External and internal responses of cervical artificial disc replacement, and anterior cervical discectomy and fusion: A finite element modeling study

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ABSTRACT

Surgical treatment for spinal disorders, such as cervical disc herniation and spondylosis, includes the removal of the intervertebral disc and replacement of biological or artificial materials. In the former case, bone graft is used to fill the space, and this conventional procedure is termed anterior cervical discectomy and fusion (ACDF). The latter surgery is termed as artificial disc arthroplasty (ADR). Surgeries are most commonly performed at one or two levels. The present study was designed to determine the external (range of motion, ROM) and internal (anterior and posterior load sharing) responses of the spines with one-level and two-level surgeries in both models (ACDF and ADR) using a previously validated finite element model (FEM) of the subaxial cervical spinal column. The FEM simulated the vertebra (cancellous core and cortical shell of the body, posterior elements – laminae, pedicles and spinous processes), discs (anulus fibers, ground substance, and nucleus pulposus), anterior and posterior ligaments of the disc and facet joints, and interspinous and supraspinous ligaments. Appropriate material properties were assigned to the spinal components. The United States Federal Drug Administration-approved Mobi-C was used for the ADR option. The FEM was exercised under pure flexion and extension moment loading of 2 Nm in the intact state. The overall ROM of the column was obtained. The hybrid loading protocol applied moments that matched the ROM in the intact spine for both one-level (C5-C6) and two-level (C5-C7) ACDF and ADR surgeries. ROM at the level(s) of surgery, termed the index level was obtained. These data along with anterior column load (ACL) and posterior column load (PCL) sharing were obtained for all surgical options at superior and inferior segments (termed adjacent segment outputs). Results for both one-level and two-level surgeries showed that ACDFs decreases ROM at the index level, while ADRs increase motions compared to the intact normal spine. The ROM, ACL, and PCL increased at both adjacent levels for the ACDF while ADR showed a decrease. Although two-level surgeries resulted in increased these biomechanical variables, greater changes to adjacent segment biomechanics in ACDF may accelerate adjacent segment disease. Decreased ROM and lower load sharing in ADRs may limit adjacent segment effects such as accelerated degeneration. Their increased posterior load sharing, however, may need additional attention for patients with suspected facet joint disease.

KEYWORDS: Artificial disc; cervical spine; finite element model; range of motion; load sharing; segmental biomechanics

INTRODUCTION

The disc and bilateral facet joints of the spine form the central path for the transmission of the external and day-to-day physiologic loads along the vertebral column (Clark and Benzel, 2005). (The hard tissue components of the cervical spine are not generally susceptible to bone loss such as osteopenia and osteoporosis); however, joints are prone to anatomical and structural changes due to their load bearing function (Yoganandan et al., 2006a; Yoganandan et al., 2006b). Load sharing occurs anteriorly through the disc and posteriorly through the facet joints. Aging, the normal degenerative process, and occupational exposures such as those in the military and certain civilian environments may lead to neck pain, cervical spondylosis, and instability (Ang and Harms-Ringdahl, 2006; Burton and Travis, 1999). Spinal disorders such as disc herniation and spondylosis are initially treated conservatively, i.e., without surgical intervention. In patients with pathologic changes and persisting symptomatology who have failed medical management, surgical intervention is the typical treatment modality. Removing the disc (discectomy) using an anterior approach is common procedure (Bailey and Badgley, 1960; Hilibrand et al., 1999; Smith and Robinson, 1958). It is still in use, albeit in modified forms (Bishop et al., 1996).

The anterior cervical discectomy approach calls for replacing the excised disc with a graft (typically autograft, allograft and/or PEEK), that acts as a fusion material, and the principal intent is to permanently constrain the motion at the pathological level (Cho et al., 2002; Fujibayashi et al., 2008). This procedure is called anterior cervical discectomy and fusion, ACDF. While these surgeries are often done at one-level, multiple level surgeries are also performed. Anterior cervical discectomy and fusions are considered the gold standard and have been successful throughout the years; however, loss of range of motion (ROM) and development of adjacent segment disease is a significant concern (Hilibrand and Robbins, 2004). This adjacent segment disease is attributed to increased motion at the adjacent segments, as the constrained motions gets transferred to the adjacent levels. This has resulted in the development of motion preserving techniques such as cervical disc arthroplasty.

In efforts to preserve range of motion and to minimize adjacent segment disease, the United States Federal Drug Administration (US FDA) has approved artificial disc arthroplasty (ADR) as an alternative to conventional ACDFs (Alvin et al., 2014; Choi et al., 2019b). The removed disc is replaced with an artificial disc and no bone graft material is used in the process. The intent is to preserve, instead of constrain motion at the pathologic level, i.e., index level, thereby, limiting the transmission of additional biomechanical stress/strain to adjacent vertebral levels. While many clinical studies are published proving the non-inferiority hypothesis of the ADRs (compared to conventional ACDFs),

there remains some controversy. The external and internal biomechanical responses of normal or intact, ACDF and ADR spines with one-level and two-level surgeries have not be carried out for the Mobi-C type device. The hypothesis is that the ADR preserves motion at the index level and reduces column loads. Based on this hypothesis, the purpose of the study is, therefore, to determine the external responses, i.e., ROM at the index level and adjacent vertebral segments, and internal responses, i.e., anterior load sharing as defined by the intradiscal pressure and posterior load sharing as defined by the facet forces, under sagittal moments using a validated spinal column finite element model.

METHODS

The current work adopted a previously validated and geometrically accurate C2-T1 osteoligamentous finite element model (FEM) of the cervical spinal column (Choi et al., 2019a; John et al., 2017). The FEM simulated the vertebra (cancellous core and cortical shell of the body, posterior elements – laminae, pedicles and spinous processes, discs – anulus fibers, ground substance, and nucleus pulposus, anterior and posterior ligaments of the disc and facet joints, ligamentum flavum, and interspinous ligaments. The types of elements and material properties assigned to these components are listed in Table 1. The artificial cervical disc Mobi-C® was the ADR (Lu and Peng, 2017). It was modeled in computer aided design software (CATIA V6, Dassault systems) and meshed in ANSA software, version 16.1.0 (BETA CAE Systems, Farmington Hills, MI). The LSDYNA (Livermore Software Technology Corporation, Livermore, CA, R9.0.1) solver was used in the study.

Table 1: Details of the FEM components

Component	Element Type	Constitutive Model	Parameters
Cortical bone	Quadrilateral shell	Linear elastic	E = 16.8 GPa, μ = 0.3
Trabecular bone	Hexahedral solid	Linear elastic	E = 400 MPa, μ = 0.3
Endplate	Quadrilateral shell	Linear elastic	E = 5.6 GPa, μ = 0.3
Facet cartilage	Quadrilateral shell	Linear elastic	E = 10 MPa, μ = 0.3
Annulus ground substance	Hexahedral solid	Hill foam	n = 2, C1 = 0.115 MPa C2 = 2.101 MPa, C3 = -0.893 MPa b1 = 4, b2 = -1, b1 = -2
Annulus fibrosus	Quadrilateral membrane	Orthotropic nonlinear elastic	Fiber angle (45-60°)
Nucleus pulposus	Hexahedral solid	Fluid	K = 1720 MPa
Ligaments	Quadrilateral membrane	Non-linear curves	Stress-strain curves-
Upper Plate	Hexahedral solid	Linear elastic	E = 210 MPa, μ = 0.3
Middle Core	Hexahedral solid	Linear elastic	E = 3 MPa, μ = 0.3
Lower Plate	Hexahedral solid	Linear elastic	E = 210 MPa, μ = 0.3

The standard surgical procedure was used to simulate ADR and ACDF at the C5-C6 level. The anterior longitudinal ligament was removed at the surgical level in both cases. In the case of ADR, a cavity was created at the implanted level for the placement of the disc prosthesis. Both the superior and inferior components of the ADRs were attached to the respective vertebral bodies using tied contact to simulate complete osteointegration of the implant with the bone to ensure no relative motion between the implant and vertebral endplates. The contact between the metal and polymer surfaces were modeled as a surface-to-surface contact definition with a coefficient of friction of 0.1. In the case of ACDF, the disc properties were altered at the surgical level to simulate the trabecular bone, and this represented complete fusion (Mackiewicz et al., 2016). In case of one level surgery, the ACDF and ADR were simulated at C5-C6 level, and in the case of two-level surgery, they were simulated at the C5-C6 and C6-C7 levels (Figure 1). It should be noted that there are the two widely operated levels for ADR (Gornet et al., 2017).

Mobi C is a three-piece metal-polymer-metal design with unconstrained center of rotation. The polymer core is made of ultrahigh molecular weight polyethylene and translates approximately 1 mm in anteroposterior and medial-lateral directions until it is stopped by two metal supports. This contrasts with other implant designs. For example, the Prestige LP disc is a metal on metal with a fixed center of rotation while the Bryan disc is a one-piece closed compartment with an unconstrained design (Alvin et al., 2014; Choi et al., 2019b; Kani and Chew, 2019). A coordinate Measuring Machine (FaroArm, Irvine, CA) was used to define the geometric representation of objects by means of accurate dense clouds of three-dimensional points from the external surface of the Mobi-C disc prosthesis. The generated three-dimensional points were imported into a software (CATIA V6, Dassault systems, Waltham, MA) and processed with a digitized shape editor to create the surface geometry from the points and meshed in the ANSA software (BETA CAE Systems, Farmington Hills, MI). Table 1 shows the details of the components defined for the ADR.

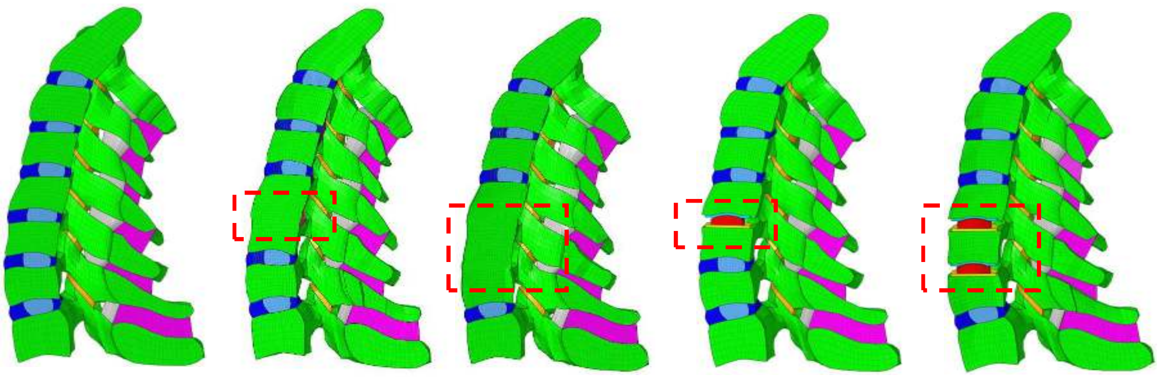

Figure 1: Intact (left), ACDF (one- and two-level, second and third from left) and ADR (one- and two-level, fourth and fifth from left) subaxial spinal column FEMs.

The model was fixed at T1 vertebra for all degrees of freedom and the moment load was applied at the rostral-most vertebra. Flexion and extension moments were simulated under two loading conditions: pure moment (2 Nm) and hybrid protocol. The hybrid loading protocol was used to understand the changes in the biomechanics of spine after implanting with the ACDF and ADR at C5-C6 and C5-C7 levels. In this method, the applied moment was varied until the overall column range of motion (ROM) of the operated spine (both ACDF and ADR) reached the magnitude determined in the intact spine under the pure moment loading. In all the three FEMs, biomechanical data such as the overall/total and segmental ROM, and anterior and posterior column load-sharing as determined by the intradiscal pressure and facet forces were obtained. These data were used to compare the changes in the external and internal biomechanical responses of cervical spine following one-level and two-level ACDFs and ADRs.

RESULTS

Responses under Pure moment

Figure 2 compares the flexion ROM responses at each segment of the subaxial spine for the intact spine and surgically altered spines under flexion. C5-C6 and C6-C7 responded with the greatest motion (10.7 deg and 10.2 deg), while the C2-C3 and C7-T1 responded with the lowest ROM (8.0 deg and 8.7 deg). A similar pattern was observed for the ADR spines for both one-level and two-level surgeries. For the ACDF spines, the ROMs were less than one degree at the index levels for both one-level and two-level surgeries. Other data are shown in the cited figure.

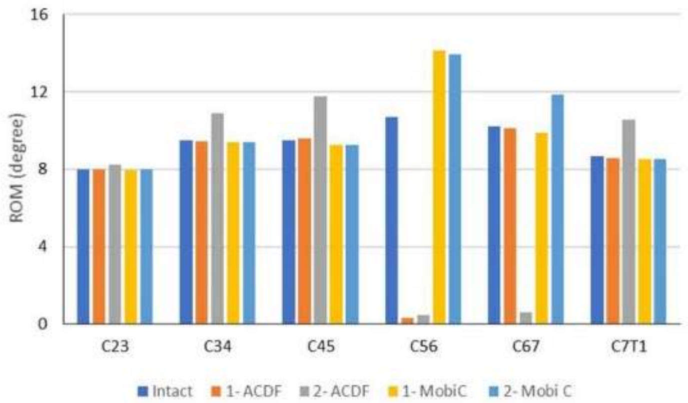

Figure 2: ROM under pure moment loading for intact and surgically altered spines under flexion.

Figure 3 compares the extension ROM responses at each segment of the subaxial spine for the intact spine under extension. C5-C6 responded with the greatest motion of 6.0 deg, and C6-C7 responded with the least motion, 4.3 deg. For the ACDF spines, the ROMs were less than one degree at the index levels for both one-level and two-level surgeries. For the one-level ADR spine, the ROM was the least at the index segment and somewhat uniform across all other segments (4 to 5 deg). For the two-level ADR, the superior index level responded with the greatest motion of 8 degrees, followed by the inferior index level at 5 deg. Other data are shown in the cited figure.

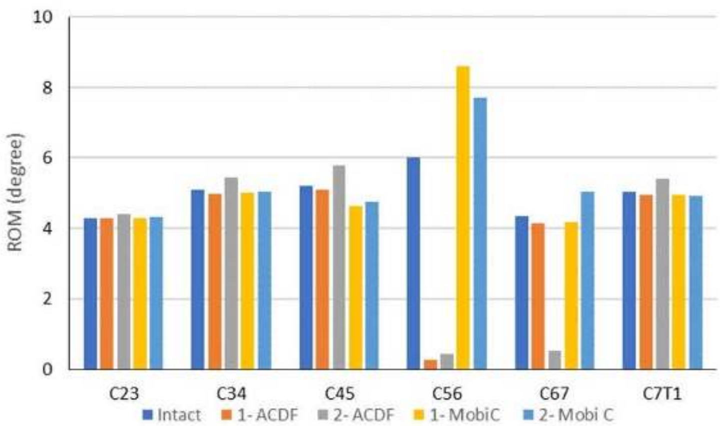

Figure 3: ROM under pure moment loading for intact and surgically altered spines under extension.

Responses under Hybrid loading protocol

Range of motion: Moments required to achieve the overall ROM of the intact spine for the one- and two-level ADCF and ADRs in flexion were 3.5 Nm, 4.5 Nm, 1.95 Nm, and 1.98 Nm, and in extension were 4.0 Nm, 6.0 Nm, 1.4 Nm, and 1.5 Nm, respectively. Figures 4 and 5 compare the magnitudes of ROM at the superior and inferior segments for the one and two level ACDF and ADR surgeries. It should be noted that the inferior segments do not correspond to the same segment: in the case of one-level surgery, the superior and inferior segments are C4-C5 and C6-7, while in the case of the two-level surgical option, they are C4-C5 and C7-T1, respectively. Compared to the intact spine, under flexion, the ACDF option resulted in increased motions of 32% and 29% for the one-level option and 44% and 42% for the two-level option (Figure 4); and under extension, these magnitudes were 28% and 18% for the one-level option and 53% and 18% for the two-level option (Figure 5). In contrast, the ADR surgeries decreased ROMs at both segments for both one- and two-level options. Under flexion, the decreases were 7% and 13% at the superior segment and 12% and 9% at the inferior segment. Under extension, they were 17% and 14% at the superior segment, and 7% and 2% at the inferior segment for the two-level surgery.

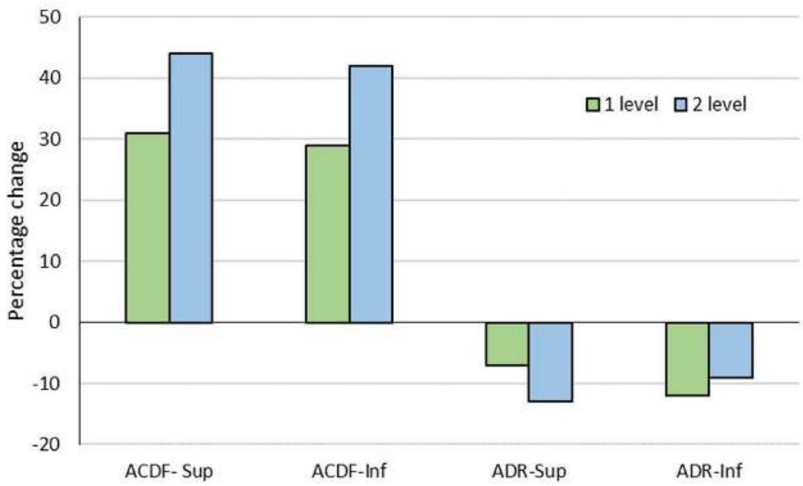

Figure 4: % change in ROM at the adjacent levels for the one- and two-level surgeries with respect to the intact spine under hybrid loading protocol under flexion.

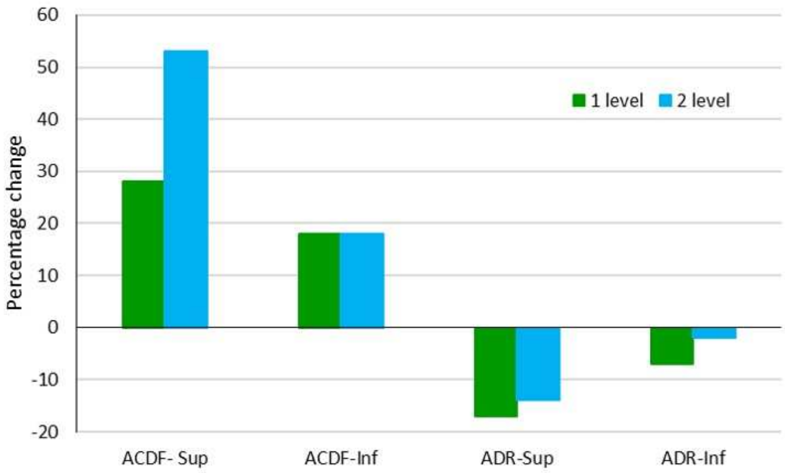

Figure 5: % change in ROM at the adjacent levels for the one- and two-level surgeries with respect to the intact spine under hybrid loading protocol under extension.

Anterior load sharing: Figures 6 and 7 compare the magnitudes of the anterior column load sharing at the superior and inferior segments for the one and two level ACDF and ADR surgeries. Compared to the intact spine, under flexion, the ACDF option resulted in increased motions of 56% and 27% for the one-level option, and 64% and 45% for the two-level option (Figure 6); and under extension, these magnitudes were 71% and 22% for the one-level option, and 93% and 31% for the two-level option (Figure 7). In contrast, the ADR produced decreased load sharing. Under flexion, the decreases were 4% and 14% at the superior segment, and 14% and 16% at the inferior segment. Under extension, they were 22% and 21% at the superior segment for the one-level surgery, and 32% and 8% at the inferior segment for the two-level surgery.

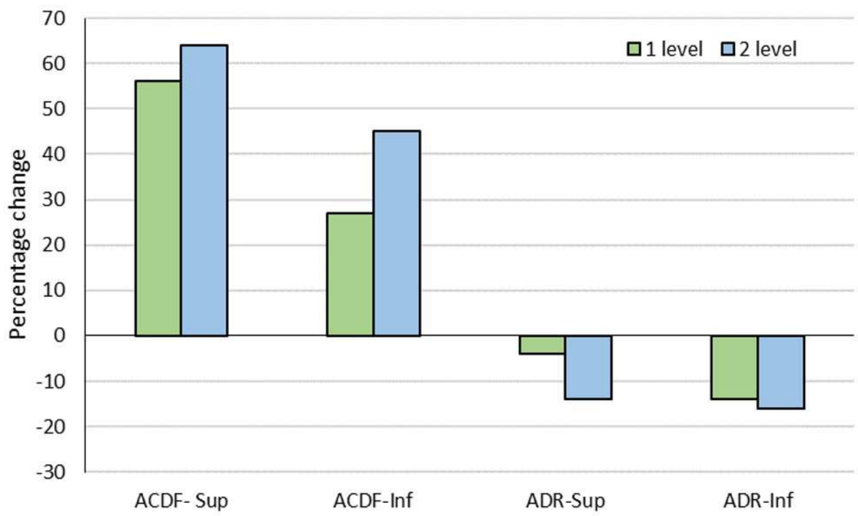

Figure 6: % change in ACL at the adjacent levels for the one- and two-level surgeries with respect to the intact spine under hybrid loading protocol under flexion.

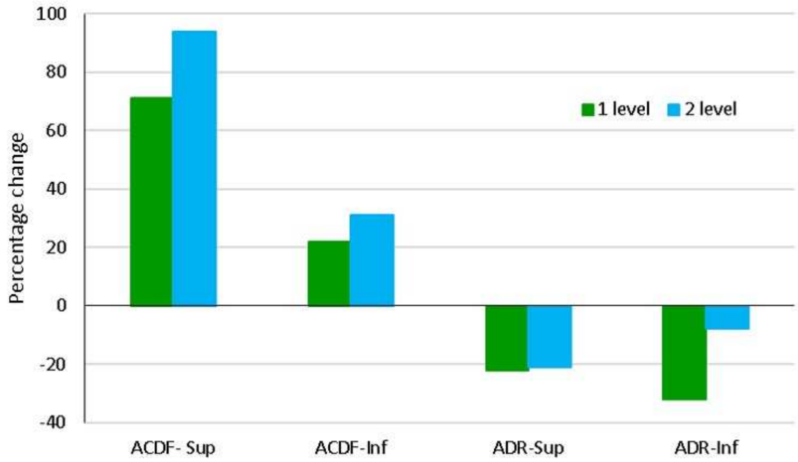

Figure 7: % change in ACL at the adjacent levels for the one- and two-level surgeries with respect to the intact spine under hybrid loading protocol under extension.

Posterior load sharing: Figure 8 compares the magnitudes of PCL at the superior and inferior segments for the one and two level ACDF and ADR surgeries. Compared to the intact spine, the ACDF option resulted in increases of 27% and 34% for one-level option, and 64% and 87% for the two-level option. In contrast, the ADR-produced decreases were 16% and 8% at the superior segment and 17% and 16% at the inferior segment.

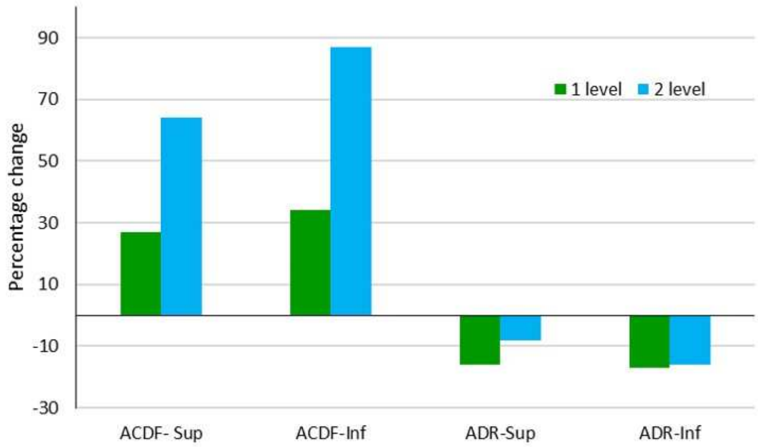

Figure 8: % change in PCL at the adjacent levels fpr the one- and two-level surgeries with respect to the intact spine under hybrid loading protocol under flexion.

Comparative evaluations with one-level surgery

Range of motion: At the superior segment, in flexion and extension, the two-level surgery resulted in increased ROM by 1.4 times and 1.9 times for the ACDF and by 1.9 times and 0.8 times for the ADR surgeries. At the inferior segment, these values were 1.5 times and 1.0 times for the ACDF and 0.8 times and 0.3 times for the ADR surgeries, respectively (Figure 9).

Anterior load sharing: At the superior segment, in flexion and extension, the two-level surgery resulted in increased ACL by 1.1 times and 1.3 times for the ACDF and by 3.1 times and 0.98 times for the ADR surgeries. At the inferior segment, these values were 1.7 times and 1.4 times for the ACDF and 1.2 times and 0.26 for the ADR surgeries, respectively (Figure 10). The actual magnitudes of anterior column loads are shown (Table 2).

Table 2: Magnitudes of anterior column load (MPa) s for different conditions of the spine.

Loading	Sup	Inf	Sup	Inf	Sup	Inf
	Intact		ACDF1		ADR1	
Flexion	0.36	0.27	0.56	0.35	0.34	0.24
Extension	0.25	0.06	0.42	0.08	0.19	0.04
Loading	Intact		ACDF2		ADR2	
Flexion	0.36	0.70	0.59	1.02	0.31	0.59
Extension	0.25	0.24	0.48	0.31	0.20	0.22

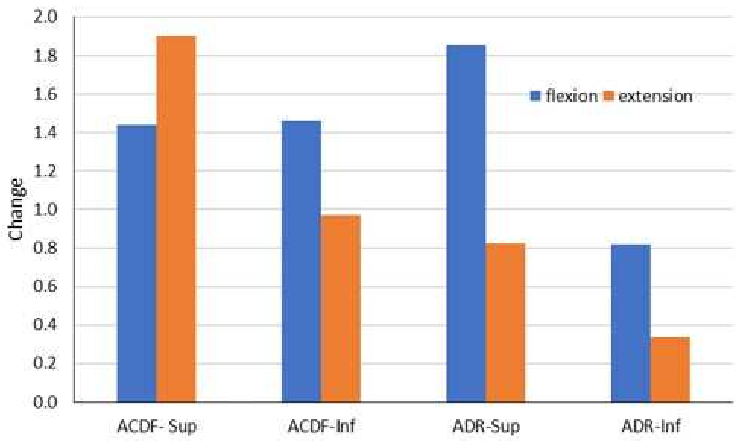

Figure 9: Change in ROM at the adjacent levels for the two-level surgeries with respect to the one-level surgery under flexion and extension for the hybrid loading protocol.

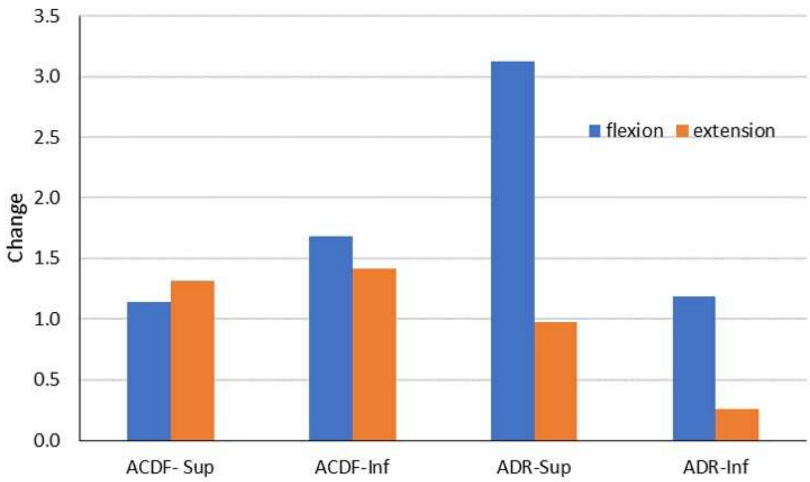

Figure 10: Change in ACL at the adjacent levels for the two-level surgeries with respect to the one-level surgery under flexion and extension for the hybrid loading protocol.

Posterior load sharing: At the superior segment, the two-level surgery resulted in increased PCL by 2.3 times for the ACDF and by 0.5 times for the ADR surgeries. These values at the inferior segment were 2.6 and 1.0 times, respectively, for the ACDF and ADR surgeries (Figure 11). The magnitudes of the loads for different conditions are given (Table 3)

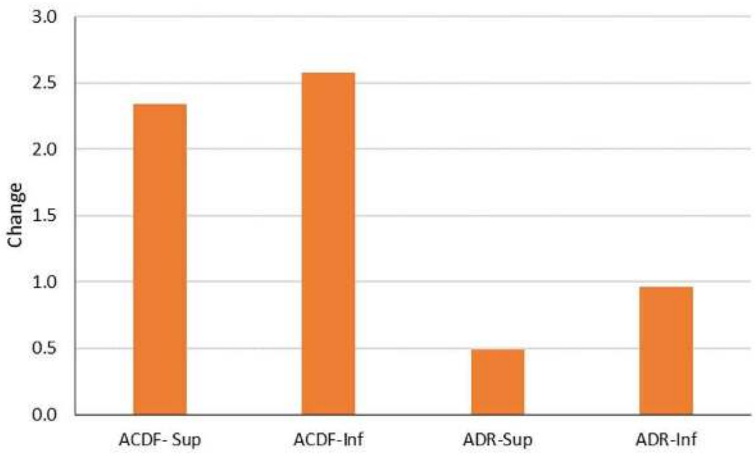

Figure 11: Change in PCL at the adjacent levels for two-level surgeries with respect to the one-level surgery for the hybrid loading protocol.

Table 3: Magnitudes of posterior column loads (N) for different conditions of the spine.

Sup	Index	Inf	Sup	Index	Inf	Sup	Index	Inf
Intact			ACDF1			ADR1		
30.0	20.8	45.7	38.1	11.8	61.1	25.1	66.3	38.0
Intact			ACDF2			ADR2		

30.0	33.3	28.2	49.1	12.5	52.7	27.6	56.3	23.6

Index level biomechanics

Figure 12 compares the changes in ROM responses for both surgeries in both modes. The ACDF surgery decreased motions for both ACDF surgeries, while ADR showed increased motions for both one-level and two-level surgeries. The motions shown in the figures for the two-level surgeries are an average of the two index levels. Figure 13 shows the posterior column loads at the index level for both ACDF and ADR in extension.

DISCUSSION

As stated in the introduction, the objective of the study was to determine the external and internal biomechanical responses of two types anterior cervical spine surgeries: anterior cervical fusion and cervical disc arthroplasty. While external responses (ROM) can be obtained from experiments using human cadavers (post mortem human surrogate specimens, PMHS), internal responses are difficult to measure from biological models (Kumaresan et al., 1999). For example, loads in the posterior column (facet forces) cannot be measured without violating the integrity of the joint itself, which when altered for placing a transducer such as load cell changes the load path of the spine, with or without ACDF and ADR. It is difficult to perform multiple surgeries on the same spine with different procedures in a PMHS experimental model while monitoring internal loads and motions. Computational models such as the finite model used in the present study is an effective tool for such evaluations (Maiman et al., 1999; Yoganandan et al., 1996; Yoganandan et al., 1987). Hence, a previously validated FEM was used in to achieve the stated objectives (Choi et al., 2019a).

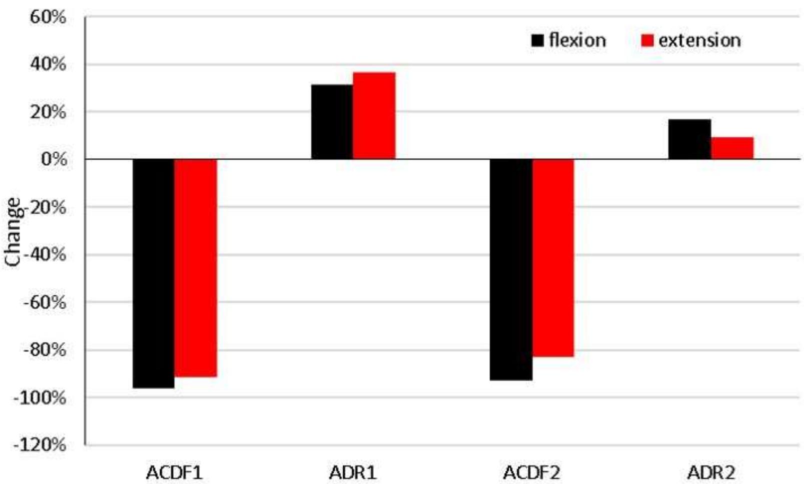

Figure 12: Change in the index level ROM for one-level and two-level surgeries under flexion and extension

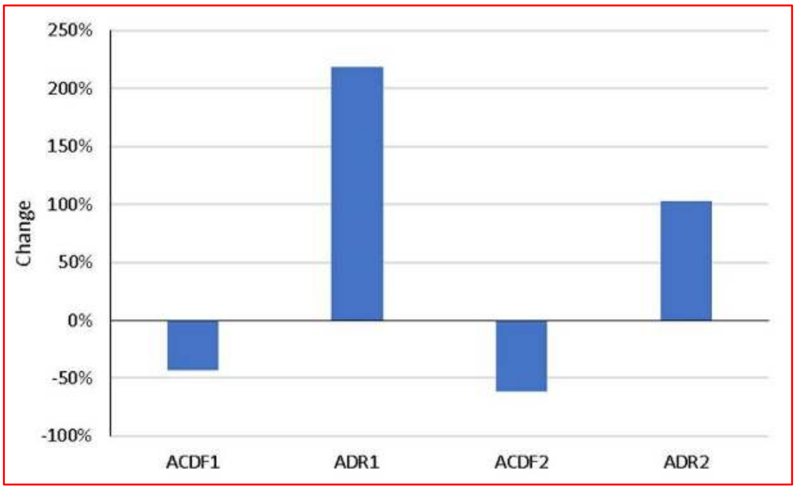

Figure 13: Change in the index level PCL for one-level and two-level surgeries under extension

Comparison with literature

A comparison of the motion obtained from the present study made with existing experimental studies (Figure 14). An earlier study 15 human cadaver spines to 1.0 Nm of flexion-extension loading (Panjabi et al., 2001). Another study subjected 13 specimens to 2 Nm of loading and both studies reported segmental motions (Wheeldon et al., 2006). Motions from the present study were within the mean ± one standard deviation compared with these results.

Different protocols have been used to apply biomechanical loads to intact and instrumented (ACDF and ADR) spines, and it is true for PMHS and FEM models. One PMHS study tested 12 C2-T2 spines by fixating the ends and applying a pure moment of 2 Nm, with and without a constant compressive force [CCF, 50 N], called the follower load (Barrey et al., 2012). The constructs were one-level (C5-C6) and two-level (C4-C6) arthrodesis and ADR. The ADR was a ball-and-socket device and arthrodesis was an interbody cage with plating. The instrumented spines were evaluated at the magnitude of the overall range of motion (OROM) corresponding to the OROM of the stiffest condition in flexion, termed the hybrid loading protocol. Another PMHS study using nine C3-T1 PMHS spines tested a different type of ADR, with and without interbody cage and plating at the C5-C6 level (Colle et al., 2013). For all conditions, 1.5 Nm pure moment loading was applied first, followed by loading with a compression-flexion device that applied a CCF of 70 N. In the second loading case, motion was limited to the OROM of the C3-T1 spine, achieved during the pure moment case. In another PMHS study, six intact C3-C7 columns were subjected to 1.5 Nm of moment and 100 N compressive force (Finn et al., 2009). The spines with ADR and fusion (securing rods to lateral mass screws and anterior plate) were subjected to the same compressive force and differing moments to achieve the same OROM as in the intact spine.

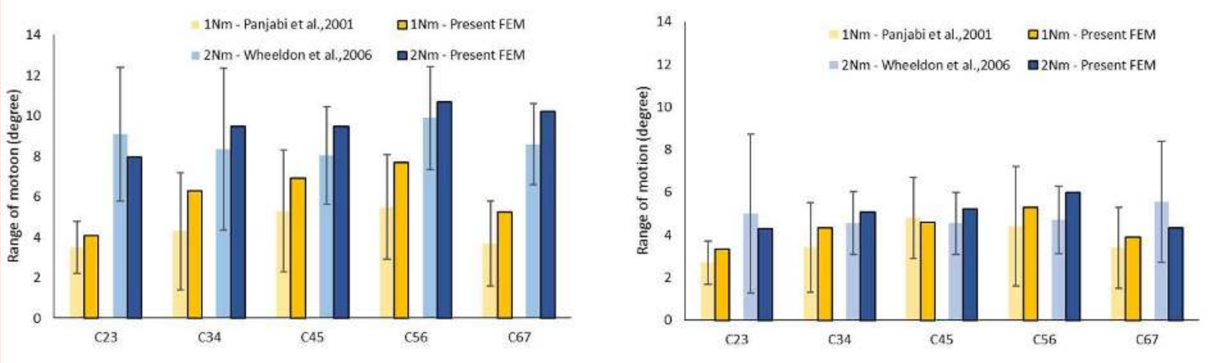

Figure 14: Comparison present results with literature (left: flexion, right: extension)

A similar paradigm is apparent in modeling studies. Using a C5-C6 functional unit FEM, compared to the intact segment, the ROM decreased with the ball and socket ADR under a CCF of 40 N and a moment of 1.6 Nm (Rousseau et al., 2008). A later study, to evaluate the ROM and facet forces with different ADRs, used a C2-C7 FEM and applied a CCF of 50 N and a moment of 1 Nm for both intact and ADR spines (Lee et al., 2011). In another study, a C3-C7 FEM was used to evaluate different ADR designs at the C5-C6 level, and the loading protocol included a CCF of 75 N and a moment of 1.5 Nm for the intact spine (Faizan et al., 2012). For the analysis of ADR models, the loading was such that the moments were limited corresponding to the OROM found in the intact spine while the applied force remained at the same magnitude. A more recent study used yet another C3-C7 FEM to evaluate the effects of different types of single level ADRs at the C4-C5 level and arthrodesis at the C5-C6 level by applying a CCF of 73.6 Nm and a moment of 1.0 Nm for the intact spine (Mo et al., 2017). The surgically altered spines were subjected to the same force but with a moment that limited the OROM to the level found in the intact spine.

These previous PMHS and FEM studies, although not all inclusive, encompassed different spinal levels, used various types of ADRs and conventional fusion types at different segments, and applied differing force and moment magnitudes, while their goals were in unison: evaluate the biomechanics of constructs; the actual comparison is difficult due to differences in study designs. The present study falls in line with these studies: it used a column FEM, simulated ADRs and ACDFs at one and two segments, and obtained the OROMs and adjacent segment motions and load-sharing patterns, under flexion and extension moments. It should be noted that while the OROMs for the instrumented spinal column were matched with the intact column for evaluating surgically altered spines, the actual motion at the surgical level(s) cannot be matched using the hybrid protocol method. From this perspective, it may be prudent to also analyze the effects of ADR and ACDF using the pure moment loading.

Data from Figures 2 and 3 shows that, for ACDF, the motion at the adjacent level increases by 0.8% at the superior level and decreases by 0.9% at the inferior level in flexion and decreases by 2.3% and 4.6% at the superior and inferior levels in extension. In contrast, the increase in magnitudes of adjacent segment motion, as shown in Figure

4 and 5 were greater (18-31% range). It should be noted that this discrepancy is due to the level of moment: Figures 2 and 3 are from pure and Figures 4 and 5 are from Hybrid moments.

Clinical implications – Index level, ACDF and ADR

The similar decrease in ROMs at the index level(s) for both one and two level ACDFs in flexion (93% to 98%), extension (83% to 92%), and posterior column load sharing (43 to 62%) suggest that the conventional options substantially rigidize the spine, the original intent of this procedure. In contrast, the ADR procedures resulted in opposite responses: increased motions in both modes (32% and 17% for one and two levels for flexion and 37% and 9% for extension). Posterior column loads, however, increased due to ADRs with one level reaching two times the intact spine and two levels attaining the same magnitude, while spreading across the two operated levels. These results show considerable load sharing due to ADR. The increased posterior column loads, in contrast to decreased load sharing in the ACDF case, underscore the potential for facet-related changes over time. From this perspective, patients with suspected facet joint issues may need to be carefully selected for this type of surgery.

Clinical Implications - Adjacent level, ACDF

While many studies are published with different ADRs, this is the first study to the best knowledge of the authors, to compare one and two level ACDF and ADR using a C2-T1 FEM. Results from the present head-to-head comparison offer some clinical insights into the behavior of two different surgical approaches for anterior fixation. As briefly described in the introduction, ACDF is the gold standard, and numerous retrospective studies have been conducted to analyze postoperative outcomes. Adjacent segment disease is a topic of clinical interest, especially in the context of revision procedures. A recent long-term (5 to 30 years) and large sample size (166 patients) study reported that more than 90% of the ACDF patients had worsened anterior and posterior osteophytes at segments immediately adjacent to the fusion (Rao et al., 2016). Degenerative changes were significantly affected ($p<0.05$) by the proximity of the level to the fusion. This observation served as a basis to focus the present study on the adjacent level kinematics and anterior and posterior column loads, determined by the range of motion and disc pressures and facet loads. For the ACDF option, compared to the one level option, at the superior level, two level surgeries resulted in increased ROM by 1.4 and 1.9 times in flexion and extension, increased anterior column load-sharing by 1.1 and 1.4 times in flexion and extension, and posterior column load-sharing by 2.3 times in extension. These results are indicative of the added 'burden' for the superior segment for the two level ACDF. The results for the inferior level also showed a similar pattern, especially for motion and anterior load column sharing. Taken together, these findings while supporting clinical observations of accelerated degeneration/adjacent segment diseases in ACDF patients, provide quantitated data on the kinematics and internal load sharing between the two components, i.e., external and internal responses delineating the biomechanics of commonly used anterior surgical procedures for cervical spondylosis and disc herniation. The identification of anterior osteophytes at adjacent levels reported in clinical literature may be due to the increased anterior load sharing in ACDF patients (Gore, 2001; Gore et al., 1986; Rao et al., 2016).

Clinical Implications – Adjacent level, ADR

A similar line of discussion is advanced for the one- and two-level ADR outputs. Compared to the one level option, at the superior level, in flexion, two-level surgeries resulted in increased ROM and anterior column load sharing by 1.9, and 3.1 times, and in extension, both parameters decreased (0.82 and 0.98 times). At the inferior level, only the anterior column load sharing had an increase (1.2 times in flexion), while all other variables were lower than the one-level surgery case. These results are a consequence of the preservation of motion at the index level, the intent of any ADR, and from this perspective, the design feature appears to have been accomplished. Furthermore, the relatively smaller changes in the magnitudes (exception of superior level anterior column loading for the two-level compared to the single level) with ADR than the ACDF indicates its role in imparting less changes to the intact spine.

Uniqueness of the Present Study

Clinical investigations provide information on surgical outcomes, and they are routinely assessed on images and quality of life metrics (Rao et al., 2016). Human cadaver spine experiments provide information on external responses, e.g., ROM (Wheeldon et al., 2006). Quantification of internal responses such as anterior and posterior load sharing is best accomplished using finite element models, and this unique feature was used in the present study to compare one- and two-level surgeries with two different approaches: conventional bone graft and artificial disc. While this approach is not novel, finite element models evaluating the effects of one- and two-level ACDF and Mobi-C have not been published. This specific ADR is approved by the US-FDA for both one- and two-level disc disorders(Beaurain et al., 2009; Davis et al., 2015; Dufour et al., 2019; Nunley et al., 2018). It is therefore, important to delineate the internal and external biomechanics under these two surgical conditions, and this is one uniqueness of this study.

Another uniqueness is to compare the anterior and posterior load sharing for both ACDF and ADR, and one- and two-level surgeries. Interpreting the bi-column loads from a clinical perspective, especially for selecting specific options for patients with suspected facet joint issues is a practical application of the results from the present research. The relatively lesser changes in the adjacent levels with this US-FDA approved ADR over the conventional ACDF supports its use by the surgeon as an effective option to treat the spinal disorder, another conclusion from this study.

CONCLUSIONS

Conventional two level ACDFs impart greater changes to adjacent segment biomechanics, i.e., external ROM and internal anterior and posterior load sharing, than one level ACDF. Such changes may accelerate adjacent segment disease over time, a phenomenon reported in clinical investigations. The use of ADRs instead of ACDFs showed limited changes in kinematics and load sharing, thus potentially limiting the adjacent segment effects due to motion preserving artificial disc arthroplasties. At the index level, however, while ACDF decreased kinematics and anterior and posterior load sharing, ADR surgeries increased load sharing in the posterior column. The considerable increase in posterior column load sharing with the ADR compared to the ACDF hitherto not quantified using this approach in the published literature, is a new finding from this study. The increased load sharing may in turn increase the development of heterotopic ossification and facet arthropathy and may subsequently act to reduce the overall motion preservation over a long term. Such issues may need additional attention in patients with suspected facet-joint involvements.

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