Finite Element Analysis Study of the Influence of Simulated Surgical Methods on Kinematics of a Model of the Full Cervical Spine

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Abstract

It is common that a surgical method is used to treatment pain and other problems that are diagnosed to be due to a severe form of degenerative disc disease (DDD). In the cervical spine, DDD is frequently seen at the C5-C6 level and the three most widely used surgical methods are anterior cervical discectomy without fusion (ACD), anterior cervical discectomy with fusion achieved with the aid of either an autologous or a synthetic bone graft (ACDF), and total disc replacement (TDR). The present study involved the determination of the influence of each of the three widely used surgical methods on the kinematics of the full cervical spine. For this investigation, we built a detailed three-dimensional solid model of the full cervical spine (C1-C7 levels), simulated surgical treatment at one level (C5-C6), applied loadings that are clinically-relevant, and used the finite element analysis method. The biomechanical parameter determined was the principal motion at each of intersegmental positions. It was found that relative to the results when an intact model was used, 1) the highest frequency of the smallest % changes in principal intersegmental motions was obtained when the TDR model was used. This finding is consistent with results that show that when ACDF and TDR are compared in randomized clinical trials, as good as or better patient outcomes were obtained when the latter was used.

Keywords: Finite element analysis (FEA); Cervical spine; Fusion; Total disc replacement

Introduction

The consensus is that degenerative disc disease (DDD), especially in its severe form, plays a key role in the etiology of a large assortment of symptomatic disorders of the spine, among which are herniation of the nucleus pulposus, discogenic pain, loss of disc height, loss of segmental mobility, development of osteophytes along the spine, myelopathy, radiculopathy, myeloradiculopathy, and traumatic instability at one or more levels [1]. In the cervical spine, when any of these disorders is diagnosed as originating from DDD and the pain/discomfort is not relieved by a conservative treatment, such as physical therapy and intermittent traction [2], the usual recourse is to use a surgical modality, the most widely used of which are anterior cervical discectomy without fusion (ACD), anterior cervical discectomy followed by fusion (ACDF), and disc arthroplasty (implantation of a total disc replacement) (TDR) [3,4]. Other surgical modalities include percutaneous nucleotomy [5] and nucleus pulposus replacement [6].

Three shortcomings of the very large body of literature on finite element analysis (FEA) of models of the cervical spine are noted. First, a model of the full cervical spine (herein, defined as C0-T1, or C0-C7, or C1-T1, or C1-C7) is used in only a few studies [7-13]. Second, to the best of the present workers’ knowledge, there are no studies in which a model of the full intact cervical spine and its modification to simulate each of the three most widely used surgical method (ACD, ACDF, and TDR) was used. Third, in two of the studies in which a full cervical spine model was used, kinematic parameters were not determined [9,13]. This omission is surprising given the fact that many normal activities of daily living involve motion of the cervical spine. Furthermore, since, in many clinical reports, data on motions are given, this omission means that only limited discussion of the clinical relevance of the reported FEA results can be undertaken.

In the present FEA study, we constructed a three-dimensional (3D) solid model of the full cervical spine (C1-C7), validated it,
and then used the validated model to determine the influence of each of the simulations of the three widely used surgical methods on the principal motion at each of the intersegmental locations in the model. For the simulations, DDD was taken to occur at a level that is commonly presented, namely, C5-C6 [14], and each of the applied loadings used is clinically-relevant [9,15].

Materials and Methods

Four 3D solid models of the full cervical spine (C1-C7) were constructed: the first was of an intact, healthy spine and the others were modifications of this spine to simulate the three surgical methods studied.

Model of intact, healthy spine

The solid model was constructed by using digitized quantitative axial computed tomography scans/images of the bony parts of the full cervical spine of a male cadaver imported from the Visible Human Project® dataset (National Library of Medicine, Bethesda, MD, USA), a 3D scanning software package (Mimics® Version 8.1; Materialise, Inc., Leuven, Belgium), a 3D medical image processing and editing software package (RapidForm® Version 2006; INUS Technology, Inc., Seoul. South Korea) and a computer-aided drawing software package (ProEngineer® Wildfire 5.0; Parametric Technology Corporation, Needham, MA, USA). These bony parts were the vertebral bodies, the posterior elements (transverse processes, pedicles, laminae, spinous processes, and facet joints), and the endplates. Constructed separately were the discs (with the annulus fibrosus and the nucleus pulposus occupying 60% and 40% of the total volume, respectively [16]) and the ligaments. The final solid model was obtained by merging the three sub-models (bony parts, discs, and ligaments) (Figure 1A).

The final solid model was meshed using an FEA software package (ABAQUS®, Version 6.13; Abaqus, Inc., Providence, RI, USA). Details of element types used for the finite element (FE) meshing and the properties of all the materials in the model are given in (Table 1). The convergence criterion used was a change of <1.5% in the rotation at C1-C2 of the model, under a loading of 1 Nm axial flexion + 73.6 N compression force, between successive changes in mesh density. Validation involved comparing the range of motion of each of the motion segments in the model, under various applied loadings, to applicable experimental results reported in the literature.

Models of simulated surgically altered spines

Each of the simulated surgically altered models was obtained by modifying the solid geometry of the INT model and, then, meshing it.

For ACD Model, the inferior endplate on the C5 vertebral body, the disc at C5-C6, the ALL at C5-C6, and the superior endplate on the C6 vertebral body were all removed. Then, the inferior surface of the C5 vertebral body and the superior surface of the C6 vertebral body were sculpted so that they fitted perfectly. These steps were consistent with the surgical method used by Nandoe-Tiwari et al. [26]. For ACDF Model, the tissues removed were the same as for the ACD Model with the exception that the empty disc space was filled with a brick-shaped graft (height and area = 100% and 85% those of the removed disc, respectively [27]). It was ensured that the posterior edge of the graft did not touch the posterior longitudinal ligament at C5-C6 and the superior and inferior faces of the graft were considered fully bonded with the inferior surface of the C5 vertebral body and the superior surface of the C6 vertebral body, respectively. These steps are consistent with a surgical procedure, namely, the Smith-Robinson method [28]. For TDR Model, the tissues removed were the same as those in ACD Model, but, in this case, the empty disc space was filled with a notional endplates-and-mobile insert TDR design (Figure 2A) ensuring that there was perfect contact between the top and bottom surfaces of the implant with the inferior surface of C5 and the superior surface of C6, respectively. In terms of materials, this notional design is comparable to four of the six TDR designs that are approved by the US Food and Drug Administration for use in clinical work, namely, Mobi-C® (LDR Spine USA, San Antonio, TX, USA), PCM® (Cervitech, Rockaway, NJ, USA), ProDisc-C® (Synthes, Inc., Philadelphia, PA, USA), and Secure®-C (Globus Medical, Audubon, PA, USA). Details of the element types used for the FE meshing and the properties of the materials are given in (Table 1).

Boundary conditions and loadings

For each of the four models (INT, ACD, ACDF, and TDR Models), the loading was applied to the superior surface of the C1 vertebral...
body while the inferior surface of the C7 vertebral body was fully fixed.

The applied loadings used were: 1) 1 Nm flexion moment + 73.6 N axial compression force; 2) 1 Nm extension moment + 73.6 N axial compression force; 3) 1 Nm left lateral bending moment + 73.6 N axial compression force; 4) 1 Nm right lateral bending moment + 73.6 N axial compression force; 5) 1 Nm counter-clockwise-acting (left) axial torsional moment + 73.6 N axial compression force; and 6) 1 Nm clockwise-acting (right) axial torsional moment + 73.6 N axial compression force The compression force simulates the weight of the head [9], while the magnitudes of the moments and the axial compression force are clinically-relevant [15].

### Biomechanical parameters determined

Under each loading, the motion at each of the intersegmental positions (that is, at C1-C2, C2-C3, C3-C4, C4-C5, C5-C6, and C6-C7) was determined, which allowed computation of the change in that motion when a surgical-simulated model was used (ACD, ACD, or TDR Model) compared to the value when INT Model was used.

### Results

#### Convergence test and model validation results

The mesh density of the converged INT model consisted of 421,160 elements and 89,161 nodes (Figure 1B). With a few exceptions, the FEA results obtained using converged INT Model are within the range obtained from experimental tests, as reported in the literature (Figure 3). Differences between some features of INT Model and those used in these experimental tests include spine section covered (C1-C7 in the present study versus C0-C7 [29], C2-T1 [30], and C2-C7 [31]) and the method and position used to apply the moments (in the present study, application of a load on the superior surface of the C1 vertebral body along an anatomical axis while the inferior surface of the C7 vertebral body was fixed in position and direction versus...

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**Table 1** Element type and elastic properties of the tissues/materials in the finite element model.

| Tissue/materiala | Element type         | Elastic propertyb | Reference          |
|------------------|----------------------|-------------------|--------------------|
| Cortical bone    | 3-noded triangular general purpose shell | $E_{11} = 9,600\, \text{MPa}$; $E_{22} = 9,600\, \text{MPa}$; $E_{33} = 17,800\, \text{MPa}$; $G_{12} = 3,097\, \text{MPa}$; $G_{13} = 3,510\, \text{MPa}$; $G_{23} = 3,510\, \text{MPa}$; $\nu_{12} = 0.55$; $\nu_{13} = 0.30$; $\nu_{23} = 0.30$ | Rho [17]; Cowin [18] |
| Cancellous bone  | 4-noded tetrahedral  | $E_{11} = 144\, \text{MPa}$; $E_{22} = 99\, \text{MPa}$; $E_{33} = 344\, \text{MPa}$; $G_{12} = 53\, \text{MPa}$; $G_{13} = 63\, \text{MPa}$; $G_{23} = 63\, \text{MPa}$; $\nu_{12} = 0.23$; $\nu_{13} = 0.17$; $\nu_{23} = 0.11$ | Ulrich et al. [19] |
| Posterior elements | 4-noded tetrahedral | $E = 3,500\, \text{MPa}$; $\nu = 0.29$ | Kumaresan et al. [20] |
| Annulus fibrosus | 4-noded tetrahedral  | $E = 4.2\, \text{MPa}$; $\nu = 0.45$ | Ha et al. [21] |
| Nucleus pulposus | 8-noded brick       | $E = 1.0\, \text{MPa}$; $\nu = 0.499$ | Ha et al. [21] Brolin and Halldin [22] |
| Endplates        | 4-noded tetrahedral  | $E = 500\, \text{MPa}$; $\nu = 0.40$ | Yoganandan et al. [23] |
| ALL              | Nonlinear tension-only spar | $E = 30.0\, \text{MPa}$ | Zhang et al. [8] |
| PLL              | Nonlinear tension-only spar | $E = 20.0\, \text{MPa}$ | Zhang et al. [8] |
| ISL, LF (C1-C2)  | Nonlinear tension-only spar | $E = 10.0\, \text{MPa}$ | Zhang et al. [8] |
| SSL, ISL, LF (C2-C7) | Nonlinear tension-only spar | $E = 1.5\, \text{MPa}$ | Zhang et al. [8] |
| CL (C1-C3)       | Nonlinear tension-only spar | $E = 10.0\, \text{MPa}$ | Zhang et al. [8] |
| CL (C3-C7)       | Nonlinear tension-only spar | $E = 20.0\, \text{MPa}$ | Zhang et al. [8] |
| A1L              | Nonlinear tension-only spar | $E = 5.0\, \text{MPa}$ | Zhang et al. [8] |
| TL               | Nonlinear tension-only spar | $E = 20.0\, \text{MPa}$ | Zhang et al. [8] |
| ApL              | Nonlinear tension-only spar | $E = 20.0\, \text{MPa}$ | Zhang et al. [8] |
| Iliac crest bone graft  | 4-noded tetrahedron | $E = 3,500\, \text{MPa}$; $\nu = 0.25$ | Natarajan et al. [24] |
| Co-Cr-Mo alloy   | 8-noded brick       | $E = 220\, \text{GPa}$; $\nu = 0.32$ | Ratner et al. [25] |
| UHMWPE           | 8-noded brick       | $E = 1\, \text{GPa}$; $\nu = 0.49$ | Ratner et al. [25] |

**a**ALL: anterior longitudinal ligament; PLL: posterior longitudinal ligament; SSL: supraspinous ligament; ISL: interspinous ligament; LF: ligamentum flavum; CL: capsular ligament; AL: alar ligament; TL: transverse ligament; ApL: apical ligament. UHMWPE: ultra-high-molecular-weight polyethylene. **b**E: modulus of elasticity; v: Poisson’s ratio. 11, 22, and 33 refer to the radial, tangential, and longitudinal axes of the bone, respectively. The literature references are for the values of the elastic properties.
example, via a spinal gimbal and an XY table [31]). Furthermore, comparison of FEA results, reported by previous workers, to the same set of literature experimental results yield the same trends as found in the present results (Figure 4). When all of these observations were taken into account, it may be concluded that INT model was validated.

Simulated surgically altered spine models

The final mesh density of each of these models is given in (Table 2) and, as an example, the meshed finite element model of TDR Model is shown in (Figure 2B).

When the whole collection of results (Figure 5 and Tables 3 and 4) is considered, it is seen that the TDR Model produced the highest frequency of the smallest % changes in principal intersegmental motions (22 times out of a possible 36 times).

Discussion

In the literature on FEA studies of models of the cervical spine, studies on comparison of the influence of the three most widely surgical methods for treating pain and problems due to DDD on kinematics are lacking. This aspect is the subject of the present work.

Relevant literature FEA studies are considered those that have both of the following two characteristics. First, the FEA study was of a model of the intact spine section and a minimum of two of the three surgical simulation models utilized in the present work. Second, the range of motion (ROM) results were obtained under the same types of applied loadings as were used in the present study. By this definition, to the best of the present worker’s knowledge, the only relevant literature FEA studies are those by Mo et al. [33] and by Faizan et al. [34]. Mo et al. [33] used a C3-C7 model and simulated ACDF and TDR at C5-C6 and applied a loading of 73.6 N preload + 1.8 Nm moments on C3. A comparison of Mo et al.’s ROM results and corresponding ones from the present study (Figure 6A) shows that, at C5-C6, the two sets of results for an ACDF model are similar. However, the TDR model results given by Mo et al. are higher than those obtained in the present work.
Neck Disability Index score, pain score, neurological parameters, consistent with results of clinical and patient outcomes (such as to the corresponding values when an intact model was used, is of the smallest % changes in intersegmental motion, relative

The finding that TDR Model produced the highest frequency of the smallest % changes in intersegmental motion, relative to the corresponding values when an intact model was used, is consistent with results of clinical and patient outcomes (such as Neck Disability Index score, pain score, neurological parameters, number of secondary surgical procedures, flexion-extension ROM, and number of adverse device-related events) from randomized controlled trials (RCTs) in which ACDF and an approved TDR design (Bryan, or Prestige LP or ProDisc-C) were compared in the treatment of symptomatic DDD at one level (C3-C4 or C4-C5 or C5-C6 or C6-C7) in patient-matched cohorts.

We note some limitations of the study. First, in the solid model, the facet joints were included as part of the posterior elements, rather than separate tissues. For a C5-C6 model subjected to 1.8 Nm flexion moment + 73.6 N axial compression force or 1.8Nm extension moment + 73.6 N axial compression force, the principal motions were ~25% greater when facet joints were included as separate entities compared to when they were included as part of the bone posterior structures. Second, in the solid model, the muscles were not included. The important role played by muscle forces in spinal motions is well recognized. However, this aspect is particularly important when dynamic or impact loading is applied. In the present study, the loading was quasi-static. Third, in the solid model of each of the simulated surgical methods, perfect bonding was assigned at the respective interfaces (for example, inferior surface of C5-superior surface of C6, in the case of ACF Model; and inferior surface of C5-superior surface of TDR design and inferior surface of TDR design-superior surface of C6, in TDR Models). In other words, the simulation was for the situation that is likely to exist several weeks after surgery. In the FEA, a Coulomb friction contact or a stick-slip contact formulation could be used to model an interface. Fourth, the solid model was built using data taken from one person and, as such, it is unknown if the results obtained have generality. This problem could be overcome by using a parametric modeling method or a parametric and patient-specific modeling method. Fifth, in the FEA, the ground substance and the fibers in the annulus fibrosus (AF) and the nucleus pulposus (NP) were each taken to be linear, isotropic, elastic materials. Other material models been used for these tissues, including hyperelastic (Mooney-Rivlin) or hyperelastic (neo-Hookean) for the annulus ground substance, nonlinear stress-strain relationship for the annulus fibers [23], hyperelastic incompressible solid for the nucleus [47]; incompressible fluid for the NP [48]; and poroelasticity for both the AF and the NP [49]. For a C4-C6 model, subjected to 100 N compression force uniformly distributed on the superior surface of the C4 vertebral body, in some tissues, such as the inferior endplate at C5, the mean von Mises stress was markedly sensitive to the constitutive model used for the AF and NP, whereas other tissues, such as the cancellous bone at C4, showed moderate sensitivity [41]. Since the present work is a parametric study, each of these the limitations applies to all the models; as such, the trends in changes in intersegmental motions and, hence, our conclusions are valid.

Conclusion

For a model of the full cervical spine (C1-C7), with simulated surgical treatment for problems due to severe DDD at C5-C6, subject to clinically-relevant loading, the highest frequency of the smallest % changes in principal intersegmental motions was obtained when TDR was simulated. This finding is in consonance with the results of many RCTs in which TDR was compared to ACDF in several patient-matched cohorts.
Figure 5  Summaries of the present principal intersegmental motion results from the INT, ACD, ACDF, and TDR models.

Table 3  Summary of % changes in principal intersegmental motion under different applied loadings*.

| Load type       | Model | C1-C2 | C2-C3 | C3-C4 | C4-C5 | C5-C6 | C6-C7 |
|-----------------|-------|-------|-------|-------|-------|-------|-------|
| Flexion         | ACD   | -10.6 | -17.6 | -16.0 | 17.0  | -62.0 | 100   |
|                 | ACDF  | 1.0   | 10.6  | 19.7  | 152   | -91.0 | 200   |
|                 | TDR   | -1.5  | -1.5  | -9.1  | -10.4 | 25.0  | -13.7 |
| Extension       | ACD   | -0.1  | -7.9  | -30.7 | -31.4 | 23.2  | -61.1 |
|                 | ACDF  | 1.0   | 9.2   | 3.0   | 99.0  | -91.0 | 70.0  |
|                 | TDR   | -0.4  | 5.0   | -1.0  | -4.0  | 17.2  | -6.0  |
| LLB             | ACD   | -0.3  | -1.0  | -4.0  | -12.3 | -15.0 | -20.3 |
|                 | ACDF  | 7.8   | 20.0  | 44.0  | 197   | -91.4 | 200   |
|                 | TDR   | -0.3  | 3.7   | -5.0  | -13   | 13.6  | -6.5  |
| RLB             | ACD   | -2.0  | -23.7 | -12.2 | -3.6  | -43.0 | -6.7  |
|                 | ACDF  | 8.5   | 22.3  | 41.2  | 196   | -91.0 | 229   |
|                 | TDR   | 0.2   | 3.0   | -5.6  | -2.0  | 31.0  | -1.4  |
| LTM             | ACD   | -0.3  | -9.0  | -15.6 | -6.8  | 6.0   | -8.5  |
|                 | ACDF  | 0.5   | 5.0   | 5.6   | 110   | -86.0 | 158   |
|                 | TDR   | -0.4  | -7.7  | -14.1 | -8.9  | 42.1  | -2.0  |
| RTM             | ACD   | 0.1   | -7.5  | -12.9 | -2.6  | 32.4  | -7.8  |
|                 | ACDF  | 0.5   | 6.0   | 3.8   | 109   | -86.0 | 82.0  |
|                 | TDR   | 0.5   | -6.0  | -2.4  | -0.1  | 30.0  | -7.8  |

*LLB: left lateral bending; RLB: right lateral bending; LTM: left (counter-clockwise) axial torsional moment; RTM: right (clockwise) axial torsional moment.
Conflict of interest

No benefits of any form have been or will be received from a commercial party related directly or indirectly to the subject of this manuscript.

Table 4 Summary of the models that yielded the smallest % change in principal intersegmental motion under different applied loadings*.  

| Loading   | C1-C2 | C2-C3 | C3-C4 | C4-C5 | C5-C6 | C6-C7 |
|-----------|-------|-------|-------|-------|-------|-------|
| Flexion   | ACDF  | TDR   | TDR   | ACD   | TDR   | TDR   |
| Extension | ACD   | TDR   | TDR   | TDR   | TDR   | TDR   |
| LLB       | ACD   | ACD   | ACD   | ACD   | TDR   | TDR   |
| RLB       | TDR   | TDR   | TDR   | TDR   | TDR   | TDR   |
| LTM       | ACD   | ACDF  | ACDF  | TDR   | ACD   | TDR   |
| RTM       | ACD   | TDR   | TDR   | TDR   | ACD   |

*LLB: left lateral bending; RLB: right lateral bending; LTM: left (counter-clockwise) axial torsional moment; RTM: right (clockwise) axial torsional moment.

Figure 6 Comparison between the present FEA results and relevant FEA results given by Mo et al. (applied loading: 73.6 N + 1.8 Nm) [33] (A), and relevant FEA results given by Faizan et al. (applied loading: 75.0 N + 2.0 Nm) [34] (B).
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