

Received: 2019.08.08

Accepted: 2019.11.05

Available online: 2020.01.20

Published: 2020.02.05

Biomechanical Comparison of 1-Level Corpectomy and 2-Level Discectomy for Cervical Spondylotic Myelopathy: A Finite Element Analysis

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Manuscript Preparation E
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Source of support: This project was supported by the Key Research Project of Shanxi Province (2017ZDCXL-SF-01-05)

Background: Anterior cervical discectomy and fusion (ACDF) and anterior cervical corpectomy and fusion (ACCF) are effective treatments for cervical spondylotic myelopathy (CSM), but it is unclear which is better. In this study, we compared the biomechanical properties of 2-level ACDF and 1-level ACCF.


Material/Methods: An intact C3–C7 cervical spine model was developed and validated, then ACDF and ACCF simulation models were developed. We imposed 1.0 Nm moments and displacement-controlled loading on the C3 superior endplate. The range of motions (ROMs) of surgical and adjacent segments and von Mises stresses on endplates, fixation systems, bone-screw interfaces, and bone grafts were recorded.

Results: ACDF and ACCF significantly reduced the surgical segmental ROMs to the same extent. ACCF induced much lower stress peaks in the fixation system and bone-screw interfaces and higher stress peaks on the bone graft. ACDF induced much lower stress peaks on the C4 inferior endplate and equivalent stress on the C6 superior endplate. There was no difference in the ROMs of surgical and adjacent segments and the intradiscal stress of adjacent levels between ACDF and ACCF.

Conclusions: Both ACDF and ACCF can provide satisfactory spinal stability. ACDF may be beneficial for subsidence resistance due to the lower stress peaks on the endplate. The ACCF may perform better in long-term stability and bone fusion owing to the lower stress peaks in the fixation system and bone-screw interfaces, and higher stress peaks in the bone graft.

MeSH Keywords: **Biomechanical Phenomena • Finite Element Analysis • Spinal Fusion**

Full-text PDF: <https://www.medscimonit.com/abstract/index/idArt/919270>

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Background

Cervical spondylotic myelopathy (CSM) is a progressive degenerative disease caused by symptomatic compression of the spinal cord. Ongoing CSM can induce a series of neurological symptoms and severely decrease the quality of life [1,2]. Once the CSM has been diagnosed, and conservative treatments do not work, surgical decompression should be performed to prevent deterioration and improve neural function [3]. An anterior, posterior, or combined anterior-posterior approach can be employed according to the clinical situation and the experience of surgeons [4]. When the soft-disc herniation involved 1 to 2 levels, accompanied by kyphosis and severe axial neck pain, an anterior approach such as ACDF and ACCF is favored [5]. Both ACDF and ACCF have been proved to be reliable and effective in spinal cord decompression, and sagittal alignment restoration and maintenance thus achieved a good clinical outcome. However, there is controversy about which method – ACDF or ACCF – is better [6]. Previous studies suggested that ACDF can shorten operation time, reduce blood loss and complication rates, and maintain greater cervical lordosis, while ACCF performed better in complete decompression, with lower risk of pseudarthrosis [7,8]. However, few studies have investigated the potential mechanism behind the differences between ACDF and ACCF. The results of previous clinical studies can help guide clinical decision-making but do not improve surgical approaches.

In the present study, the biomechanical properties of ACDF and ACCF, which cannot be observed in clinical research and traditional biomechanical study of cadavers, were investigated using finite element analysis (FEA) to provide information for improving the surgery method.

Material and Methods

The FEA modeling method has been described in our previous study [9]. Briefly, the computed tomography (CT) images of the lower cervical spine were obtained from a healthy man (aged 31 years, 76 kg, and 177 cm) at 1-mm intervals. The CT images were then processed in Mimics 19 and 3-Matic 11 software (Materialise, Inc., Belgium) and Hypermesh 14.0 (Altair Technologies, Inc., CA, USA) to reconstruct the C3–C7 FE model. This FE model consisted of cortical bones, cancellous bones, intervertebral discs, endplates, articular cartilages, and ligaments. The bony structures and articular cartilages were meshed into tetrahedron elements (C3D4). The joint space between the articular cartilage was set at 0.5 mm. The endplates were modeled as shell elements (S3) with 0.4-mm thickness. The intervertebral disc consisted of nucleus pulposus, annulus ground substance, and annulus fiber; the first 2 were meshed into triangular prism elements (C3D6) and the annulus

fiber was constructed as a net structure with 15° to 45° angles to the horizontal plane. Annulus fiber and intervertebral ligaments were modeled as a truss element (T3D2) with tension-only property. The interface between adjacent cartilages was set as frictionless sliding contact; all other interfaces were set as tie constraint. The constructed intact C3–C7 spine is shown in Figure 1.

To simulate the 2-level ACDF clinical scenario, the C4–C5 and C5–C6 intervertebral disc and corresponding anterior longitudinal ligaments were removed for decompression, then titanium interbody cages filled with bone graft were molded and inserted into the intervertebral space. A ventral cervical plate was fixed to the C4, C5, and C6 vertebral bodies by 6 screws to stabilize the surgical segment (Figure 1E). To simulate 1-level ACCF surgery, the C4–C5 and C5–C6 intervertebral discs, two-third of the C5 vertebral body, and corresponding anterior longitudinal ligaments were removed; a titanium cage filled with bone graft was inserted into the intervertebral space; and a ventral cervical plate was fixed to C4 and C6 vertebral body by 4 screws (Figure 1F). The internal fixation systems were designed and simplified according to the dimension of instruments used in clinic and the intact model. The fixation systems were meshed into a mixed element (C3D6 and C4D10). The screw-vertebra interface was assigned with a tie constraint. The endplate-cage/interbody cage and endplate-bone graft interfaces were set as node–face contact with a friction coefficient of 0.07. The material properties are shown in Table 1.

For all FE model, the inferior endplate of the C7 vertebral body was fully fixed. An axial preload of 73.6 N was applied to simulate the physiological compression. A moment of 1 Nm was applied at the superior endplate of C3 vertebra to simulate flexion, extension, lateral bending, and axial rotation. The corresponding overall movement angles of the intact model induced by 1 Nm moment, named displacement-controlled loading, were then applied to the ACDF and ACCF models to simulate the physiological range of motion. The segmental ROM and von Mises stress of the endplate, internal fixation system, screw-vertebral body interface, and bone graft were observed.

Results

Model validation results

The intersegmental ROM at C3–C4, C4–C5, C5–C6, and C6–C7 of the intact model were 7.9, 7.2, 5.2, and 4.2, respectively, in flexion-extension; 2.6, 3.7, 2.1, and 2.8, respectively, in lateral bending; and 4.5, 3.7, 2.3, and 2.3, respectively, in axial rotation. The intersegmental ROMs of the present intact FE model were slightly different from those reported in a biomechanical study [10] but showed good agreement with those of other

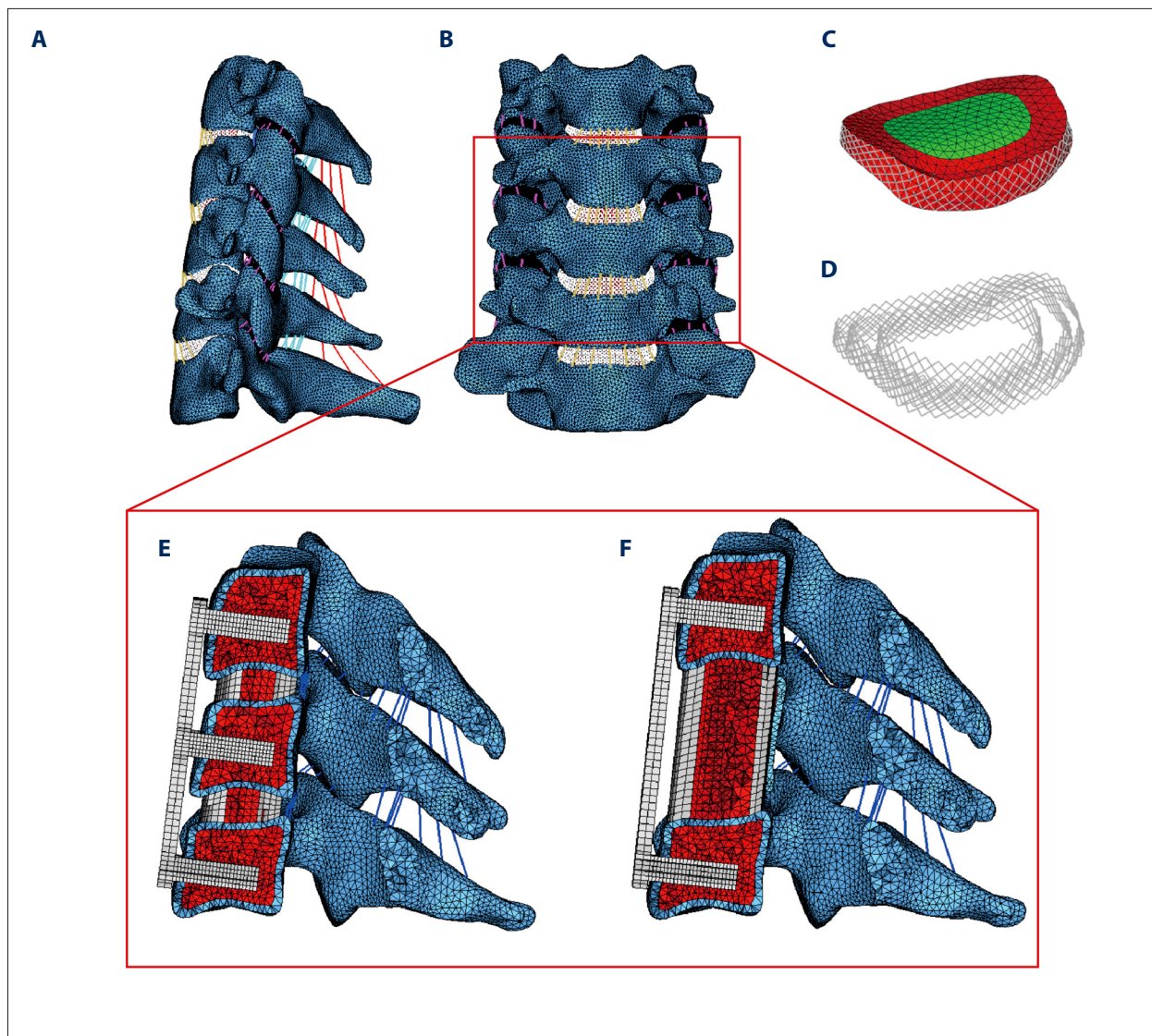


Figure 1. Finite element model of intact C3–C7 cervical spine (**A**, Frontal view; **B**, Lateral view; **C**, Intervertebral disc; **D**, Annulus fiber). (**E**) Finite element model of ACDF. (**F**) Finite element model of ACCF. ACDF – anterior cervical discectomy and fusion; ACCF – anterior cervical corpectomy and fusion.

FE studies [11,12] (Figure 2), which indicated that the intact lower cervical spine model was successfully constructed and could be used for further investigation.

ROMs of the surgical constructs

As shown in Figure 3A, the C4–C6 surgical segment ROMs of the intact lower spine model in flexion-extension, lateral bending, and axial rotation were 12.4, 5.1, and 6.2, respectively. After surgery, the ROMs of ACDF and ACCF under 1 Nm moment were significantly reduced to 0.1 and 0.4, respectively, in flexion-extension; 0.9 and 0.7, respectively, in lateral bending; and 0.4 and 0.3, respectively, in axial rotation. When the displacement-controlled loading was applied, the ROMs of

ACDF and ACCF model were significantly reduced to 0.8 and 1.0, respectively, in flexion-extension; 0.7 and 0.8, respectively, in lateral bending; and 1.3 and 1.0, respectively, in axial rotation. Comparing with that under 1 Nm moment, the ROMs increased in flexion-extension and axial rotation but were almost unchanged in lateral bending.

Fixation systems stresses

The maximum von Mises stresses of the fixation systems are shown in Figure 3B. Displacement-controlled loading significantly increased the stress of the fixation systems in ACDF and ACCF in flexion (87.6%, 80%), extension (362.9%, 386.2%), left bending (603.0%, 638.8%), right bending (578.1%, 473.0%),

Table 1. Material properties assigned to the finite element models.

Material	Elastic modulus (MPa)	Poisson ratio	Cross-sectional area (mm ²)
Bones			
Cortical bone	12000	0.3	–
Cancellous bone	100	0.2	–
Endplate	500	0.4	–
Posterior structure	3500	0.25	–
Bone graft	3500	0.3	–
Facet cartilage	10	0.4	–
Implants			
IFC/APC/screw	110000	0.3	–
Ligaments			
ALL	30	0.4	6.1
PLL	20	0.4	5.4
LF	10	0.4	50.1
ISL	10	0.4	13.1
CL	10	0.4	46.6
Intervertebral disc			
Annulus fiber	450	0.45	
Annulus ground			
Substance	Hyperelastic, Mooney-Rivlin C10=0.56 C01=0.14		
Nucleus pulposus	Hyperelastic, Mooney-Rivlin C10=0.12 C01=0.09		

ALL – anterior longitudinal ligaments; PLL – posterior longitudinal ligaments; LF – ligamentum flavum; ISL – interspinous ligaments; CL – capsular ligaments.

left rotation (366.9%, 431.1%), and right rotation (319.2%, 351.7%), as compared with 1 Nm moment. The fixation system stress peaks were much higher in ACDF than that in ACCF under 1 Nm moment and displacement-controlled loading in extension (9.2%, 4.0%), left bending (44.4%, 37.4%), right bending (33.3%, 57.8%), left rotation (53.3%, 34.8%), and right rotation (52.7%, 41.7%); but were lower in flexion (15.9%, 11.2%). The distribution contour of von Mises stress of the fixation system in ACDF and ACCF are shown in Figure 4.

Bone-screw interfacial stresses

Figure 3C shows the maximum von Mises stresses of the bone-screw interfaces. Compared with 1 Nm moment, displacement-controlled loading significantly increased the stress of the bone-screw interfaces in ACDF and ACCF in flexion (80.8%, 75.3%), extension (453.8%, 443.7%), left bending (487.5%, 496.9%), right bending (384.8%, 365.4%), left rotation (311.7%, 340.0%), and right rotation (337.1%, 315.7%). The bone-screw interface stress peaks were much higher in ACDF than in ACCF under 1 Nm moment and displacement-controlled loading in extension (6.9%, 9.0%), left bending (24.5%, 22.6%), right bending (24.2%,

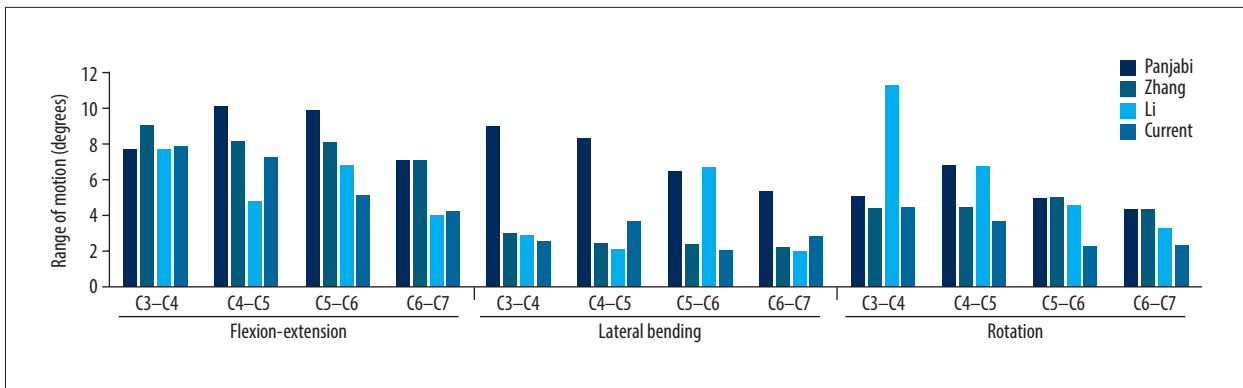


Figure 2. Comparison of ROM between the current intact model and the results of previous studies. ROM, range of motion.

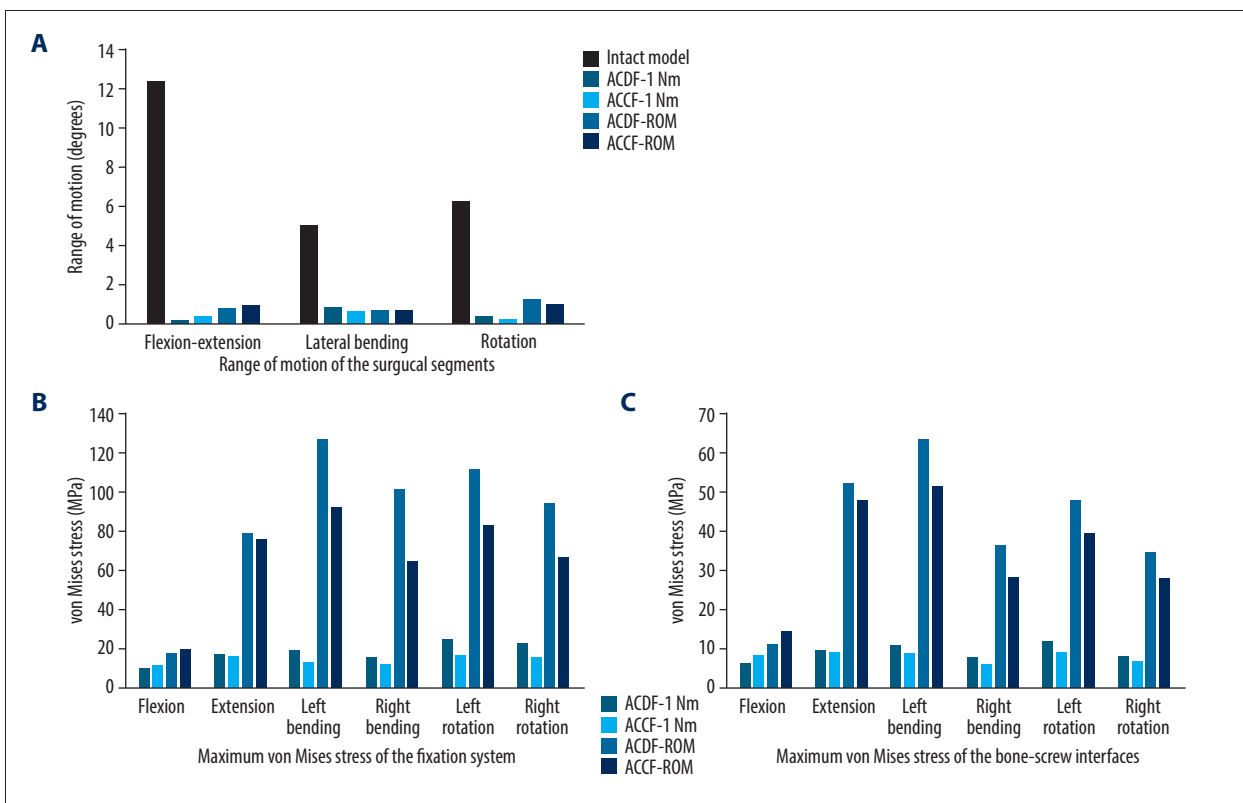


Figure 3. Parameters representing the postoperative spinal stability. (A) Range of motion of the surgical segments. (B) Maximum von Mises stresses of the fixation systems. (C) Maximum von Mises stresses of the bone-screw interfaces.

29.3%), left rotation (29.0%, 20.7%), and right rotation (18.2%, 24.3%); but were lower in flexion (30.7%, 26.7%).

ROMs of the adjacent segments

The ROMs of adjacent segments are shown in Figure 5A and 5B. Under the load of 1 Nm moment, ACDF and ACCF induced C3–C4 and C6–C7 segmental ROMs quite that were very similar to those of the intact model. The ROMs of C3–C4 segment of ACDF and ACCF induced by displacement-controlled loading were higher than that induced by 1 Nm moment in

flexion-extension (82.0%, 85.2%), lateral bending (56.8%, 50.6%), and rotation (45.5%, 13.2%). The ROMs of C6–C7 segments of ACDF and ACCF induced by displacement-controlled loading were also higher than that induced by 1 Nm moment in flexion-extension (100.7, 87.3%), lateral bending (32.7%, 40.0%), and rotation (118.1%, 106.4%). However, no differences in the ROMs of C3–C4 and C6–C7 segments was found between ACDF and ACCF.

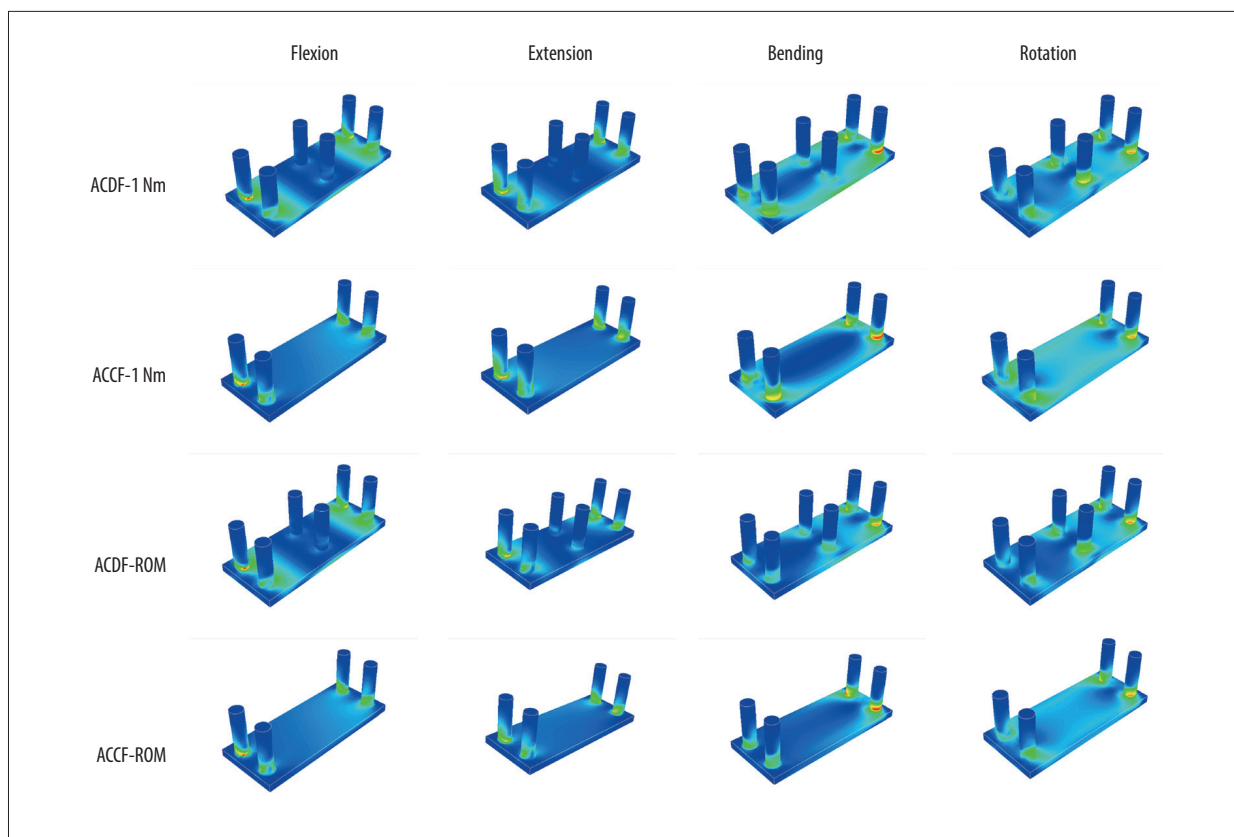


Figure 4. Stress distribution on the fixation systems. ACDF – anterior cervical discectomy and fusion; ACCF – anterior cervical corpectomy and fusion. 1 Nm – 1 Nm moment; ROM – displacement-controlled loading.

Adjacent intervertebral disc stresses

The maximum von Mises stresses of adjacent intervertebral discs are shown in Figure 5C and 5D. Under the load of 1 Nm moment, both ACDF and ACCF induced C3–C4 and C6–C7 intervertebral disc stresses that were quite similar to those of the intact model. The C3–C4 intervertebral disc stresses of ACDF and ACCF induced by displacement-controlled loading were much higher than those induced by 1 Nm moment in flexion (103.6%, 98.2%), extension (6.5%, 6.6%), left bending (282.2%, 298.9%), right bending (194.5%, 203.3), left rotation (120.2%, 133.9%), and right rotation (80.9%, 84.0%). The C6–C7 intervertebral disc stresses of ACDF and ACCF induced by displacement-controlled loading were also much higher than those induced by 1 Nm moment in flexion (102.6%, 100%), extension (211.6%, 211.6%), left bending (240.8%, 252.4%), right bending (183.5%, 200.8%), left rotation (159.4%, 168.8%), and right rotation (115.2%, 119.7%). The C3–C4 and C6–C7 intervertebral disc stresses showed no difference between ACDF and ACCF under any load condition.

Endplate stresses

The maximum von Mises stress of endplates of ACDF and ACCF are shown in Figure 6. The maximum von Mises stress of C4 inferior endplates of ACDF and ACCF induced by displacement-controlled loading were much higher than that induced by 1 Nm moment in flexion (73.6%, 70.6%), extension (371.5%, 184.3%), left bending (570.6%, 486.2%), right bending (370.1%, 321.7%), left rotation (333.5%, 334.9%), and right rotation (248.4%, 270.4%). The maximum von Mises stress of C6 superior endplates was also increased; the magnitude of increases was (84.6%, 79.2%), (524.8%, 461.6%), (775.4%, 419.0%), (580.3%, 524.6%), (317.3%, 347.6%), and (267.4%, 281.9%), in the same types of movement. The C4 inferior endplate stress peaks of ACCF were higher than that of ACDF in flexion (63.6%, 60.7%), extension (80.4%, 8.8%), left bending (43.5%, 25.5%), right bending (27.7%, 14.6%), left rotation (32.7%, 33.1%), and right rotation (27.2%, 35.2%) under 1 Nm moment and displacement-controlled loading. In C6 superior endplates, differences in von Mises stress were not evident. The distribution contour of von Mises stress of C4 inferior and C6 superior endplates are shown in Figure 7.

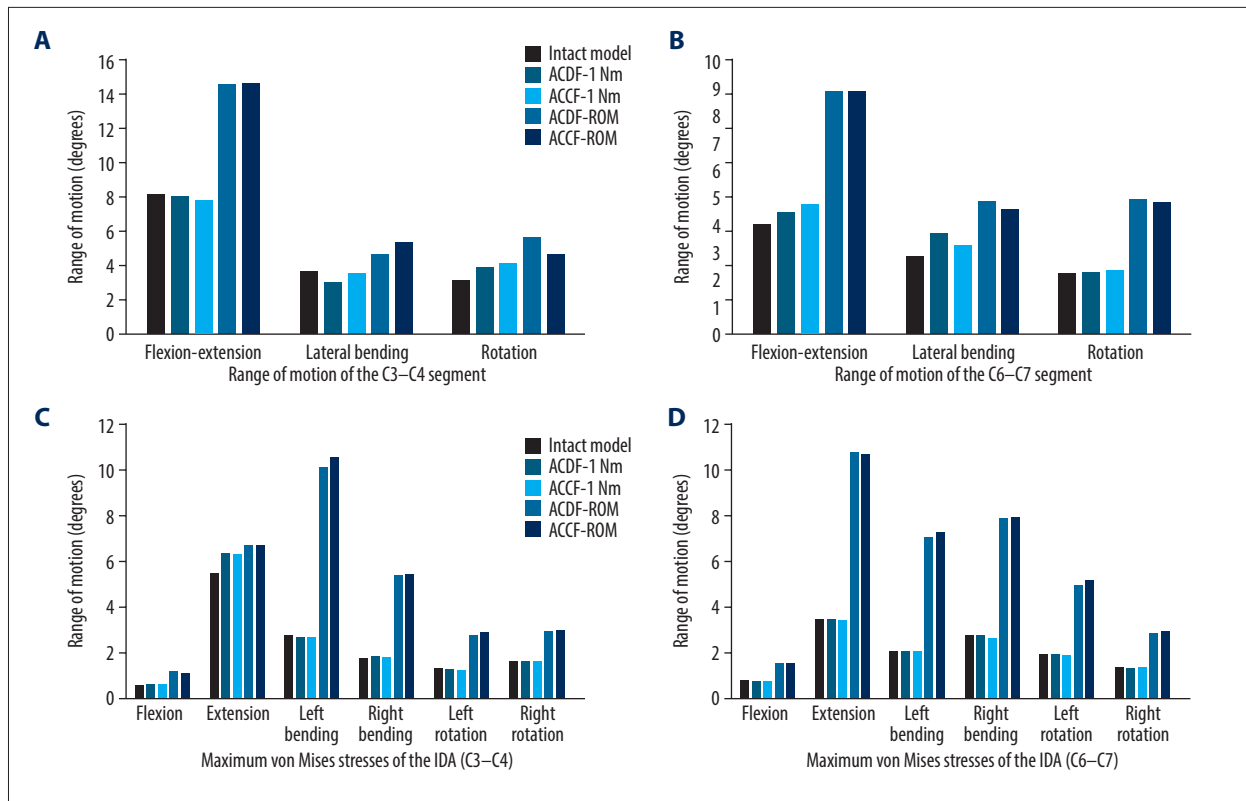


Figure 5. Parameters representing the degeneration of adjacent levels. (A) Range of motion of the C3-C4 segment. (B) Range of motion of the C6-C7 segment. (C) Maximum von Mises stresses of the C3-C4 intervertebral disc. (D) Maximum von Mises stresses of the C6-C7 intervertebral disc.

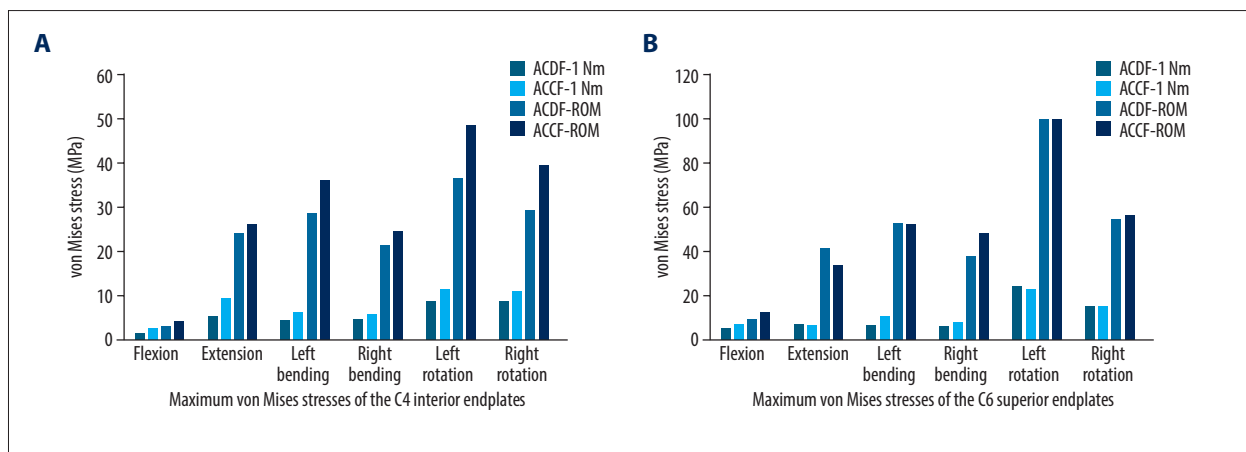


Figure 6. Maximum von Mises stresses of the endplates. (A) Maximum von Mises stresses of the C4 inferior endplates. (B) Maximum von Mises stresses of the C6 superior endplates.

Bone graft stresses

Figure 8 shows the maximum von Mises stresses in the bone grafts. The bone graft stress peaks of ACDF and ACCF induced by displacement-controlled loading were slightly higher than that induced by 1 Nm moment in flexion (73.9%, 61.9%), and much higher in extension (705.7%, 673.6%), left

bending (959.7%, 1008.2%), right bending (763.8%, 930.0%), left rotation (320.0%, 472.1%), and right rotation (266.4%, 449.8%). Under 1 Nm moment, the bone graft stresses in ACCF were higher than that in ACDF in flexion (82.6%), extension (51.4%), left bending (26.9%), right bending (37.9%), and right rotation (16.1%), but were slightly lower in left rotation (3.6%). When displacement-controlled loading was applied, the bone

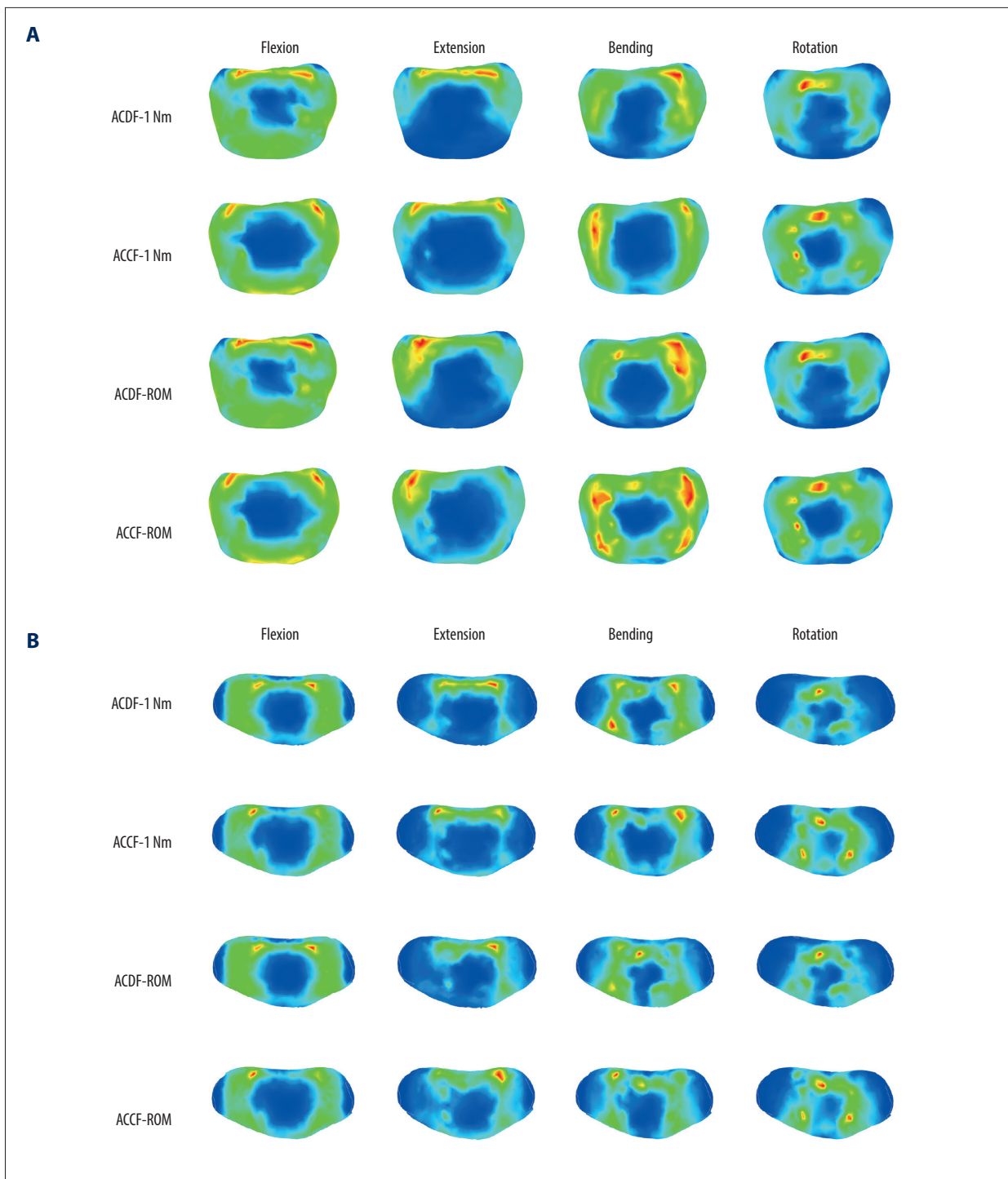


Figure 7. Stress distribution on the endplates. **(A)** Stress distribution on the C4 inferior endplates. **(B)** Stress distribution on the C6 superior endplates. ACDF – anterior cervical discectomy and fusion; ACCF – anterior cervical corpectomy and fusion. 1 Nm – 1 Nm moment; ROM – displacement-controlled loading.

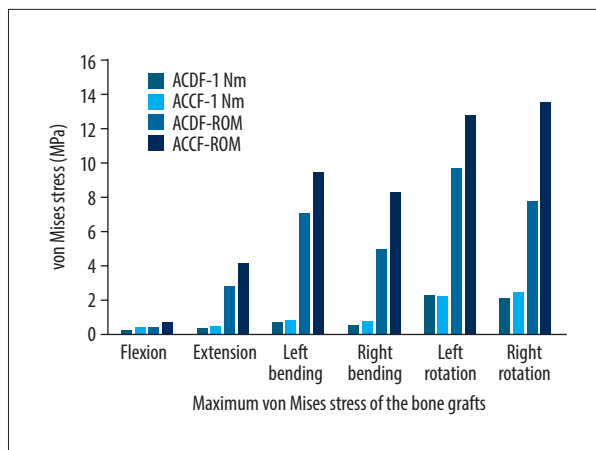


Figure 8. Maximum von Mises stresses of the bone grafts.

graft stresses in ACCF were higher than that in ACDF in flexion (70%), extension (45.4%), left bending (32.7%), right bending (64.5%), left rotation (31.5%), and right rotation (74.3%).

Discussion

Spinal stability

This study investigated the postoperative biomechanical stability provided by 2-level ACDF and 1-level ACCF. Results showed that both ACDF and ACCF surgery can clearly reduce the ROMs of surgical segments. The decreased magnitude ranged from 79.2% to 93.5% when the physiological range of motion was achieved, which was consistent with the results of previous studies [13,14]. The decrease in surgical segmental ROMs suggested that ACDF and ACCF can both provide satisfactory stability after surgery. No apparent difference in ROMs of surgical segments was found between ACDF and ACCF, which suggests that the ventral plate, not the cage or interbody cage, is the critical factor in maintaining the postoperative spinal stability. A possible explanation for this might be that the ventral plate has carried most of the load of the anterior column. Studies of Reidy and Darrel compared the cervical spine loading characteristics of ACCF with use of static and dynamic plates, finding that the cage carried very little load when the static plate was used [15,16].

Fixation system-related risks

The ACDF surgery significantly increased the stress peaks in the fixation system and bone-screw interface compared to ACCF surgery in most motion conditions. The reason for this difference may be that the reconstructed anterior column of ACCF construct was mainly composed of a titanium cage and that of ACDF consisted of 2 titanium interbody cages and a C5 vertebral body. The anterior column of the ACCF construct

has much higher stiffness and shared more of the load of the fixation system because of the high elastic modulus of the titanium cage [17,18]. The higher stress in the fixation system and bone-screw interface may weaken the fatigue resistance and long-term stability of the ACDF construct, which would increase the risks of non-fusion and pseudarthrosis. In a systematic review by Sheng, the non-fusion rates were 18.4% in 2-level ACDF and 5.1% in 1-level ACCF [8]. Furthermore, the extremely high stress in the fixation system and bone-screw interface may induce screw loosening and plate-screw breakage, which have also been reported in previous studies [19]. This result indicated that there still room for improvement of ACDF surgery. Further studies are required to improve the strength of the anterior cervical fixation system and the screw purchasing forces to alleviate the fixation system-related complications. It was noteworthy that the fixation system used in this study was static plate, which is the most widely used due to its advantages in reducing nonunion rates. However, the static plate may be too rigid to translate the load to the cage and graft, which may be the primary reason for the high stress in the fixation system and bone-screw interface [20]. The dynamic plate may be a potential solution to minimize the rate of fixation system loosening and breakage. As reported by previous studies, a dynamic plate allowed controlled settling and promoted load shearing across the cage and graft, thus reducing the stress on the fixation system and improving fusion [15,21]. However, the benefits of dynamic plates have been largely theoretical, and the actual clinical effects of dynamic plate use need further research.

Adjacent-level complications

The hypermobility and increased intradiscal pressures at adjacent levels induced by eliminating motion at the surgical segment were considered to be the cause of adjacent levels degenerative disease. In this study, the ROMs and intradiscal pressures of the adjacent levels in ACDF and ACCF induced by 1 Nm moment were very similar to those of the intact model, and were significantly increased when displacement-controlled loading was applied. A similar phenomenon was reported by Jason [22], who found that when the postoperative cervical spine was forced to reach the physiological motion range, the adjacent intradiscal pressures increased by 45.3% to 73.2% compared to the intact specimens. This result indicated that to maintain the physiological motion of the cervical spine after anterior cervical decompression and fusion surgery, the degenerative lesions of adjacent levels were inevitable. Van Eck investigated the rate of revision surgery and the occurrence of adjacent segment disease in patients undergoing ACDF for CSM and found that among all causes of revision surgery, the rate of adjacent segment disease was 47.5% [23]. No difference in the motions and intradiscal pressures at adjacent levels between ACDF and ACCF was found in the study.

Previous studies compared the clinical outcome of ACDF and ACCF in 2-level cervical spondylosis myelopathy and found no difference in adjacent-level ossification and complications [24], suggesting that neither ACDF nor ACCF perform better in the degenerative changes in the adjacent levels, and other techniques, such as use of an anterior non-fusion fixation device, should be considered to address this problem [25].

Cage subsidence

A migration of the cage or interbody cage into the adjacent vertebral body was defined as cage subsidence, which has been commonly reported in ACDF and ACCF [26,27]. The occurrence of cage subsidence was considered to be closely correlated with endplate conditions such as the degree of osteoporosis and loaded stress [28]. The removal of the endplate in ACDF was found to increase the rate of cage subsidence from 18% to 58% [26]. In the present study, ACCF induced much higher stress on the C4 inferior endplates and a similar stress on the C6 superior endplates compared to ACDF, which indicated that ACCF was more vulnerable to cage subsidence than ACDF. This result is consistent with that of previous clinical studies [8,13,29]. The mean rate of cage subsidence after ACDF was reported as 21.2% [30], while in ACCF, cage subsidence occurred in 80% of patients [31]. Therefore, it is necessary to consider the potential risk of cage subsidence and related complications such as kyphosis and severe implant failure when ACCF is used. As mentioned earlier, the higher endplate stress of ACCF compared to ACDF may be caused by the difference in the reconstructed anterior column. The reconstructed anterior column of ACCF comprised only a titanium cage while that of ACDF was composed of titanium cages and the C5 vertebral body. The presence of the C5 vertebral body reduced the total elastic modulus of the reconstructed anterior column, and thus decreased the load sharing and the endplate stress. Attention should be paid to optimal cage design to minimize the stress concentration on the endplate to improve the clinical outcome of ACCF.

Bone graft stress

The spinal stability provided by a fixation system is temporary; permanent spinal stability depends on reliable bone fusion after anterior cervical decompression and fusion surgery [32]. Solid bone fusion is the ultimate goal of both ACDF and ACCF. According to the Wolff law [33], the stress distribution on the bone graft is a critical factor that affects bone fusion. An appropriate level of stress is necessary for bone formation, and the lack of stress induces bone resorption. In the present study, we found that the stresses on the bone graft in ACCF were significantly higher than in ACDF, which indicated that ACCF might be superior to ACDF in fusion. This conclusion is consistent with previous studies. In a systematic review

study, Jiang analyzed the published literature ranged from Jan 1969 to Dec 2010 to compare the fusion rate of ACCF and ACDF, finding that the fusion rates of ACCF were much higher than those of ACDF, especially for multilevel cervical spondylosis [6]. Some researchers attributed the inferiority of ACDF in fusion to the increased number of fusion surfaces [34]; this may be a potential cause, but it is impossible to correct after the ACDF surgery method is chosen. The lower stress on the graft found in the study indicated that optimizing the stress distribution on the graft promise is an effective method to improve the fusion rate of ACDF. Goldberg used a dynamic plate as its load sharing property to enhance the stress on the graft to promote fusion, reporting that the cervical fusion with dynamic plating was slightly higher than that with static plating [35]. However, other studies found a higher rate of non-union in dynamically plated patients than in statically plated ones [36]. These conflicting results might be due to the different designs of the dynamic plate. A dynamic plate can enhance stress on the graft to promote fusion, but also induces micromotion between the endplate and graft, thus leading to pseudarthrosis. Future studies should focus on optimizing the structure of the dynamic plate or designing other fixating systems to simultaneously enhance graft stress and stabilize the contact between the endplate and the graft.

Limitations

Our study has several limitations. To reduce the computation load, the morphology of the fixation system, cage and interbody cage, and bone graft have been simplified and these simplifications inevitably affected the actual stress distribution. In addition, the FE models used in the present study were constructed using the computational tomography images obtained from a healthy person, so the results derived from this study may be not applicable to cases with severe pathological changes such as kyphosis and severe osteoporosis. Further studies are required to expand the range of application of models to completely clarify the biomechanical characteristics of ACDF and ACCF.

Conclusions

Both ACDF and ACCF can provide satisfactory stability for the postoperative low cervical spine. The ACDF approach induced less stress on the endplates, which can decrease the risk of cage subsidence, while the ACCF approach induced less stress on the fixation system and bone-screw interfaces and higher stress on the bone graft, which can benefit long-term spinal stability and bone fusion. Based on these results, corresponding measures should be implemented to improve the clinical outcome of ACDF and ACCF.

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