental C5-C6 ROMs after C5/C6 arthroplasty using the Prestige LP are listed in Table 3. In the symmetrical models, the predicted flexion-extension ROM following ACDR increased by 1.53 degrees and rotation ROM increased by 2.57 degrees comparing to the intact model. The flexion-extension, lateral bending, and rotation ROMs of the severe asymmetrical model were 2.22 degrees, 2.41degrees, and 1.22 degrees bigger than those of the symmetrical model following ACDR.

Table 3
Range of motions at C5-C6 level in different models.

| | | **Flexion-extension** | **Lateral bending** | **Rotation** |
|---|---|---|---|---|
| Symmetrical | Intact | 9.22 | 12.46 | 9.14 |
| Symmetrical | ACDR | 10.75 | 12.38 | 11.71 |
| Moderate asymmetrical | ACDR | 12.02 | 12.93 | 10.9 |
| Severe asymmetrical | ACDR | 12.97 | 14.79 | 12.93 |

Table 4
Facet contact forces (FCFs) of intact and C5/C6 ACDR model in different positions.

| | **Intact** | | **ACDR** | | | | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Neutral | | Neutral | | Flexion | | Extension | | Left bending | | Right bending | | Left rotation | | Right rotation | |
| | Left | Right | Left | Right | Left | Right | Left | Right | Left | Right | Left | Right | Left | Right | Left | Right |
| Symmetrical | 3.73 | 4.97 | 4.50 | 5.28 | 2.92 | 3.22 | 12.70 | 14.39 | 11.93 | 0.09 | 0.05 | 12.49 | 0.96 | 13.85 | 13.19 | 2.37 |
| Moderate asymmetrical | 0.08 | 5.51 | 0.59 | 5.04 | 1.15 | 4.67 | 0.94 | 19.23 | 5.46 | 0.19 | 0.07 | 16.96 | 1.18 | 21.30 | 12.50 | 0.05 |
| Severe asymmetrical | 0.09 | 6.19 | 0.81 | 7.22 | 1.42 | 5.94 | 1.44 | 24.57 | 6.52 | 1.68 | 0.57 | 39.25 | 1.72 | 24.78 | 15.87 | 1.40 |

Table 5
Facet capsule strains of intact and C5/C6 ACDR model in different positions.

| | **Intact** | | **ACDR** | | | | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Neutral | | Neutral | | Flexion | | Extension | | Left bending | | Right bending | | Left rotation | | Right rotation | |
| | Left | Right | Left | Right | Left | Right | Left | Right | Left | Right | Left | Right | Left | Right | Left | Right |
| Symmetrical | 3.25 | 3.37 | 14.03 | 14.62 | 4.66 | 6.71 | 22.76 | 25.85 | 31.89 | 5.66 | 8.94 | 39.04 | 35.5 | 10.45 | 9.24 | 39.95 |
| Moderate asymmetrical | 2.44 | 1.66 | 16.51 | 14.64 | 6.24 | 6.80 | 30.41 | 32.11 | 56.94 | 21.75 | 15.35 | 52.46 | 49.59 | 13.04 | 68.22 | 57.72 |
| Severe asymmetrical | 3.56 | 2.56 | 18.09 | 15.91 | 9.92 | 9.43 | 35.15 | 37.38 | 67.73 | 29.86 | 21.17 | 62.66 | 56.80 | 15.34 | 78.19 | 89.56 |

# Facet contact force

In the symmetrical model, when only the 75N axial loading was applied, the left and right facet contact forces (FCFs) at the C5/C6 level were about the same (4.50 N vs. 5.28 N). The left and right FCFs decreased when the cervical spine flexed, whereas increased when extended. The left FCF increased whereas the right FCF decreased when the cervical spine bent to the left and rotated to the right and vice versa. In the asymmetrical models, the FCFs on the right were bigger than those on the left in the neutral position under 75 N preload. The right FCFs in asymmetrical model increased whereas the left FCFs decreased under flexion and extension when comparing to those in the symmetrical models. The severe asymmetrical model had bigger FCFs than the moderate asymmetrical model under flexion and extension, on both sides. In the extension position, the FCF of the severe asymmetrical model on the right (24.57 N) nearly doubled the FCF (14.39 N) of the symmetrical model. When bending to the left side, the FCFs increased on the left side whereas decreased on the right and vice versa. Interestingly, the left-side FCFs of the asymmetrical model were smaller than that of the symmetrical model. When bending to the right side, the right sided FCF of the severe asymmetrical model was about 5.44 folds of the neutral position, and 3.14 folds of the symmetrical model. The left-sided FCFs decreased, whereas the right-sided FCFs increased when the cervical spine rotated to the left, and vice versa. In the left rotation position, the FCFs of the severe asymmetrical, moderate asymmetrical, and symmetrical models were 24.78 N, 21.30 N, and 13.85 N, respectively. (Fig. 2)

In the symmetrical model, the stress distribution was symmetrical between the right and left facet joints when in the neutral, flexion, and extension positions. The stress was concentrated on the cephalad part of the facet joints under flexion, whereas on the caudal part under extension. The stress concentrated on the left facet at the cephalad part when the cervical bent to the left, and vice versa. In the left rotation position, the stress is concentrated on the cephalad part of the right facet and on the caudal part of the left facet. In the moderate and severe asymmetrical models, the stress distribution was more complex. Stress was concentrated on the cephalad part of the facets in every position. For each position, the stress distribution on the left facets were similar between the symmetrical and asymmetrical models. (Fig. 3)

# Facet capsule strain

In the symmetrical model, under the 75N preload, the facet capsule stresses on both sides increased by nearly 4.3 times following the simulated ACDR compared to those in the intact model. Facet capsule strains decreased when the cervical spine was flexed but increased when it was extended. The left-sided facet capsule strains increased, whereas the right capsule strains decreased when the cervical spine bent to the left and rotated to the left, and vice versa. In the moderate and severe asymmetrical models, in each position, the capsule strains on both sides were bigger than the symmetrical model. In the left bending position, the left-sided maximum capsule strain of the moderate asymmetrical model (56.94 MPa) and the severe asymmetrical model (67.73 MPa) was 1.79 times and 2.12 times that of the symmetrical model (31.89 MPa), and the right-sided maximum capsule strain of the moderate asymmetrical model (21.75 MPa) and the severe asymmetrical model (29.86 MPa) was 3.85 times and 5.28 times that of the symmetrical model (5.66 MPa). The changes in maximum capsule stresses under right bending, left rotation, and right rotation followed the same pattern. (Fig. 4)

When the cervical spine was flexed, capsule strains increased at the posterior portion, whereas they increased predominantly at the lateral portions when the cervical spine was extended. In the left bending position, the right side of the left facet capsule and the left side of the right facet capsule increased, and vice versa. When the cervical spine rotated to the left, the right-sided capsule strain increased at the left-posterior portion. The strain on the left-side capsule increased at the left-posterior and right-anterior portions. When the cervical spine rotated to the right, the right-sided capsule strain increased at the right-posterior portion in the symmetrical and asymmetrical models. However, the left-sided capsule strain increase shifted from the right-posterior portion in the symmetrical model to the left-anterior portion in the severe symmetrical model. (Fig. 5)

## Discussion

Long-term clinical results demonstrated high clinical success rate and satisfaction rate after lumbar TDR. Kitzen et al[14]. reported a 79.6% satisfaction rate after lumbar TDR at a mean 12.3-years follow-up. Park et al.[15] reported 76.9% clinical success rate and 87.2% satisfaction rate after lumbar TDR at mean 5 years follow-up. David et al.[16] also reported 82.1% clinical success rate at 13.2 years follow-up. In an IDE study, Radcliff et al. suggested the lumbar TDR were safe and effective for single-level lumbar DDD[17]. They also found no significant increase of radiological presence of facet joint degeneration at 7-year follow-up[17]. However, many clinical studies demonstrated facet joint degeneration (FJD) after lumbar TDR, some requiring reoperation. Shim et al. observed degradation of FJD in 32% patients after ProDisc after 36–40 months and in 36.4% patients after CHARITE after 36–48 months[18]. Park et al. found 12 of 41 (29%) TDR levels had FJD progression after 32.2 months (26–42 months) and related with female, malposition, and 2-level TDR[19]. Early-results from the Norwegian Spine Study Group demonstrated that significantly more patients either had newly-onset or progressed FJD at the surgical level in the lumbar TDR group compared with the rehabilitation group (34% vs. 4%, P < 0.001) at two-year follow-up[20]. In a later long-term study, they found similar FJD rate at 8-year follow-up compared with the 2-year follow-up but found no association between FJD and clinical outcomes[21]. Pimenta et al. followed 15 patients for 7 years to find 7 patients with FJD and only 1 symptomatic[22]. On the contrary, some authors believed FJD was the main cause of postoperative back pain for lumbar TDR patients. Siepe et al. confirmed facet joint pain by fluoroscopically guided spine infiltrations in 9.1% patients after L4/L5 TDR, 28.1% after L5/S1 TDR, and 60% after bi-segmental TDR, using ProDisc II[23]. They in a later study observed that progression of FJD was present in 20% of all facet joints at 53.4 months follow-up and more common at lumbosacral junction, and progression of FJD was associated with lower ROM at the index level and inferior VAS and ODI scores[24]. A small portion of patients with facet joint complains eventually received revision surgery. Siepe et al. reported 29 revisions of 201 patients, among those 2 re-operations were due to facet complains[25]. David et al. reported 10.4% (11 of 106) reoperation rate at the index-level after lumbar TDR[16]. Five patients underwent revision surgery due to symptomatic FJD[16]. Punt et al. performed revision surgery for 75 patients after lumbar TDR, among them 25 patients (33.3%) present with FJD[26]. Schmitz et al. reported that 85% of the 48 patients who had revision present with FJD and concluded that FJD was the most important cause for revision after lumbar TDR[27]. In vitro cadaveric study and finite element analysis suggested that abnormal loading and aberrant kinematics of the facet joints after lumbar TDR contributed to the FJD[28–31].

Presently, long-term results of ACDR for single-level and two-level cervical spondylosis have been published. These studies showed that ACDR, performed with different prostheses, showed higher or comparable overall successful rate and satisfaction rate, lower incidence of adjacent level degeneration, fewer revisions at either the index level or the adjacent levels, comparing to ACDF[1–6]. Interestingly though, these long-term studies did not specifically describe the facet joint degeneration after ACDR. Ryu et al. reported progression of FJD in 19.4% (7 of 36) levels treated with ACDR (1 of 19 Bryan and 6 of Prodisc-C) at 24 months follow-up[7]. They found that anterior placement of the Prodisc-C was associated with FJD. They argued that anterior placement of the Prodisc-C could increase the load on the facet joints at the index level. Meisel et al. on the other hand, in their multi-center study composed of 200 ACDR patients using Active-C, observed no FJD progression at the 4-year follow-up[8]. On the contrary to the limited clinical data on the post-operative alteration of the facet joints after ACDR, many cadaveric studies and finite element studies assessed the effect of ACDR on the facet joints.

Chang et al. used strain gauges provided by Vishay Micro-Measurements, Inc. in a nondestructive manner to measure the facet joint force after single-level ACDR using Prestige[32]. The results showed that facet joint force increased under flexion, extension, lateral bending, and axial rotation[32]. The results also showed 95.4% increase of the facet joint force under extension and 19.7% under flexion after insertion of the Prodisc-C prosthesis, but 6.4% decrease of and insignificant change of facet joint force under lateral bending and rotation[32]. Interestingly, Crawford et al. used the same strain gauge to find mild decrease of facet load during flexion and no significant change during extension[33]. Jaumard et al.[34] and Bauman et al.[35] further complicated the scenario regarding the cadaveric study of ACDR. They placed a tip pressure transducer inside the facet joint capsule without cutting it open. The results showed no significant alteration of the facet pressure in flexion and extension, but significant increase of facet pressure in ipsilateral lateral bending and torsion[34,35]. Partel et al. placed the film senser into the facet joint after cutting the facet capsule to directly record the facet force after C4/C5 disc replacement using Prodisc-C[36]. The facet forces under extension, lateral bending, and rotation were 28.75N, 55.33N, and 61.36N in the intact model, and 41.87N, 58.96N, and 58.31N in the ADCR model. The results indicated the Prodisc-C only increased the facet force under extension. These inconsistent results reflected the hardship of measuring the facet force/pressure in the cadaveric specimens. The indirect measurements by strain gauges needed complex process of calibration, yet still lacked validation. The number of gauges used, and the position of gauge placed would also significantly affect the estimation of the facet force. The film sensor could provide direct reading of the facet force but required cutting open the facet capsule. The changed biomechanics of the functional spinal unit due to cutting of capsular ligaments would therefore result in inaccurate evaluation of the effect of ACDR on facet force.

Finite element (FE) analysis provided an alternative approach to study the facet force after ACDR. Lee et al. constructed a cervical model (C2-C7) with simulated ACDR at C5/C6 level by either Prodisc-C or Mobi-C[37]. The results showed increased facet force about two times larger than that in the intact model under extension. The capsular ligament tension increased under flexion in the Prodisc-C model (34 MPa) and the Mobi-C model (25.8 MPa), comparing to the intact model (20 MPa). Gandhi et al. changed the material properties of the IVD to simulate its degenerative state and studied the post-operative biomechanics of the cervical spine after ACDR by either Bryan or Prestige LP at C5/C6 level[38]. The results demonstrated considerable increase of facet force under all loadings except left lateral bending in the Prestige LP model. A team from Medical College of Wisconsin presented a series of FE studies comparing the biomechanical effect of different prostheses, including Bryan, Prodisc-C, Prestige LP, Mobi-C and Secure-C[39–42]. The facet force increased in all ACDR models under extension, with Bryan and Secure-C to a lesser extend while Mobi-C, Prestige LP and Prodisc-C to a greater extend. Besides, comparing to the intact model, facet force increased under flexion for Prodisc-C, Prestige LP, increased under lateral bending for Bryan, Prestige LP, Mobi-C and Secure-C, all to various degrees.

These prostheses differed in structure design, number of components, bearing surfaces, and articulation design. Kinematic degrees of freedom, bult-in stiffness, and patients- or surgical-related factors all played crucial parts in the postoperative biomechanics of the cervical spine[43]. Presently, center of rotation, one of the prosthetic traits, has been meticulously studied. Ahn et al. simulated three types of CORs and studied their impact on the cervical facet force, a fixed COR at the disc level, a fixed COR below the endplate (6.5 mm below the disc level), and a mobile COR at the level[44]. The results showed that the fixed COR at the disc level considerably increased the facet force under extension and lateral bending (364.5 N and 104.9 N) comparing to the intact model (14.3 N and 51.5 N)[44]. The lower fixed COR increased facet force under extension to a lesser degree (91 N), whereas the mobile COR did not build up the facet force under all loadings[44]. Galbusera et al. showed that when the COR was fixed, lower the COR (close to the physiological position under the endplate) would result in lower facet force in extension and lateral bending[45]. Rousseau et al. corroborated this finding by showing the posterior and lower positioned COR result in smallest facet force[46]. Faizan et al. showed that the facet forces were smaller for the design with inferior ball component indicating lower position of the COR[47]. Mo et al. showed that prosthesis with mobile COR resulted in smaller maximum facet stress compared to the fixed COR[48]. In short, these studies suggested that ACDR would result in increased facet force at the index level, which was believed to be a risk factor for development or progression of FJD. The increased FCFs could cause micro-injury to the facet joints. The accumulation of these micro-injuries with daily neck activity in time could therefore initiate or accelerate the FJD process.

Apart from the prosthesis- or surgical-related factors, here we present an anatomical-related factor that could alter the facet force and facet capsule strains after ACDR. We previously reported that facet tropism could cause increase of FCFs comparing to symmetrical model[10]. This finding was in accordance with previous biomechanical and FE studies[49–51]. Further, in this study we showed that, ACDR together with facet tropism, could magnify the abnormal facet loading caused by facet tropism alone. In theory, abnormally increased facet loadings could be related to the FJD. Though evidence regarding the effect of facet tropism on FJD after ACDR was not available now, data from the lumbar spine supported the hypothesis that facet tropism could be associated with FJD progression after total disc replacement (TDR). Shin et al. observed that the FJD progression levels had significantly larger facet tropism than the non-FJD progression levels at the 36-month follow-up after lumbar TDR using ProDisc-L[52]. Besides, we previously observed higher facet joint degeneration rate at the cervical level with facet tropism in cervical spondylosis patients[9]. Yet, long-term observation on the alteration of the facet joints after ACDR are needed to validate such theory in the cervical spine.

This study has some limitations. First, the cartilages of the opposing facet joints were simulated as flat components. In the real world, the surface of facet joints has various shapes. The flat type was only one of many. However, the purpose of this study was to examine the effect of facet tropism on facet stress distribution, and the flat form of the facet joint was the most straightforward candidate for presentation. In certain circumstances, a more complex analysis may be necessary. Second, the cervical model was based on a young male with no symptoms. The results should be applied with caution due to the possibility that degenerative changes of the cervical spine would complicate their interpretation. Thirdly, only the Prestige LP cervical disc replacement was evaluated in this study. Prestige LP was considered a semi-constrained artificial disc. Constrained versus unconstrained artificial discs, such as Prodisc-C vivo or Mobi-C, might have a distinct effect on the cervical spine in the presence of facet tropism. Additional investigation is necessary to determine the influence of artificial disc design on facet stress distribution when facet tropism is present.

The existence of face tropism could considerably increase facet contact force and facet capsule strain after ACDR, especially under extension, lateral bending, and rotation. Facet tropism also could result in abnormal stress distribution on the facet joint surface and facet joint capsule. Such abnormality might be a risk factor for post-operative facet joint degeneration progression after ACDR, which needs long-term clinical study to verify. Nevertheless, facet tropism might worth paying attention to when ACDR was considered as the surgical option.

## Declarations

### Ethical Approval

This study was approved by the Ethical Committee of West China Hospital of Sichuan University. All patients had given the informed consent to allow their information to be used in research purposes.

### Competing interests

The authors declare that they have no competing interests.

### Authors' contributions

XR and JL wrote the main manuscript text. XR, JL and JQZ made substantial contribution to analysis and interpretation of data. RX, JL, BYW, and KKH made substantial contributions to conception and design. HL gave the final approval of the version to be published. All authors read and approved the final manuscript.

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None

### Availability of data and materials

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

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## Figures

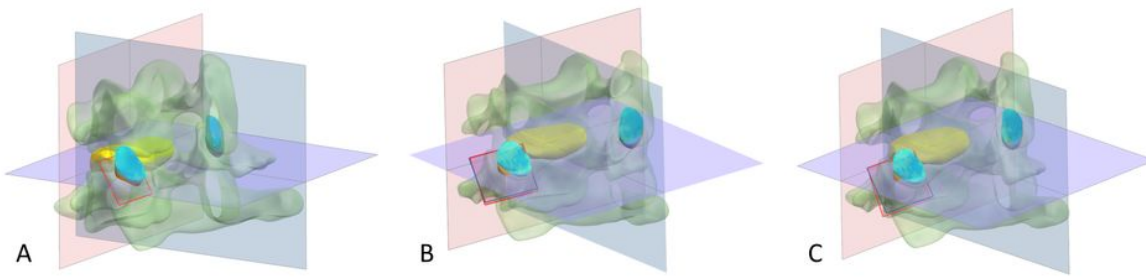


**Figure 1**

Illustration of symmetrical model (A), moderate asymmetrical model (B), and severe asymmetrical model (C) at the C5/C6 level.

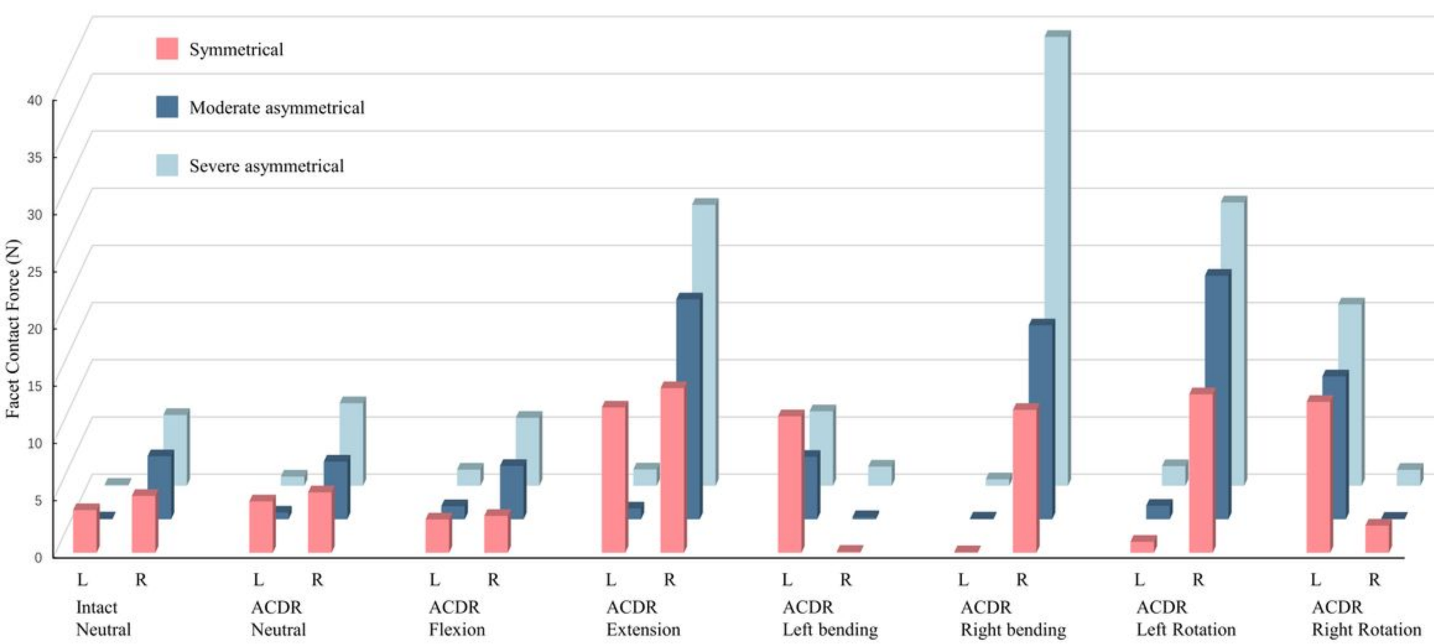


**Figure 2**

Facet contact forces (FCFs) at the C5/C6 level in the symmetrical model, the moderate asymmetrical model, and the severe asymmetrical model before simulated ACDR in neutral position under 75 N preload and after simulated ACDR in neutral position under 75 N preload and different motion condition under 1 N•m moment plus 75 N follower load.(L, left side. R, right side)

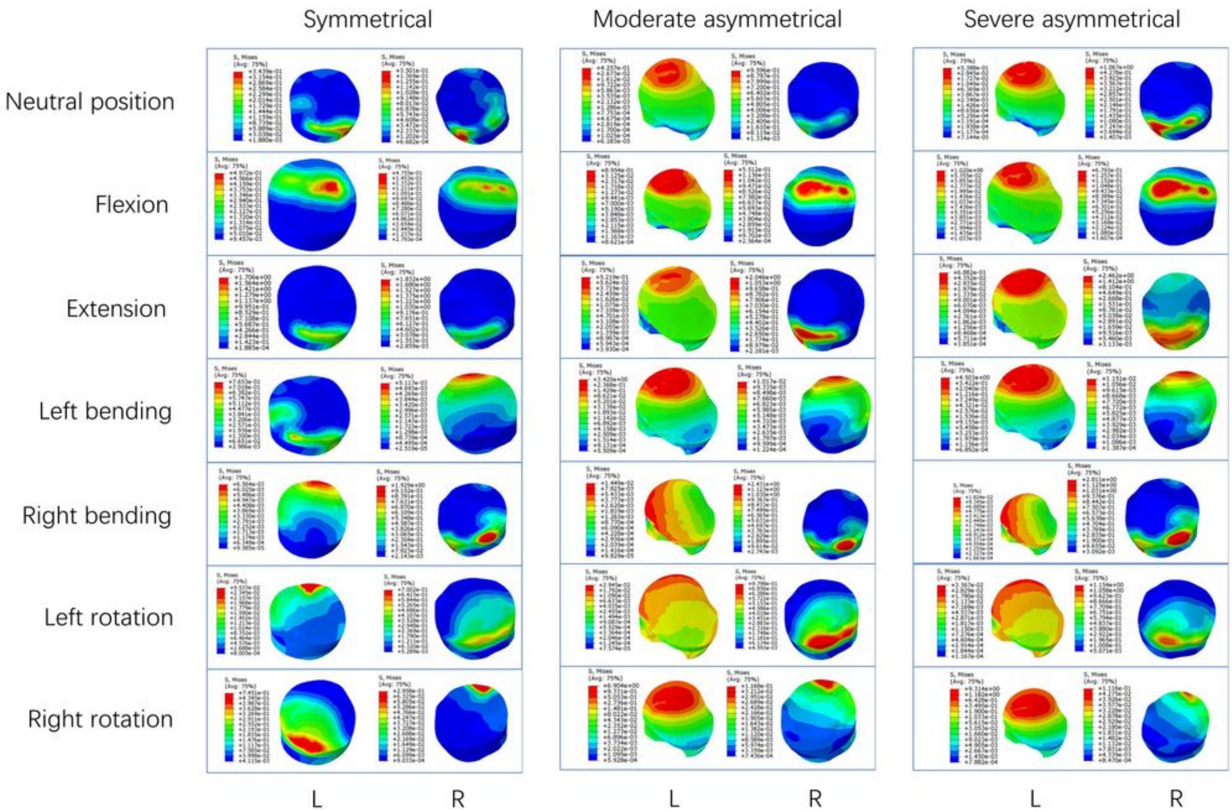


**Figure 3**

Stress distribution on the joint surfaces of the C6 superior articular process after simulated C5/C6 ACDR in the symmetrical model, the moderate asymmetrical model, and the severe asymmetrical model in neutral position under 75 N preload and different motion condition under 1 N•m moment plus 75 N follower load.(L, left side. R, right side)

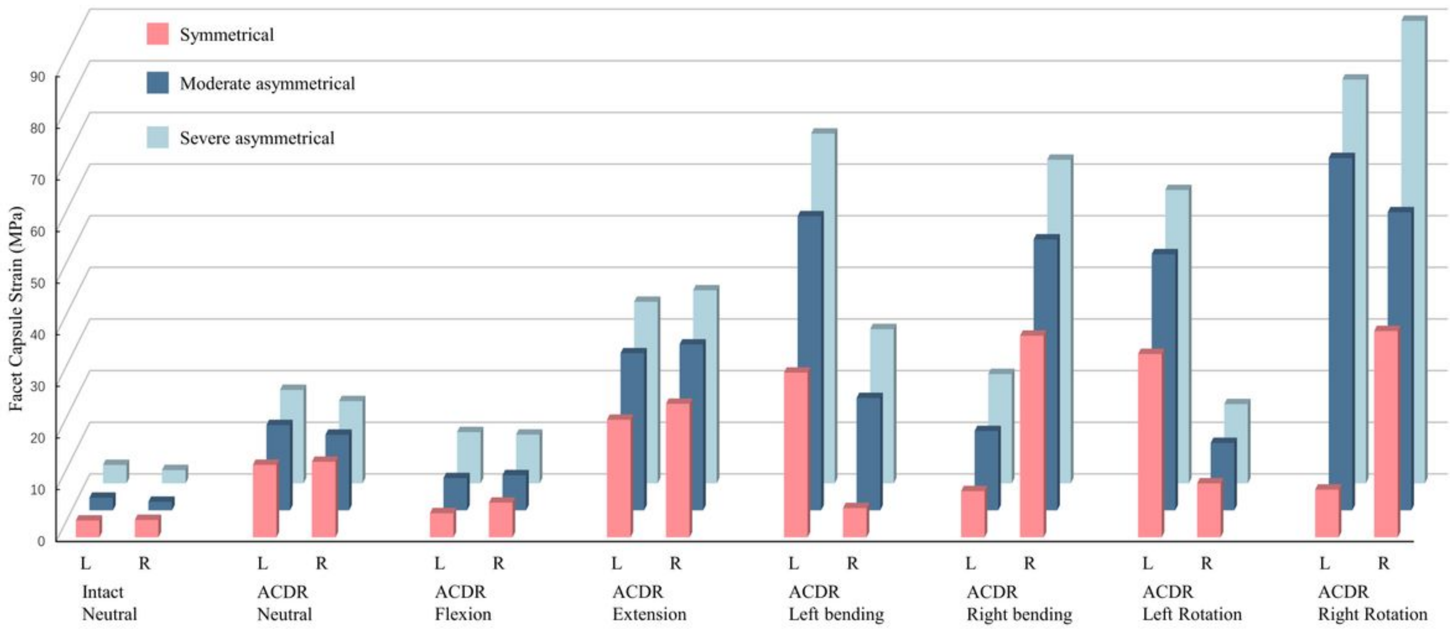


Figure 4

Facet capsule strain at the C5/C6 level in the symmetrical model, the moderate asymmetrical model, and the severe asymmetrical model before simulated ACDR in neutral position under 75 N preload and after simulated ACDR in neutral position under 75 N preload and different motion condition under 1 N•m moment plus 75 N follower load. (L, left side. R, right side)

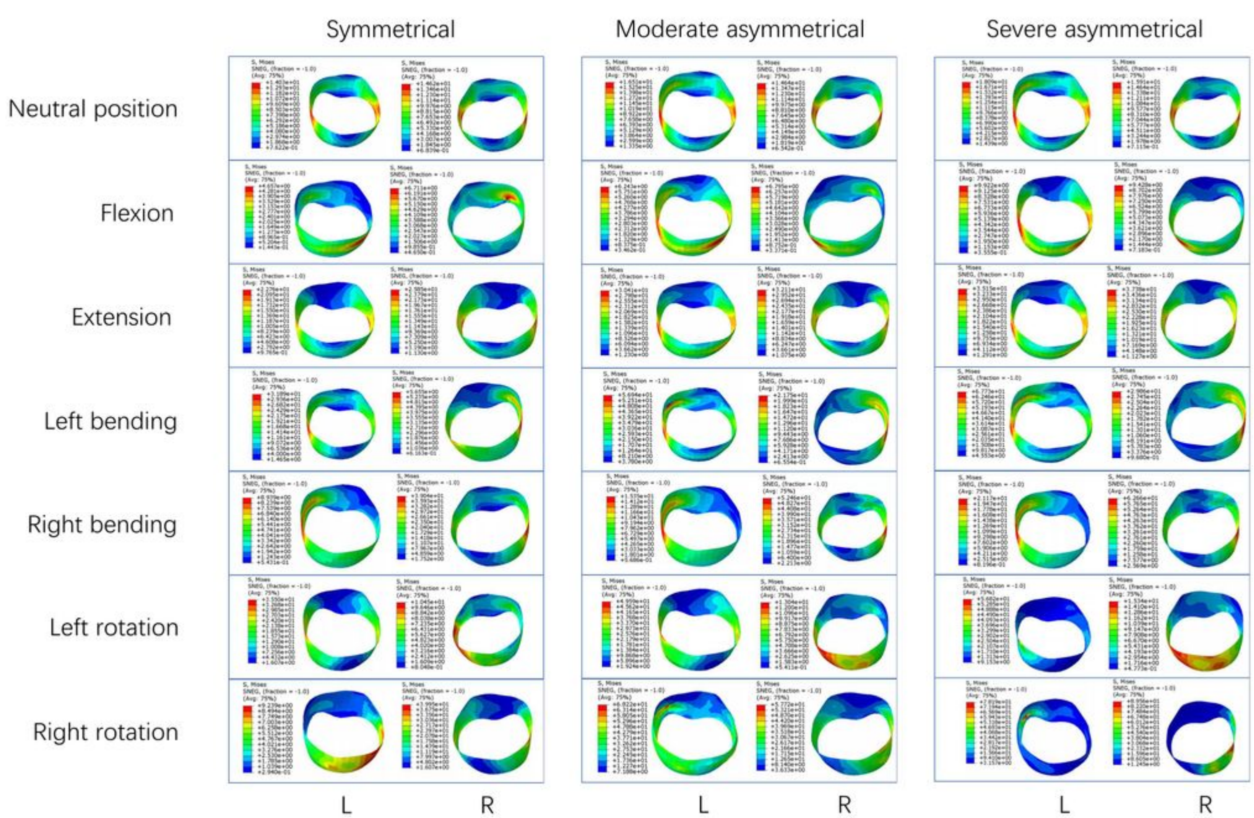


Figure 5

The C5/C6 Facet capsule strain distribution after simulated C5/C6 ACDR in the symmetrical model, the moderate asymmetrical model, and the severe asymmetrical model in neutral position under 75 N preload and different motion condition under 1 N•m moment plus 75 N follower load.(L, left side. R, right side)