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Development and Validation of a Three-Dimensional Finite Element Model of Cervical Spine (C3-C5)

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ABSTRACT

The objective of this investigation is to develop a detailed, non-linear asymmetric three-dimensional anatomically and mechanically accurate FE model of complete middle cervical spine (C3-C5) using Hypermesh and MSC.Marc software. To achieve this goal, the components of the cervical spine are modeled using 20-noded hexagonal elements. The model includes the intervertebral disc, cortical bone, cancellous bone, endplates, and ligaments. The structure and dimensions of each spinal component are compared with experimentally measured values. In addition, the soil mechanics formulation of MSC.Marc finite element software is applied to model the mechanical behaviour of vertebrae and intervertebral discs as linear isotropic two-phase (biphasic) material.

The FE simulation is conducted to investigate compression, flexion\extension and right\Left lateral bending modes. The simulation results are validated and compared closely with the published experimental data and the existing FE models. In general, results show greater flexibility in flexion and less flexibility in extension. The flexion/extension curves are asymmetric with a greater magnitude in flexion than in extension. In addition, the variations of the predicted lateral C4-C5 disc bulge are investigated and the results show that the maximum disc bulge occurs at the C4-C5 anterior location.

INTRODUCTION

The need for reliable data in biomedical applications demands accurate and fast solution techniques. The existing methods are limited to numerical and experimental approaches while there is no explicit analytical method. Experimental techniques yield the most accurate results but they are highly expensive and time consuming. Hence, there is an increasing surge into utilizing numerical solutions amongst which Finite Element Method (FEM) is the most popular one. The FE method is an essential part of many of today's engineering activities. It was first developed in the 1950s in the aircraft industry and continues to be an indispensable tool in the design of most of the critical components in today's aircraft, [1]. This method is applied to simulate and obtain the benchmarks for stress and strain developed under biomechanical loading in different areas of human-body. Cervical spine is one of the prevalent areas of a human body which is under high-risk of injuries. Therefore, more accurate and reliable methods and FE models would help to reduce the risk of injury where the safety is the most important issue of the state-of-the-art designs.

The inherent complexity of cervical spine is the main reason for the existence of the very simplified FE models in the literature. Although simplification reduces the computational time, it reduces the accuracy of the FE model. For example, among various existing cervical FE models in the literature, the geometry of the vertebrae were modeled very simplistic or assumed to be symmetric, which reduces the reliability and accuracy of the existing models, [2, 3, 4, 5, 6 and 7]. In this study, the authors have paid great deal of attention to construct an anatomically accurate asymmetric three-dimensional finite element model of the complete middle cervical spine of a 28 year-old healthy male.

Moreover, one of the key components of any accurate FE analysis is the accurate modeling of the material physics. Many experimental observations suggested that it would be more realistic to consider biological tissue as a multi-phase system. However, the majority of the existing FE cervical models modeled the bone and intervertebral disc as an elastic or single phase material, [8, 2, 9 and 6]. Therefore, in this study, the vertebrae and intervertebral discs were modeled as linear, isotropic and two-phase material.

NOMENCLATURE

e_0	=	Initial Void Ratio
E	=	Young Modulus
n	=	Porosity
k_0	=	Initial Permeability
ν	=	Poison Ratio

MODELING

A previously developed surface cervical spine model of a 28 year-old healthy male was imported into finite element pre-processing software Hypermesh 8.0, Altair Engineering Inc., Troy, MI, and MSC.Marc, MSC Software Copr., Los Angeles, U.S.A, for geometry validation and mesh construction. The dimensions of each cervical spine components were measured and compared against experimentally measured values, [10, 11, 12 and 13] which are known as 50% percentile males or mid-sized male. The validation showed that the cervical spine components were within one standard deviation of experimental data. Using Hypermesh, the surfaces of each cervical spine components were meshed using the 20-noded hexagonal element. Because of the geometric complexity of the vertebrae and to control the distribution of mesh density at different locations, a mixture of fine mesh (smaller size elements) and a sparse mesh (larger size elements) were used. This was done to ensure the geometry of vertebrae was precisely modeled. While meshing the surfaces, the quality of the elements was checked and poor quality elements were remeshed.

The mesh size was chosen so that the aspect ratio and element distortion of each element was within the acceptable tolerance. The tolerance for aspect ratio was chosen to be .5 or less and distortion ratio was chosen to be .2 as recommended by Hypermesh.

The C3 to C5 vertebrae were modeled as deformable bodies, consisting of separate elements for cortical bone and cancellous bone since the mechanical properties of cancellous and cortical bone are different (**Figure 1**). To create the cortical bone and bony endplates, the surfaces of the vertebrae were meshed using 2D quadrilateral shell elements. The 2D elements were converted to 3D solid hexagonal elements by offsetting them to an appropriate thickness. Even though the thickness of the cortical bone varies depending on each vertebral body and on the location (anterior and posterior), [14], in this study, an average measured anterior and posterior cortical thickness value was used. Therefore, the cortical bone thicknesses of 0.46 mm, 0.48 mm and 0.53 mm were assigned to C3, C4 and C5 vertebrae, respectively. For boney endplates, the superior and inferior elements of the C3-C5 vertebrae body were given thicknesses as experimental measured values, [14]. All cortical bones and endplates were created using two layer solid elements.

The geometric profile of cancellous bone in the vertebrae was model using the inner elements of the cortical bone. The inner elements of cortical bone were used to create a surface. This surface constituted the outer surface of cancellous bone. Again, using Hypermesh, the surface was meshed and was imported into pre-processor MSC.Marc software to convert the shell elements to 20 noded hexahedral elements.

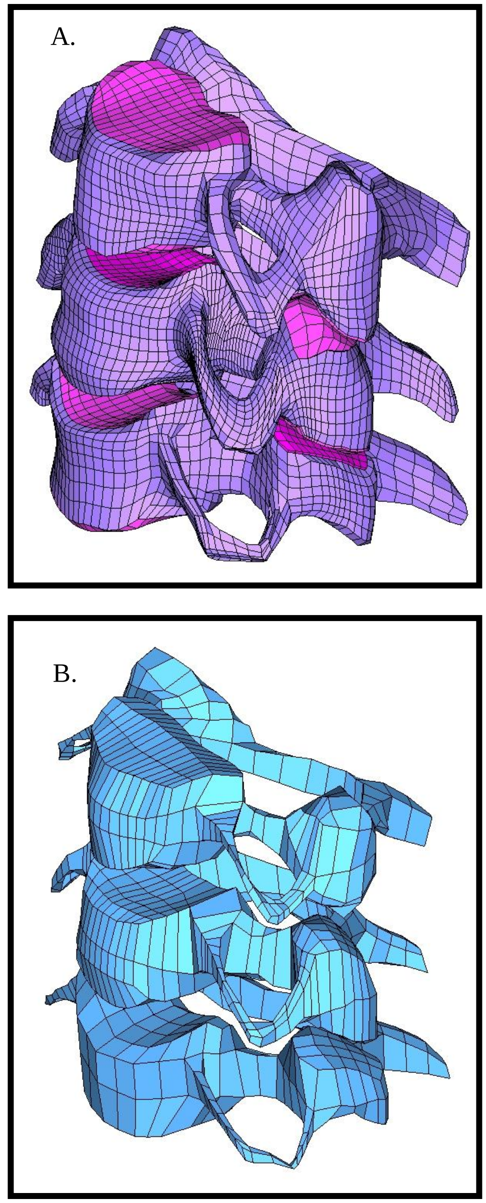

Figure 1: A. C3-C5 cortical bone and B. C3-C5 cancellous bone.

In this study, the facet joints of C3 to C5 vertebrae (vertebrae end processes) were modeled using solid elements for articular cartilage. The size of the articular cartilages was modeled using experimental measured dimensions, [12]. The shape of articular cartilages was dependent on the vertebral geometry. All articular cartilages were created using one layer of solid elements. The articular cartilage elements were attached to the vertebrae using tied contact to allow for attachment of dissimilar meshes. Furthermore, a finite sliding frictionless contact was defined between articular cartilages so that if there is any contact due to applied load, a penetration between the surfaces could be avoided.

The Intervertebral Disc (IVD) was divided into two components: nucleus pulposus and annulus fibrosus (**Figure 2**). The geometries of the C3-C4 and C4-C5 discs were imported into the Hypermesh 8.0. The anterior and posterior heights of each disc were measured and compared with experimental measurements of 50% percentile male values, [10], and it was found that the dimensions were within one standard deviation of the experimental values. The surface of each disc was meshed using quadrilateral shell elements. The shell elements were exported into MSC.Marc finite element software. By using the hex mesher module of MSC.Marc, the shell elements of the each disc were converted to 20-noded solid elements.

In this study, five major ligaments, the anterior longitudinal ligament (ALL), the posterior longitudinal ligaments (PLL), the ligamentum flavum (LF), the interspinous ligament (ISL), the supraspinous ligament (SSL) were considered, and modeled using 20-noded solid elements. The ligaments were created using one layer of solid elements. The location of each ligament was determined according to anatomical descriptions. These ligaments were modeled using experimentally measured dimensions, [15]. The detailed three-dimensional finite element model of C3-C5 middle cervical spine is shown in **Figure 3**.

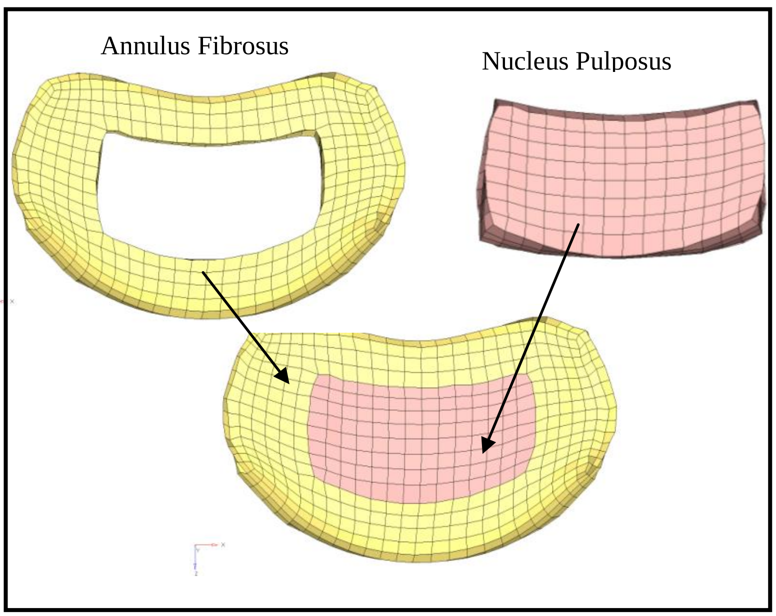

Figure 2: Superior view of the C3-C4 Invertebral disc.

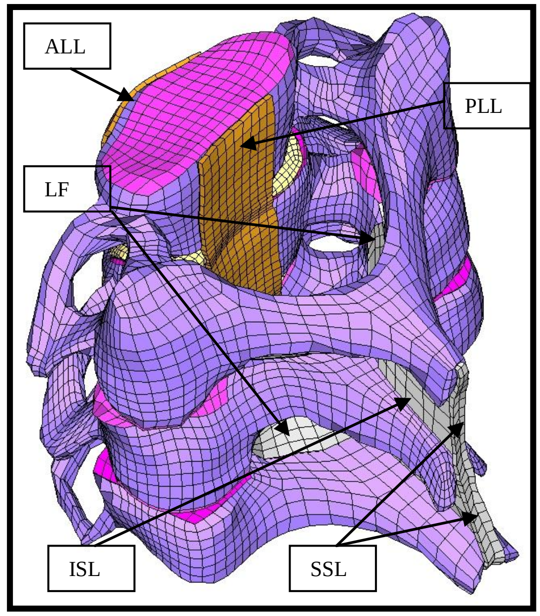

Figure 3: The C3-C5 finite element model.

MATERIAL PROPERTIES

A number of theories related to soft tissue's mechanical behavior have been developed, and many experimental observations suggested that it would be more realistic to consider biological tissues as a multi-phase system. Therefore, two theories of multi-phase behavior have been proposed to study the soft tissue. They are biphasic theory and poroelastic theory. Most of the previous models considered bone and the discs as a single phase elastic or viscoelastic material, [8, 2, 9 and 6]. Therefore, in this study, the vertebrae and intervertebral discs were modeled as linear, isotropic and two-phase material.

The poroelasticity theory is different from the biphasic theory of mixtures in that the former specifies a continuous distribution of pores in the solid matrix whereas the latter specifies a continuous distribution of solid and fluid phases. Consequently, the poroelasticity theory uses the displacement of the continuum and the average relative fluid displacement as field variables, whereas the biphasic theory uses the solid and absolute fluid displacement as field variables. This difference in the primary unknowns leads to different expressions of the field equations, for detailed discussion please consult Prendergast *et al.* [16]. Given this difference between the biphasic approach used for soft tissues and the poroelastic approach developed for soil mechanics, it has been shown that poroelastic models are equivalent to biphasic models provided that the fluid phase is inviscid, [16, 17 and 18] . Prendergast *et al.* [16] compared three commercial FEM programs MARC, DIANA, and SWANDYNE with the model of Spilker *et al.* [19] for a confined compression test of a biphasic cartilage layer. They concluded that the soil mechanics capability of finite element codes can be used to model biphasic tissues with reasonable accuracy, for both linear and nonlinear cases. Based on their findings, this study uses the soil mechanics capability of MSC.Marc to model biphasic tissues for linear case, given the fluid in this study is assumed inviscid. Hence, the components of vertebrae and intervertebral discs were modeled as a linear isotropic biphasic model, and the ligaments were modeled as a linear isotropic single phase model. In the biphasic theory, the tissue is modeled as a two phase immiscible mixture, consisting of an intrinsically incompressible solid phase (collagen and proteoglycan) and an intrinsically incompressible fluid phase (interstitial water).

Upon choosing an appropriate material model for all components, the next step was to input accurate material properties of the spinal component into the material model. However, material properties of the human cervical spine components reported in the literature vary drastically depending on the approach, method, and specimen preparation, [8, 9, 20, 21, 22, 23 and 24]. In this study, material properties of each component of the cervical model were varied within the range of values reported in the literature. This was done so that an optimum match of the model output was obtained in comparison to the published experimental data under compression. Hence, numerous analyses were performed in an iterative manner so that one specific set of material properties (Table 1 & 2) proved to be an optimum matches to the experimental published data.

In the current formulation, it is assumed that the fluid is of a single phase. Thus, it was required to define both the solid and fluid properties. For the fluid phase, it was necessary to identify the density of fluid, dynamic Viscosity, fluid bulk modulus and porosity *n* properties. In the majority of the soft tissue studies, the fluid phase was considered to have the same properties as that of water, [21, 22 and 24]. Consequently, this study used same assumption. The porosity *n* is the ratio of the volume of fluid to the total volume of the solid. The porosity is given in MSC.Marc through the INITIAL POROSITY or the INITIAL VOID RATIO option which in this study initial void ratio $e = n/(1-n)$ was selected.

In addition, the solid properties were selected from previous literatures, [9, 20, 23 and 24]. Table 1 summarized the properties that were used for vertebrae and intervertebral discs components. The material properties of ligaments were taken from previous published literatures and tabulated in Table 2, [9].

Table 1: Material properties of Vertebrae and Intervertebral discs components.

Spinal Components	**Soil (Solid) Phase Properties**				**Fluid Phase Properties**			
	Elastic Modulus E (MPa)	Poisson's Ratio ν	Initial Permeability k_0 ($m^4N^{-1}s^{-1}$)	Density [Kg/m^3]	Density of Fluid [Kg/m^3]	Dynamic Viscosity [N/m^2]	Fluid Bulk Modulus [N/m^2]	Initial Voids ratio e_0
Cortical Bone	10,000 [6]	0.29 [6]	1.0×10^{-20} [1]	1.83×10^3 [3]	1000	2.1×10^9	$2.2 \times 10^{7*}$	0.02 [1]
Cancellous Bone	100 [6]	0.29 [6]	1.0×10^{-13} [1]	$1.12 \times 10^{3®}$	1000	2.1×10^9	$2.2 \times 10^{7*}$	0.40 [1]
Endplate	500 [6]	0.4 [6]	7.0×10^{-15} [1]	1.12×10^3 [4]	1000	2.1×10^9	$2.2 \times 10^{7*}$	4.00 [1]
Articular Cartilage	500 [6]	0.4 [6]	7.0×10^{-15} [1]	1.12×10^3 [4]	1000	2.1×10^9	$2.2 \times 10^{7*}$	4.00 [1]
Annulus Fibrosis (AF)	4.2 [6]	0.45 [6]	3.0×10^{-16} [1]	1.061×10^3 [5]	1000	2.1×10^9	2.2×10^7 [2]	2.33 [1]
Nucleus Pulpous (NP)	1.5 [1]	0.49 [6]	3.0×10^{-16} [1]	1.342×10^3 [5]	1000	2.1×10^9	3.35×10^6 [2]	4.00 [1]

[1] Argoubi *et al.,* [27]
[2] Wu J.S.S *et al.,* [21]
[3] Lee C.K *et al.,* [22]
* Assumed to be the same as Annulus Fibrosis value.
[4] Loret B *et al.,* [23]
[5] Simon B.R *et al.,* [24]
[6] Ha S.K, [8]
® Assumed to be the same as Endplates .

Table 2: Material Properties of Ligaments

Ligaments	**Posterior Longitudinal Ligament (PLL)**	**Anterior Longitudinal Ligament (ALL)**	**Supra-Spinous Ligament (SSL)**	**Inter-Spinous Ligament (ISL)**	**Ligamentum Flavum(LF)**
Elastic Modulus E (MPa)	20 [6]	54.5 [6]	1.5 [6]	1.5 [6]	1.5 [6]
Poisson's Ratio ν	0.3 [7]	0.3 [7]	0.3 [7]	0.3 [7]	0.3 [7]

[6] Ha S.K, 2[8]
[7] Scifert J *et al.*,[9]

FE ANALYSIS

To ensure that the process of modeling and assigning material properties to various components of the cervical spine were accurate and suitable, the FE model of C3-C5 cervical spine was validated against the results reported in the literature. A reasonable agreement between the simulation results and published data was considered as the validation of the model. Once the model was validated, further analyses were performed. The present model was validated under compression, flexion and extension, and right and left lateral bending load configurations. The predicted responses were compared against the published experimental and existing analytical results under the same boundary and load configurations, [25, 26 and 27]. In this study, biphasic formulation of MSC.Marc was adopted under the hypothesis of large deformations. The analyses were implicit, quasi-static and incorporated nonlinear geometry. The finite element model of C3-C5 cervical spine was subjected to an axial compressive load of 600 N centered over superior nodes of C3 vertebrae. The C3-C5 cervical segment was constrained in all degree of freedom at the C5 inferior vertebral body. The force-displacement response of FE model in middle vertebra was determined and compared against the *in vitro* experimental measurements performed by Shea *et al.* [25]. Furthermore, the lateral disc bulge at its mid-height under axial compressive load at anterior and posterior locations of the C4-C5 intervertebral disc was obtained.

The FEM model of C3-C5 was also used to predict the flexion, extension and lateral bending motions. Therefore, for all boundary conditions except compression the following pure moments of 0.33, 0.5, 1.0, 1.5 and 2.0 N.m were applied to the superior end of the C3 vertebra, while the inferior nodes of C5 were constrained in all directions. The response of the vertebrae to these pure moments was evaluated in the sagittal plane. The published experimental and FE simulation results of cervical spinal segment are all in the form of angular displacement. For this reason, our comparison is made based on the angular displacement as well. The resulting external angle response curves were obtained and compared against the experimental data of Wheeldon *et al.* [26 and 27]. Furthermore, the lateral disc bulge of each loading modes were computed.

RESULTS

The force versus displacement curve of C4 vertebra as a result of 600N compressive load is shown as **Figure 4.** The percentage of discrepancy between this analysis and Shea *et al.* [25] experimental test was approximately 4%. At 600N axial compressive load, the C4 vertebra was compressed by an average of 0.82 mm. Generally, the predicted mechanical response of the C3-C5 model under axial compressive loading agreed reasonably well with the published experimental measurements. The simulation showed that the compressive loading was coupled with a significant extension motion that agrees with Panjabi *et al.* [28] and Shea *et al.* [25], who reported various compression loads coupled with an extension loading conditions. Also, it was observed that the nucleus pulposus forces the annulus fibrosus to bulge outward in both anterior and posterior directions.

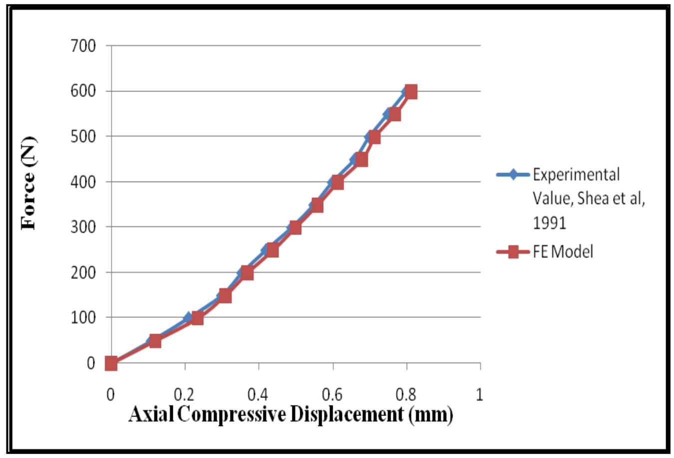

Figure 4: Comparison of the force-displacement response between present and the Shea *et al.* [25] results.

Validation of the model in flexion and extension was particularly important, since these are the primary modes of spinal loading. Under flexion and extension bending moments, the biomechanical responses of the C3-C4 and C4-C5 segment results corresponded well within experimental results, [26], see **Figure 5 and 6**. In flexion, the acceptable ranges (experimental range) of rotation at 2 Nm moments for C3-C4 and C4-C5 segments are 4.7 to 12 and 5.2 to 11 degrees, respectively. In this study, the maximum rotation degree at 2 Nm moments for C3-C4 and C4-C5 vertebrae was found to be 6.7 and 5.5 degrees respectively which fall within the experimental range. In extension, the acceptable range of rotation for both C3-C4 and C4-C5 vertebrae are 3.5 to 6 degree. Current result showed that the maximum degree of rotation for C3-C4 is 5.7 degree and for C4-C5 vertebra is 4.8 degree which fall within the experimental range. Note negative and positive in **Figure 5** and **Figure 6** indicate the extension and flexion response.

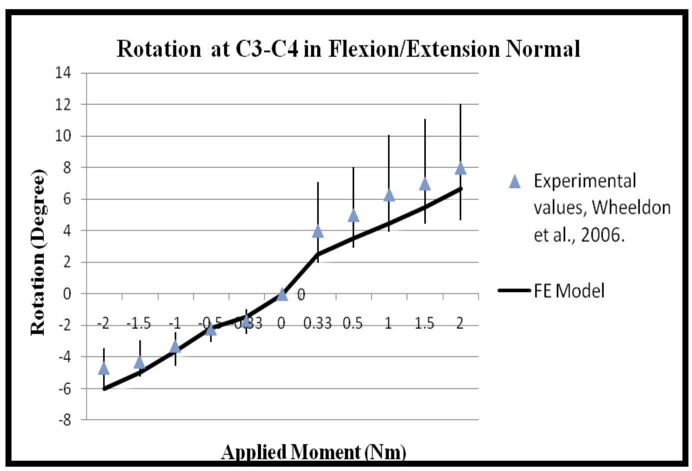

Figure 5: Comparison of the C3-C4 moment-rotation relationship between current results and Wheeldon *et al.* [26] results.

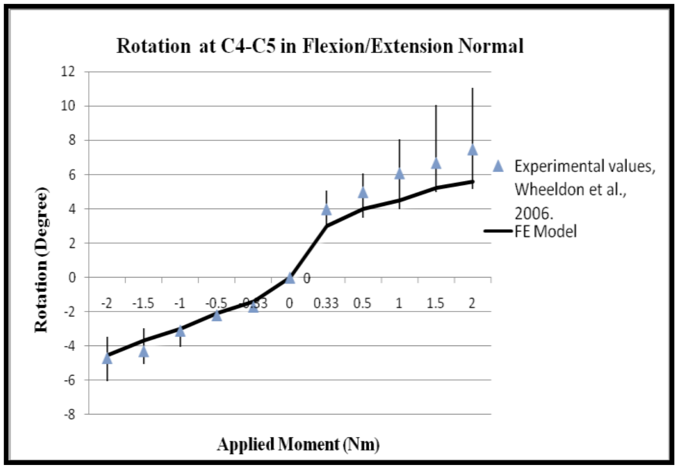

Figure 6: Comparison of the C4-C5 moment-rotation relationship between current results and Wheeldon *et al.* [26] results.

In addition to flexion/extension, right and left lateral bending moments were applied to the model and the results showed that they were within one standard deviation of the experimental results (**Figure 7**). Due to the limited *in vitro* studies carried out on upper cervical spine under lateral bending, the results of C4-C5 segment was used for model validation purposes. For C4-C5 segment, the maximum experimental ranges of rotation at 2 Nm during right and left lateral bending are 2.7 to 6 and 2 to 6 degrees, respectively. This study has shown the maximum rotation of 4 degree for right and 3 degree for left lateral bending that correspond to experimental data. Note that negative and positive in **Figure 7** indicate the left and right response, respectively.

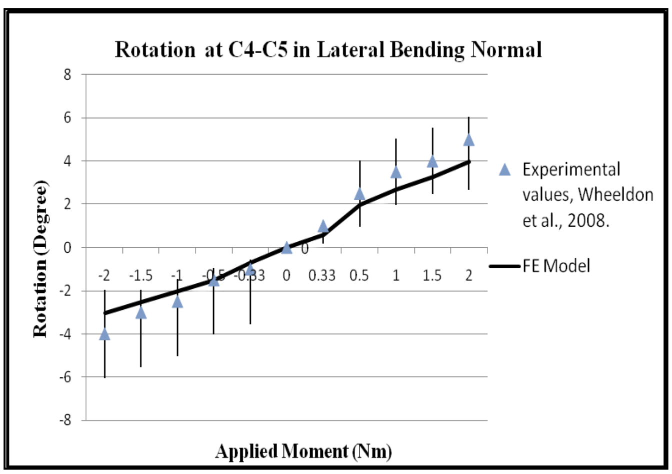

Figure 7 : Comparison of the C4-C5 moment-rotation relationship between current results and Wheeldon *et al.* [27] results.

Under axial compressive load and flexion/extension bending moments, the median, minimum and maximum values of C4-C5 disc bulge at its mid-height of anterior and posterior locations were summarized in **Table 3.** The results under compression show that the maximum bulge occurs at the anterior location of the C4-C5 disc which this is in agreement Ng et al. [2] observation. Under both flexion/extension, the maximum disc bulge occurs at the C4-C5 anterior location. However, in flexion the disc bulges outward in anterior location and in extension disc bulges inward in anterior location. In flexion, the nucleus pulposus tended to shift backward (towards posterior location) even though the disc bulged outward in the anterior. Furthermore, the disc bulges inward (concave) in the posterior location under flexion. Under extensional loading, the disc tends to shift inward at anterior location (disc bulge in) and it bulged outward in the posterior. The displacement on nucleus pulpous is opposite to the disc bulge out at posterior location. This is identical to the disc behavior under flexion. This behavior of the intervertebral disc agrees with qualitative observation of White and Panjabi [29]. Also, in flexion, the superior vertebrae was inclined to move in the anterior direction, and vice versa for the extension case which is similar to the motions reported by Moroney *et al.* [30]. Furthermore, the C4-C5 disc bulge of both left and right lateral bending moments at right and left locations of the disc were measured and summarized in **Table 4**. The results showed that in right lateral bending, the disc bulges more toward the right hand side and vice verse in left lateral bending. It was found that the applied lateral bending moment produced complementary axial rotational motion. This means that applied lateral bending moment induces small axial rotation (twisting of the vertebrae). This happens primarily because of the asymmetric geometry of cervical vertebrae. This coupled motion agrees well with experimental results reported by Moroney *et al.* [30].

Table 3: Median, Maximum and minimum values in millimeters of disc bulging evaluated for anterior and posterior locations.

Loading Direction	Location	C4-C5 Disc Bulge[mm]		
		Median	Max	Min
Compression	Anterior	0.57	0.64	0.45
	Posterior	-0.44	-0.58	-0.32
Flexion	Anterior	0.79	0.93	0.55
	Posterior	0.28	0.36	0.12
Extension	Anterior	-0.62	-0.67	-0.56
	Posterior	-0.45	-0.52	-0.37

Table 4: Median, Maximum and minimum values in millimeters of disc bulging evaluated for left and right locations.

Loading Direction	Location	C4-C5 Disc Bulge[mm]		
		Median	Max	Min
Right Lateral Bending	Left	-0.32	-0.52	-0.13
	Right	0.61	0.80	0.43
Left Lateral Bending	Left	0.34	0.46	0.22
	Right	-0.1	-0.20	-0.02

CONCLUSION

Anatomically accurate and detailed three-dimensional finite element model of the human middle cervical spine (C3-C5) using actual geometric data was developed. All the vital anatomic features of the middle cervical spine, such as the intervertebral disc, cortical bone, cancellous bone, endplates and ligaments were modeled. In most published FE cervical models, the vertebrae were generated in a manner such that there was mid sagittal plane of symmetry. However, in reality that's not the case. Therefore, in this study, non-symmetrical three-dimensional finite element model of complete middle cervical spine was created.

To do so, the structure and dimensions of each vertebrae and discs were compared with experimentally measured values of Gilad *et al.* [10] Panjabi *et al.* [11], Panjabi *et al.* [12]. Furthermore, the ligaments structure and dimensions were generated using experimental measurements of Yoganandan *et al.* [13]. Using Hypermesh 8.0 and MSC.Marc, the geometries of vertebrae and discs were meshed and the structure of the ligaments and spinal cord were created using 20-noded hexagonal elements.

This study used the soil mechanics formulation of MSC.Marc to model the vertebrae and intervertebral discs as linear biphasic tissue. Also, the ligaments were modeled as a linear isotropic model. A number of different experimental studies were chosen to address compression, flexion, extension and lateral bending modes of loading to gauge the accuracy of model as thoroughly as possible. The current finite-element model was validated against the *in vitro* experimental measurements performed by Shea *et al.* [25] and Wheeldon *et al.* [26 and 27].

It was important to develop a realistic finite element model that can effectively simulate the general finite displacement (displacement is large) of the human cervical spine. The current finite-element model can predict the biomechanical response of the human middle cervical spine (C3-C5) by using the nonlinear contact and geometric options under axial compressive force of up to 600 N and was validated against published data.

The flexion/extension curves were determined to be asymmetric about zero reference moment. In all curves, the mean value of the rotation under the extension load was determined to be less than the mean value of the flexion load. This asymmetric result is in agreement to that reported by Wheeldon *et al.* [26]. However, disagrees to that described by Panjab *et al.* [31], Nightingale *et al.* [32]. The cause of the difference can be explained by the anatomic asymmetry of the human neck in the sagittal plane, as well as an asymmetric vertebral body shape. It was found that the biomechanical response of the C3-C4 and C4-C5 model correlated well within Wheeldon *et al.* [26] experimental results.

In flexion and extension rotational responses of the motion segments, a "slack" effect can be seen. The initial portions of the responses have a large slope until to approximately 0.33 Nm where the ligaments are not playing as large of a role. Thus, the slop suddenly changes; this is defined as "slack". Once the ligaments have been stretched to a certain point, they influence the rotational response more significantly rather than the vertebral geometry or other factors. During flexion, the interspinous ligament, ligamentum flavum and posterior longitudinal ligament stretched. Conversely, during extension, anterior longitudinal ligament stretched. In addition, the flexion and extension motion is affected by the facet joints. In flexion the superior facet surfaces slide up and forward while in extension the superior facet surfaces slide down and backwards. The facet joint gap also undergoes a narrowing in extension and a widening in flexion, [26].

In addition to extension and flexion, the right and left lateral bending response of the C4-C5 vertebrae were computed and compared against the Wheeldon *et al.* [27] experimental results. These FEM results of lateral bending were within one standard deviation of the experimental results.

The lateral disc bulge of the C4-C5 was measured for both anterior and posterior locations. Under both compression and flexion/extension, the maximum disc bulge occurs at the C4-C5 anterior location. Under right lateral bending, the disc bulges more toward the right hand side and vice verse in left lateral bending.

Finally, the responses of the current FE model of cervical spine were validated at compression, flexion, extension and lateral bending. Therefore, this model will be used for further analysis and that is to consider spinal cord response into the model.

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