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[Invited Review Paper]

Recent advances in finite element modeling of the human cervical spine †

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Abstract

The human cervical spine is a complex structure that is the most frequently injured site among all spinal injuries. Therefore, understanding of the cervical spine injury and dysfunction, and also biomechanical response to external stimuli is important. Finite element (FE) modeling can help researchers to access the internal stresses and strains in the bones, ligaments and soft tissues more realistically, and it has been widely adopted for spine biomechanics research. Although in recent years numerous techniques have been developed, there are no recent literature reviews on FE models of the cervical spine. Our objective was to present recent advances in FE modeling of the human cervical spine in terms of component modeling, material properties, and validation procedures. Model applications and further development are also discussed. The integration of new technologies will allow us to generate more accurate and comprehensive model of the cervical spine, which can increase efficiency and model applicability. Finally, the FE modeling can help to facilitate diagnosis, treatment, and prevention technologies for cervical spine injuries.

Keywords: Cervical spine; Finite element modeling; Material properties

1. Introduction

The human spine supports the weight of the body and protects the spinal cord. The cervical spine is the most frequently injured site among all spinal injuries, and injuries in this region can be life threatening [1, 2]. Biomechanical models, including in vitro, in vivo, and Finite element (FE) models, can help gain a greater understanding of the mechanisms of spinal injury and dysfunction and spinal responses to various treatment and clinical problems [3]. FE modeling can evaluate stresses and strains in complex structures of the spinal bones, ligaments, and soft tissues more realistically, and it has been widely adopted for spine biomechanics research [4].

Earlier models of the cervical spine had been developed to predict only spinal motion or displacement response to loading, where the models describe vertebrae as simple rigid bodies and other connective tissues as beam or spring elements [5-9]. Later, more anatomically correct and three-dimensional (3D) FE models, from individual segment to functional motion unit models, were used to investigate the cervical spine [10-16]. Yoganandan et al. [10-13] developed an FE model of the C4-C6 human cervical spine from Computed tomography (CT) scans and validated their model against experimental results. They described the stress distributions in cervical spine components under different loads. Maurel et al. [14] created a lower cervical spine model using parameterized bone geometries and then analyzed mechanical behavior of the cervical functional unit. Goel and Clausen [17] developed a C5-C6 motion segment model and used it to predict load sharing through the spinal elements in response to various loading modalities. Teo and Ng [16] developed a lower cervical spine C4-C6 model with important anatomic features such as facet joints, a posterior arch, intervertebral discs, and ligaments. They validated their model under axial compression, flexion, and extension loads. Later, significant improvements were made in procedures for the reconstruction of bone geometries and in validating models against experimental data sets generated from various investigators [18-21]. Recently, several groups have developed detailed 3D FE models of the cervical spine, including models in children and women [22-29]. Typically, vertebral bone geometries were created based on CT images, and connected by Intervertebral discs (IVDs) which consist of nucleus pulposus and annulus ground substance reinforced with annulus fibers. Nonlinear cable or truss elements are used for ligaments, while facets are modeled with surface interactions. In addition, some studies incorporated spinal cord tissues into human cervical spine models to investigate Spinal cord injuries

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(SCIs) [30-34].

Several review studies have been published on the computational modeling of the cervical spine [35-39]. Jager [35] first reviewed the physical properties and experimental validation data for computational models of the cervical spine. Huelke and Nusholtz [36] reviewed clinical, laboratory, and mathematical studies on the biomechanics of the human cervical spine injuries. Panjabi [3] briefly reviewed the FE models of the cervical spine in comparison to other biomechanical models. Yoganandan et al. [37] focused on the developments in model construction, material properties, loading and boundary details, and validation of all existing cervical spine FE models. Fagan et al. [38] reviewed FE analysis in spine research and presented modeling of the cervical spine components. Although these authors have presented comprehensive reviews on FE modeling of the human cervical spine, there are no literature reviews on FE models of the cervical spine that include new modeling information, such as the spinal cord, nerve roots, Cerebrospinal fluid (CSF), and other specific parts. Therefore, we have reviewed the improved components in cervical spine FE modeling, from bone components to spinal cord tissues, because of potential interest to clinicians and biomechanical engineers.

2. Model geometry and meshing

2.1 Vertebra

The data conversion procedure for developing the FE model of vertebra from CT data is shown in Fig. 1. Once the Computer-aided design (CAD) or solid model is generated, model meshing is performed using various mesh generation methods. The FE models are typically generated using a mixture of tetrahedral, wedge, and hexahedral elements, but FE models rarely consist entirely of hexahedral elements [40]. Hexahedral elements are better than tetrahedral elements because hexahedral elements can increase the analysis accuracy and decrease the number of overall element meshes [24]. However, most studies use tetrahedral elements to generate cervical spine models [19, 21, 28, 29, 41-43]. Lee et al. [25] used hexahedral and tetrahedral elements for vertebral body and posterior bony structure, respectively. Panzar and Cronin [44] developed cancellous bone using hexahedral elements, while cortical bone and endplates in the vertebra were both modeled using shell elements. Laville et al. [45] created a method to automatically generate parametric and subject-specific hexahedral mesh generation of the lower cervical spine. Kallemeyn et al. [22, 40] proposed a hexahedral mesh generation method using multiblock techniques to create a subject-specific FE model of the cervical spine. They released an open-source software called IA-FEMesh [46] to make hexahedral mesh development easier for anatomic modeling, which was also used in an FE model of the C2-T1 cervical spine of an adult male [27]. Dong et al. [26] adopted a multiblock meshing approach to generate hexahedral meshes for the cervical spine of a ten-year-old child. In addition, many existing cervical

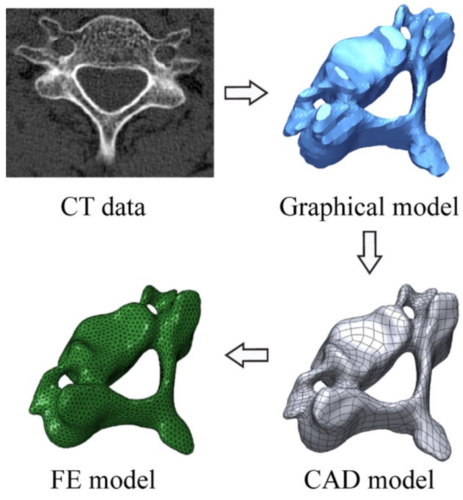

Fig. 1. Data conversion procedure for developing an FE model of a vertebra.

spine models are symmetric about the mid-sagittal plane, which can reduce the time required to generate models, but asymmetric model geometry should be considered in order to analyze asymmetrical variations and to obtain accurate results.

2.2 Intervertebral disc

For soft structures and other connective tissues within the bony anatomy, additional imaging techniques may be needed for more accurate and detailed modeling. In IVD modeling, the complex structure, which consists of annulus ground with multiple layers of fibers surrounding the fluid nucleus, has been simplified. In general, the disc geometry is obtained after vertebral modeling, where the cranial and caudal surfaces of the disc are created based on the surfaces of the endplates. Then the annulus ground and nucleus is separately generated based on the ratio between the two substances. The nucleus area was 40 % - 50 % of the cross-section of the total disc [26, 27]. The annulus fibers are positioned with a mean of 25 degree against the horizontal plane, and the content in annulus ground accounted for 20 % of the volume of the annulus [19, 25]. In many cases, disc models have been composed of three parts: Annulus ground and nucleus pulposus, both as solid elements, and annulus fibers, as a tension-only linear element [18, 19, 21, 25, 29, 42, 47-49]. However, more accurate disc models have been included in some cervical spine models by incorporating fluid elements for the nucleus pulposus [22, 27, 44, 50].

2.3 Facet joints

The facet joints are synovial joints and are observed between articular processes of two adjacent vertebrae. The joint consists of hard and soft tissue structures, providing resistance to physiologic and traumatic loads and maintaining strength and stability of the spine [12]. The facet joints are often mod-

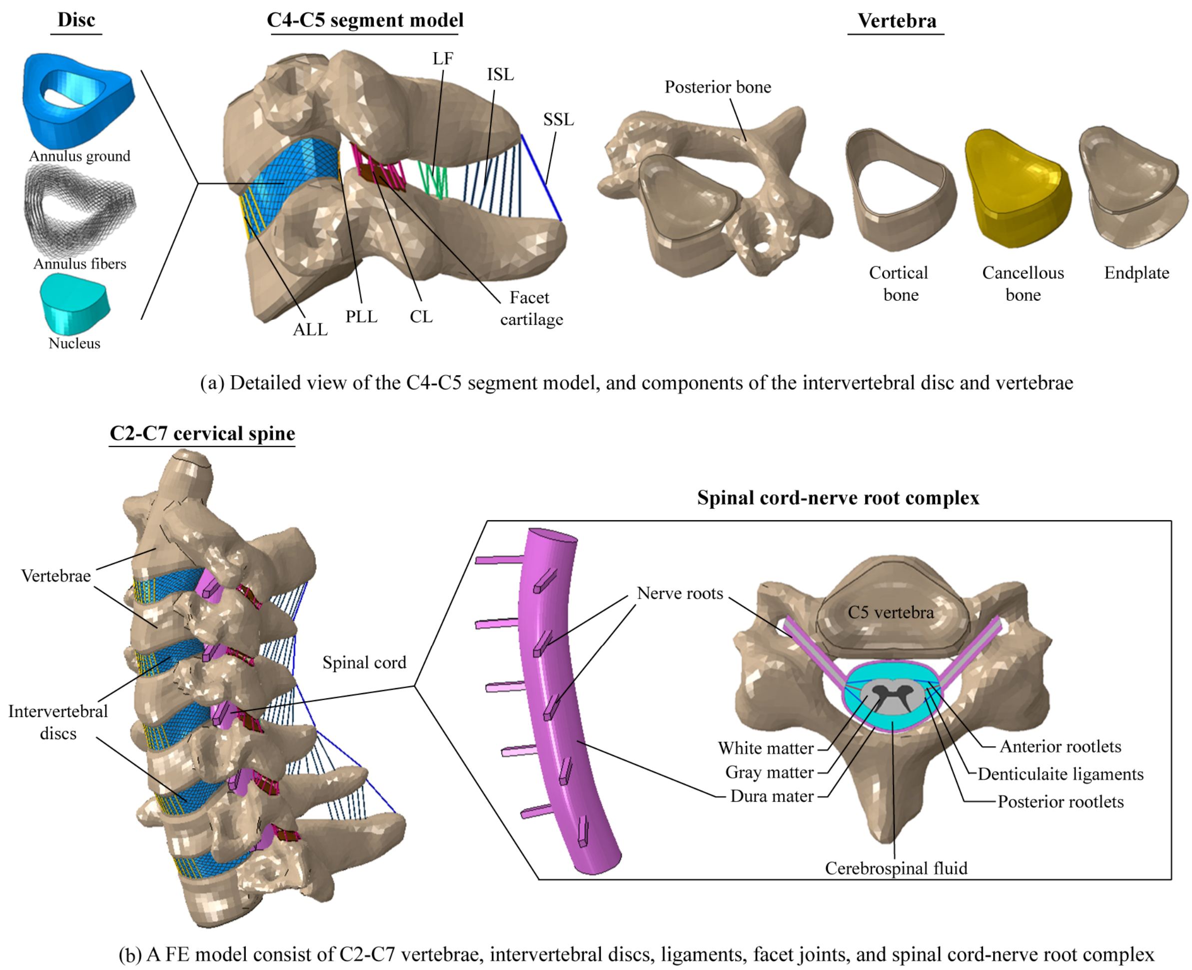

Fig. 2. (a) Detailed view of the C4-C5 segment model; (b) FE model of the C2-C7 cervical spine with the spinal cord-nerve root complex structure.

eled using solid elements, gap elements, and contact elements. Solid elements were used to model articular cartilage on each articular surface with a gap that ranges from 0.35-1.4 mm [25, 26, 44, 51]. Erbulut et al. [27] utilized gap contact elements to simulate facet joints, and a number of studies have considered facet joints as sliding contact elements [19, 21, 42, 48, 52].

2.4 Ligaments

The cervical spine has six major ligaments: Anterior longitudinal ligament (ALL), Posterior longitudinal ligament (PLL), Interspinous ligament (ISL), Supraspinous ligament (SSL), Capsular ligament (CL), and Ligamentum flavum (LF). The main role of these ligaments is to provide stability and flexibility during motion. The ligaments were usually represented as tension-only truss elements, where the origin and insertion locations could be defined by using anatomical references. Also, the membrane [28] and spring elements [20] have been used to generate the cervical ligaments. Since the linear elements were used for modeling of the ligaments, the Cross-sectional area (CSA) of the ligaments was assigned for each component. However, the CSA of the ligaments vary between studies and, because limited experimental data is available, further characterization of ligaments is needed to accurately represent in the models. The model development details are described in Fig. 2(a).

2.5 Spinal cord tissues

The inclusion of the spinal cord into the cervical spine model adds complexity to the models; therefore, few studies have included the spinal cord in their FE models of the human cervical spine [30-34, 53]. In these studies, the geometries of the spinal cord were created based on the geometry of the cervical column and anatomic measurements of the human spinal cord, for example, the anterior-posterior and lateral diameter of the spinal cord. Spinal cord diameters and the thickness of the Cerebrospinal fluid (CSF) layer in the human cervical spine can be found elsewhere [54-56]. However, the spinal cord is composed of white matter, gray matter, dura mater, CSF, Denticulate ligaments (DLs), and nerve rootlets within the dural sheath. Thus, additional information is necessary when developing a detailed spinal cord model. Scifert et al. [30] first developed a C5-C6 cervical segment model with

Table 1. Summary of material properties for select FE models of the cervical spine.

	Wheeldon et al. [20]		Kallemeyn et al. [22]		Lee et al. [25]		Panzer and Cronin [44]	
	C4-C7 model		C2-C7 model		C2-C7 model			
Components	Material properties	Type	Material properties	Type	Material properties	Type	Material properties	Type
Vertebra Cortical bone Cancellous bone Posterior bone Endplate Facet cartilage	12 GPa/0.30 100 MPa/0.20 3.5 GPa/0.25 600 MPa/0.30 10.4 MPa/0.40	Solid	10 GPa/0.30 450 MPa/0.25 3.5 GPa/0.25 - -	Solid	12 GPa/0.29 450 MPa/0.29 3.5 GPa/0.29 500 MPa/0.40 10 MPa/0.40	Solid	16.8 GPa/0.30 100-300 MPa/0.1-0.3 - 5.6 GPa/0.30 10 MPa/0.40	Solid
Disc Nucleus Annulus ground Annulus fibers	3.4 MPa 1-4.8 MPa 500 MPa	Solid Solid Rebar	Incompressible 4.2 MPa/0.45 450 MPa/0.30	Fluid Solid Rebar	1 MPa/0.49 3.4 MPa/0.40 110 MPa/0.30	Solid Solid Truss	K = 1.72 GPa Hill foam Orthotropic elastic	Fluid Solid Shell
Ligaments ALL PLL LF ISL CL SSL	Nonlinear force-displacement curve	Spring	Nonlinear hyperelastic material	Truss	10 MPa/0.30 10 MPa/0.30 1.5 MPa/0.30 1.5 MPa/0.30 10 MPa/0.30 1.5 MPa/0.30	Truss	Nonlinear force-displacement curve	Truss

the spinal cord. For impact scenarios, Kimpara et al. [53] added a spinal cord model into the Total human model for safety (THUMS) in which the spinal cord model consisted of hexahedral elements (Representing white matter, gray matter, and CSF) and shell elements (Representing DLs, pia, and dura mater). Greaves et al. [31] modeled the C4-C6 cervical spine with a spinal cord model for distinct injury scenarios. They modeled the spinal cord and dura mater using brick and shell elements, respectively. Recently, Khuyagbaatar et al. developed a C1-T1 cervical spine that included a spinal cord-nerve root complex model [34]. In this model, the spinal cord consists of white matter, gray matter, dura mater, CSF, nerve roots, and DLs. They used hexahedral elements to represent white matter, gray matter, and dura mater with nerve roots; spring elements for DLs and the nerve rootlets inside the dura mater; and fluid elements to represent the CSF (Fig. 2(b)).

3. Material properties

3.1 Vertebra

Most studies have utilized homogeneous and isotropic material definition for various components of the cervical spine. The material properties adopted in some FE models of the cervical spine are summarized in Table 1. Vertebrae are generally modeled as a linear elastic material with elastic modulus (E) and Poisson's ratio (ν). The most commonly used values were E = 12 GPa and ν = 0.3 for cortical bone, E = 450 MPa and ν = 0.25 for cancellous bone, and E = 3.5 GPa and ν = 0.25 for posterior bone. Some studies have utilized orthotropic elastic [43, 44] and isotropic elastic-plastic material [28] to model cancellous bone and vertebrae, respectively. In addition, the densities of the spinal components were necessary during dynamic analysis [1, 24, 44].

3.2 Intervertebral disc

As mentioned, the IVD consists of annulus ground with multiple layers of fibers surrounding the fluid nucleus. Most studies assumed that the disc is a linear elastic material with E = 1 MPa for the nucleus and E = 3.4 - 4.2 MPa for annulus ground [1, 18, 19, 22, 25, 42, 48]. Incompressible fluid (Bulk modulus of 1.72 GPa) and Hill foam materials have also been used for the nucleus and annulus ground, respectively [26, 44]. Recent studies have also utilized nonlinear hyperelastic and viscoelastic models for either nucleus or annulus ground modeling [2, 27-29, 43]. The annulus fibers were usually modeled with five to six layered structures with an elastic modulus of 110 – 450 MPa using truss, membrane, and rebar elements [18, 19, 22, 25, 27, 48, 51]. Recently, the stress-strain curve was used to model nonlinear behavior of the annulus fibers using spring elements [2, 29]. The nonlinear and orthotropic elastic characteristics of the annulus fibers were represented with membrane elements [26, 28, 44].

3.3 Facet joints

The articular cartilage in the facet joints was modeled using a linear elastic material with E = 10 MPa and ν = 0.4 [25, 26, 28] or E = 23.8 MPa and ν = 0.3 [29]. Surface-to-surface contact conditions were applied on each facet cartilage with a friction coefficient of 0 to 0.1. Studies have also treated the facet joints as a contact between adjacent vertebrae, where they created contact surfaces directly on the vertebral bone without facet cartilage [21, 42, 48]. Nonetheless, facet joint modeling is not well described in most studies.

3.4 Ligaments

Recent studies have characterized the nonlinear behavior of ligaments using force-deflection curve or stress-strain properties based on experimental findings [17, 57, 58]. Yoganandan et al. [57] obtained force-deflection and stress-strain data of the human cervical spine ligaments using in situ axial tensile tests on ligaments from 25 cadavers. Geol and Clausen [17] defined nonlinear characteristics of the ligaments using a stress-strain relationship. Mattucci et al. [58] provided a detailed measure of force-displacement data of a younger population by performing tensile tests. However, some studies modeled the ligaments using linear elastic truss elements [25, 51].

3.5 Spinal cord tissues

Animal and human spinal cord modeling studies have represented the spinal cord as hyperelastic and viscoelastic with a hyperelastic shear modulus ranging from 2 kPa to 40 kPa [32-34, 59-63]. Greaves et al. [31] also used a linear elastic property for the spinal cord model. The linear elastic materials used to model the dura had an elastic modulus between 5 MPa and 80 MPa. Both fluid and solid materials were used to model the CSF. Kimpara et al. [53] represented the CSF with linear solid elements. The CSF layer has been demonstrated with Newtonian fluid characterized by a viscosity of CSF [32, 60]. All of these studies used data from previous animal studies due to a lack of studies on the material properties of the human spinal cord [64-66]. Hung et al. [64] observed the nonlinear force-deformation relationship of the cat spinal cord. Ichihara et al. [65] identified differences between the mechanical properties of white matter and gray matter in the bovine cervical spinal cord. Fiford and Bilston [66] observed a nonlinear stress-strain response using the rat spinal cord. Quantitative data on the tensile properties of the nerve roots and DLs are available [67-69].

4. Model validation

4.1 Cervical spine model

Validation is an important step in FE models to ensure the model behaves like a real structure under the same experimental conditions. When an FE model shows good agreement with the experimental data, the model can provide reliable and accurate biomechanical responses. There are several studies that provide experimental data for validation of FE models of the cervical spine [70-72]. Panjabi et al. [70] measured the intersegmental movement patterns of the cervical spine in flexion, extension, bilateral axial torsion, and bilateral lateral bending under a pure moment of 1.0 Nm. Wheeldon et al. [71] provided flexion and extension data of a normal cervical spine from a young individual. They applied moments ranging from 0.33 Nm to 2.0 Nm to the cervical spine. Nightingale et al. [72] provided flexion-extension responses and bending strength of the male cervical spine with a maximum moment of 3.5 Nm. For validation of an FE model, the structural loads and constraints should be carefully considered in the context of the actual experimental setup or physiological conditions. The cervical spine is an articulation structure where the boundary conditions can be specified at the superior and inferior vertebrae. In general, the inferior surface of the lowermost vertebra is firmly fixed, while the superior vertebra can be unconstrained or fixed with some degree of freedom. Loading is applied on the superior plane of the uppermost vertebra of the cervical spine, for example bending moments along the flexion-extension, lateral bending, and axial rotation planes.

FE models have been directly compared to the predicted responses inferred from experimental results. In most cases, the predicted intersegmental Range of motions (ROMs) in flexion, extension, lateral bending, and axial rotations were used to validate the models with a maximum bending moment of 2.0 Nm [1, 18, 20, 22, 25, 27, 29, 42, 49, 73]. Panzer et al. [23] validated their model based on flexion-extension response at the segment level with a bending moment up to 3.5 Nm, the tension response of the ligaments, and the head kinematic response to impact. Dong et al. [26] analyzed model response under a bending moment of 2.5 Nm. Kallemeyn et al. [22] validated their model using in-house experimental data obtained from specimens used for mesh development. Zhang et al. [24] used kinematic responses of the head and spine segments to validate the C0-T1 model under dynamic conditions. Mustafy et al. [74] validated a C2-C3 segment model using ROM, contact pressure in facet joints, and failure forces in ligaments. In addition, the model response was predicted under a bending moment with compressive follower loads of 50 N – 73 N to mimic the in vitro cadaver testing protocol in either the ASTM or ISO standard of implant test [42, 48, 50]. All of these studies showed good agreement with the experimental results.

4.2 Spinal cord model

The spinal cord models have been verified by using cord displacement or reaction forces under static and dynamic loads. However, it is difficult to replicate realistic loading and boundary conditions for the spinal cord, and many simplifications and assumptions were made in previous modeling studies. Generally, the top and bottom surface of the spinal cord were constrained, and contact between the spinal cord and vertebrae was created during the analysis. Greaves et al. [31] validated a spinal cord model against previous experimental data based on the reaction force at the indenter tip in a static compressive load up to 2.1 N [64]. The dynamic impact test has also been used to validate a spinal cord model by using three types of pellets with the pellets impacting the spinal cord at an initial velocity of 4.5 m/s [32]. The displacement behavior of the spinal cord was measured and compared against the ex vivo studies [75, 76].

5. Model applications

FE models are widely used in clinical and practical applications to improve knowledge of the etiology and pathology of injuries as well as to simulate surgery and test spinal instrumentation for the cervical spine. FE model use has since been expanded to the field of SCI medicine. Cervical spine response and soft tissue damage were predicted in quasi-static [21] and dynamic impact conditions (e.g., Near-vertex drop impact, rear and frontal impact, whiplash injury) using cervical functional spinal unit [74] and head-neck complex models [1, 23, 24]. These models can be used for automotive safety applications to improve occupant safety during various crash scenarios.

Several studies have simulated the surgical procedures on the cervical spine, for example, laminectomy with facetectomy [18], open door and double door laminoplasties [77], hemilaminectomy and minimally invasive approach for spinal lesions [48], and resection of the uncovertebral joint [29]. The goal of these studies was to better understand the impact of various surgeries on cervical spinal biomechanics. FE modeling techniques have also been used to evaluate the effects of artificial discs [19, 25, 50, 51, 73, 78, 79], stand-alone disc implants [42], and anterior fixation plates [43] as well as fixation screws in lower [80] and upper cervical spine [81]. These studies compared the results between an intact cervical spine model and an implanted model under various loadings.

In the field of SCI medicine, different types of injuries to the spinal cord were simulated. Greaves et al. [31] conducted a primary analysis of contusion, dislocation, and distraction injury mechanisms of the spinal cord. Later, Khuyagbaatar et al. [32] investigated the biomechanical response of the spinal cord in various grades of contusion, dislocation, and distraction injury. In addition, the effect of cervical laminectomy, laminoplasty, and hemilaminectomy on the spinal cord response was investigated [33]. The causes of injuries to the spinal cord and nerve roots after laminectomy were explained from a biomechanical point of view using a FE analysis [32, 34].

6. Future directions of the FE modeling of the cervical spine

We reviewed recent FE models of the human cervical spine with a focus on the modeling of cervical spine structural components, which included both bony and soft tissues, load and boundary conditions, and validation procedures. Although there has been less effort in developing FE models of the cervical spine than on the development of the lumbar or thoracic spine, much improvement has been made. The articles covered in this paper can provide fundamental information for the development of a generic model, but there has been a tendency to integrate a detailed joint model into the motion-driven musculoskeletal model to estimate stress/strain and contact pressures on the joints during various activities. Here are some recommendations for further improvements in model development:

- There are some studies that propose an automatic or semi-automatic mesh generation method based on geometrical parameters. The fully automated model generation technology for subject-specific modeling could represent the next level of modeling for clinical and practical application with an advance of medical imaging technology.
- Most research groups were using 2D linear elements for soft tissue modeling, which does not represent the actual anatomical characteristics. The more detailed and appropriate 3D nonlinear volumetric modeling for ligaments and muscles may accurately predict the cervical spine response.
- Precise determination of the material properties of the cervical spine components in both healthy and pathological conditions based on individualized experimental data is necessary to construct patient-specific models that can be utilized effectively for treatment and surgical planning purposes.
- Dynamic FE modeling by integrating the physiologically realistic loading conditions derived from a motion capture system can be powerful tools, which may hopefully improve our understanding of the biomechanical functioning of the cervical spine during various activities.

The integration of new technologies will allow us to generate a more accurate and comprehensive model of the cervical spine, which can increase efficiency and model applicability. Then, the level of confidence and prediction accuracy in the modeling studies will be enhanced. Finally, FE modeling can help facilitate diagnosis, treatment, and prevention technologies for cervical spine injuries.

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Nomenclature

FE : Finite element
3D : Three-dimensional
CT : Computed tomography
IVD : Intervertebral disc
SCI : Spinal cord injury
2D : Two-dimensional
CAD : Computer-aided design
ALL : Anterior longitudinal ligament
PLL : Posterior longitudinal ligament
ISL : Interspinous ligament
SSL : Supraspinous ligament
CL : Capsular ligament

LF : Ligamentum flavum
CSA : Cross-sectional area
CSF : Cerebrospinal fluid
DLs : Denticulate ligaments
E : Elastic modulus
υ : Poisson ratio
ROM : Range of motion

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