Automated Finite Element Meshing of the Lumbar Spine: Verification and Validation with 18 Specimen-Specific Models

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Abstract

The purpose of this study was to seek broad verification and validation of human lumbar spine finite element models created using a previously published automated algorithm. The automated algorithm takes segmented CT scans of lumbar vertebrae, automatically identifies important landmarks and contact surfaces, and creates a finite element model. Mesh convergence was evaluated by examining changes in key output variables in response to mesh density. Semi-direct validation was performed by comparing experimental results for a single specimen to the automated finite element model results for that specimen with calibrated material properties from a prior study. Indirect validation was based on a comparison of results from automated finite element models of 18 individual specimens, all using one set of generalized material properties, to a range of data from the literature. A total of 216 simulations were run and compared to 186 experimental data ranges in all six primary bending modes up to 7.8 Nm with follower loads up to 1000 N. Mesh convergence results showed less than a 5% difference in key variables when the original mesh density was doubled. The semi-direct validation results showed that the automated method produced results comparable to manual finite element modeling methods. The indirect validation results showed a wide range of outcomes due to variations in the geometry alone. The studies showed that the automated models can be used to reliably evaluate lumbar spine biomechanics, specifically within our intended context of use: in pure bending modes, under relatively low non-injurious simulated in vivo loads, to predict torque rotation response, disc pressures, and facet forces.

1. Introduction

Finite element (FE) modeling of the lumbar spine holds promise for pre-clinical evaluation of new devices, but subject-specific modeling and simulation remain challenging due to the uncertainty of material properties and boundary conditions (Dreischarf et al., 2014).

Alternatively, population-based modeling has the potential to offer information on device efficacy across a broad range of virtual patients, and parameter uncertainty can be accommodated with well-established probabilistic methods. Modeling a large number of subjects to represent a target population has been impractical, however, due to the complexity of spinal geometry and the time consuming process of model creation. A new method for automatic generation of FE meshes of the lumbar spine addresses this limitation (Campbell and Petrella, 2015), but it has not been formally evaluated, which was the focus of the present study.

Verification and validation (V&V) are the primary methods for evaluating the reliability and accuracy of an FE model within its context of use. Several papers have been written in the last decade highlighting the importance of V&V and providing guidelines for these methods in the field of biomechanics (Anderson et al., 2007; Erdemir et al., 2012; Henninger et al., 2010; Viceconti et al., 2005). These authors have also helped to reinforce a standard vocabulary around V&V. Namely, verification is the process of confirming the computational methods solve the governing equations correctly and accurately. Validation involves evaluating how accurately the model simulates the real physical system of interest.

Jones and Wilcox presented a specific framework for V&V in the spine (Jones and Wilcox, 2008), and highlighted the importance of presenting verification results in the form of FE mesh convergence. However, relatively few studies based on segment FE models of the spine have reported detailed investigations of mesh convergence. Some studies of the lumbar spine have

1 evaluated the FE mesh qualitatively (Eberlein et al., 2004, 2001; Tsouknidas et al., 2013), but the
2 study by Ayturk and Puttlitz is one of the few to report a detailed quantitative mesh convergence
3 analysis (Ayturk and Puttlitz, 2011). They evaluated a multi-segment lumbar spine model using
4 multiple bending modes and outcome metrics with a convergence criterion of 5%.

5 Most validation studies for segment FE models of the spine have also used indirect validation
6 (Jones and Wilcox, 2008). Indirect validation for a spine FE model usually involves comparing
7 FE output metrics to a range of experimental data from multiple specimens to demonstrate that
8 the FE predictions fall within a reasonable experimental spread. This is in contrast to direct
9 validation which typically involves developing an FE model of a specific specimen and
10 comparing FE results to experimental data from the same specimen. Direct validation of segment
11 models of the spine presents significant challenges as both the geometry and material properties
12 of a specific specimen need to be known to develop an accurate subject-specific FE model.

13 The purpose of the current study was to seek broad verification and validation of lumbar
14 spine FE models created using a previously published automated algorithm (Campbell and
15 Petrella, 2015). We defined our context of use as: prediction of torque-rotation response, disc
16 pressure, and facet force in flexion/extension, axial rotation, and lateral bending, under typical *in*
17 *vivo* simulated loads. A detailed, quantitative mesh convergence study was conducted to verify
18 appropriate mesh density created by the automated method. Semi-direct validation was
19 performed by using calibrated material properties and precise boundary conditions for a single
20 specimen from a prior experimental study (Coombs et al., 2013; Rao, 2012). A thorough indirect
21 validation was also performed for 18 full lumbar FE models using generalized material
22 properties and a range of experimental outcomes reported in the literature.

2. Materials and methods

2.1. FE Model Details

Briefly, the automated algorithm used to create the FE models in this study accepts STL geometry of lumbar vertebrae as input and identifies 1,306 landmarks characterizing the key biomechanical features of each bone, such as endplate contours, ligament attachment points, and facet contact surfaces. Landmark data are then used as the basis for automated fitting of a pre-existing template FE mesh (Coombs et al., 2013; Rao, 2012) to the subject-specific bone geometry. All models are oriented with the vertical axis of L4 aligned with gravity. Lastly, soft tissues are also added to create a complete Abaqus (Simulia, Johnston, RI, USA) FE model of the specimen. The automated algorithm runs in Matlab (MathWorks, Natick, MA) and requires approximately 90 minutes to complete for a typical multi-segmental lumbar specimen (L1-L5). Additional details of the automated methods may be found in (Campbell and Petrella, 2015).

Three specific workflows were used to evaluate the FE models created by the automated algorithm (Fig. 1). Specimen geometry, material properties, and the FE solver were chosen deliberately for efficiency and for direct comparison to previous work where appropriate. A single representative L4-L5 functional spinal unit (FSU) was used for mesh convergence (Fig. 1a). Semi-direct validation was performed using the automated method with bone geometry and calibrated material properties (Fig. 1b), both of which were obtained from a prior study (Rao, 2012). And, indirect validation was based on a comparison of results from automated FE models of 18 individual specimens to a range of data from the literature (Fig. 1c). For mesh convergence and indirect validation, a set of generalized material properties was synthesized from the literature (Table 1).

1 The bones and endplates were represented by 7470 elements total for each vertebra and were
2 modeled as rigid, similar to several previous studies (Cegoñino et al., 2014; Coombs et al., 2013;
3 Dreischarf et al., 2014; Little et al., 2007; Moramarco et al., 2010; Rao, 2012). Since the bone
4 elements were rigid, their only function in the model was for visualization of the bone surface
5 and confirmation of correct soft tissue attachment. Seven relevant spinal ligaments were
6 represented using sets of nonlinear tension-only connector elements as described in Campbell
7 and Petrella (2015). The nonlinear ligament properties were based on exponential fits from the
8 literature (Ayturk and Puttlitz, 2011; Nolte et al., 1990; Rohlmann et al., 2006).

9 The annulus fibrosis was modeled using the Holzapfel-Gasser-Ogden material formulation in
10 Abaqus with 3108 linear hexagonal elements per disc. This material model allows a continuous
11 hexagonal mesh to represent a NeoHookean ground matrix with fiber reinforcement. The
12 continuous mesh is of particular importance using the automated methods because the shape and
13 size of the disc elements change depending on the specimen geometry. The Holzapfel-Gasser-
14 Ogden material model allows fiber angles to be defined numerically so that orientation of the
15 fiber network is maintained irrespective of morphological changes in the continuum. Hybrid
16 elements were used for the annulus fibrosis to allow incompressible behavior without shear
17 locking (Eberlein et al., 2004, 2001). The annulus fibrosis matrix properties were derived from
18 Eberlein et al. (2004, 2001). The annulus fibrosis properties were divided into anterior, lateral,
19 and posterior regions (Rao, 2012). A different fiber stiffness was defined in each region
20 (Eberlein et al., 2004, 2001; Malandrino et al., 2013) and the fiber angle was based on the
21 average angle for each region in a regression model (Holzapfel et al., 2005). The generalized
22 material parameters are summarized in Table 1.

The nucleus pulposus was modeled as a fluid cavity in Abaqus using 2154 quadrilateral reduced integration surface elements (SFM3D4R) per disc to define the nucleus boundary. The fluid cavity definition in Abaqus represented the nucleus pulposus with homogeneous fluid pressure. The cavity was defined with the bulk modulus of water based on the template model (Rao, 2012) and consistent with other fluid cavity models from the literature (Charles et al., 2013; Niemeyer et al., 2012). The facet cartilage was modeled using rigid hexagonal elements with 864 elements for each section of superior cartilage and 336 elements for each section of inferior cartilage. Rigid contact surfaces were created based on the geometry of each set of rigid cartilage elements. The facet contacts were defined using frictionless softened contact with a linear pressure-overclosure relationship to define interaction of the sets of rigid cartilage surfaces. The initial gap between the cartilage surfaces was defined relative to the subchondral bone as identified on the CT for each specimen (Campbell and Petrella, 2015). The ligament endpoints, superior, and inferior surfaces of the discs, and interior surface of the cartilage elements were all rigidly fixed to the bones where they attached.

2.2. *Mesh Convergence Study*

The mesh convergence study was performed to confirm adequate mesh density such that the key variables in our context of use were not influenced significantly by mesh density. The mesh was considered converged if the key output variables changed less than 5% when the mesh density was doubled. A single L4-L5 FSU from one arbitrarily chosen auto-generated FE model was used. Pure moment loads up to 10 Nm were applied to the model in each of the six primary bending modes (flexion, extension, left/right axial rotation, and left/right lateral bending). Rotation, disc pressure, and facet contact force were used as output variables.

2.3. *Semi-direct Validation Study*

The semi-direct validation study was conducted to evaluate the FE automation method by comparing simulation results to experimental measurements and results from a manually created FE model. The semi-direct validation comparison specifically assessed the ability of the automation algorithm to create a valid, properly configured subject-specific FE model. Material properties were obtained from a prior study (Rao, 2012) that calibrated an FE model to experimental tests on a lumbar spine with progressive sectioning of the ligaments. The experimental study reported torque rotation curves for each level of the L1-S1 lumbar spine in all six primary bending modes for both the experiment and calibrated model. In the current study, CT data for the same specimen was used as input to the automated FE model generation method (Campbell and Petrella, 2015). The published method was designed for L1-L5 models, so the method was adapted for use with S1 for the semi-direct validation study only. The material properties and boundary conditions were identical to the Rao model, and the simulations were run in Abaqus Explicit to replicate their methods. The sacrum was fixed in space and pure torque was applied to L1. A 450 N follower load was applied during flexion and extension, consistent with the experiments. An example of the model in each of the six bending modes is shown in Figure 2. The torque-rotation results of the current study were compared to the experimental results and to Rao's FE results. RMS (root mean squared) error was calculated for each bending direction. RMS error was reported for the full L1-S1 rotation and an average RMS error was reported for the FSUs by averaging the RMS error for rotation of each of the five FSUs.

2.4. *Indirect Validation Study*

1 The indirect validation study was conducted to broadly evaluate the biomechanics of models
2 created with the automated algorithm by comparing a range of outcomes from several models
3 (i.e., several specimens) to experimental data from the literature. This evaluation also provided
4 an opportunity to compare outcomes from the auto-generated models to other lumbar spine FE
5 models in the literature.

6 Eighteen full lumbar FE models (L1-L5, see Fig. 3), were created using the automation
7 algorithm. Inputs to the automated algorithm were STL models of bone geometry manually
8 segmented from anonymized CT scans of varying resolution and unknown demographic
9 background. The same set of generalized material properties described above (Table 1) was used
10 for all 18 models. The outcome metrics of interest for the indirect validation were rotation, disc
11 pressure, and facet force. A set of 12 simulations was run for each of the models consistent with
12 other recently published validation studies (Ayturk and Puttlitz, 2011; Dreischarf et al., 2014).
13 Six simulations were conducted to evaluate the six pure bending modes to *in vitro* experimental
14 data up to 7.5 Nm. Six additional simulations were conducted to compare the model to *in vivo*
15 experimental data. The simulated *in vivo* loading conditions were based on Dreischarf et al.
16 (2014) and included one pure compression run and bending runs with follower loads in all of the
17 six primary modes except flexion (Dreischarf et al., 2014). Flexion was excluded because
18 Dreischarf et al. concluded that their originally proposed torque was not consistent with *in vivo*
19 loading.

20 The results of the indirect validation study were compared to a range of available data in the
21 literature that contained the variables of interest and replicated the simulated loading conditions.
22 Both *in vivo* and *in vitro* data were used for torque-rotation and disc pressure. Only *in vitro* data

were used for facet force. In total 186 experimental data ranges were used for comparison to each of the 18 FE models.

3. Results

3.1. Mesh Convergence Results

The mesh convergence results showed the percent change between the original mesh and dense mesh for each of the key variables of interest. There was less than a 5% difference in each of the key variables between the two mesh densities, and most variables had less than a 2% change (Figure 4). The original mesh, therefore, was found to produce well-converged results with element edge lengths of approximately 1-1.5 mm throughout the disc and facet contact interfaces.

3.2. Semi-direct Validation Results

The semi-direct validation involved a comparison of the automated FE model from the current study to the experimental results and the FE results from Rao (2012). The torque rotation curves are presented for each of the three datasets in all six bending modes (Figure 5). The rotations were also evaluated at each of the five FSU levels. The RMS error for the automated FE model for L1-S1 rotation ranged from 0.6-1.1 degrees versus Rao 2012 and 1.7-3.3 degrees versus the experiments (Table 2). The RMS error for the FSUs averaged over all five levels ranged from 0.1-0.2 degrees versus Rao 2012 and 0.3-0.6 degrees versus the experiments (Table 2).

3.3. Indirect Validation Results

The indirect validation study includes results from 216 simulations; 12 runs for each of the 18 automatically generated L1-L5 FE models. A total of eleven of the 2016 simulations failed to converge at some point in the run. The results of the FE simulations have been compared to several experimental data sets. The torque-rotation results for *in vitro* flexion, extension, and lateral bending show a wide range of responses due to geometrical variation alone (Figure 6). Many of the specimen geometries produced results that fell within the bounds of the experiments (Guan et al., 2007), but several geometries produced results that were stiffer than the experiments (especially in lateral bending) or softer than the experiments (primarily in extension). The experimental data were reported every 0.5 Nm from 0-4 Nm (128 data ranges). The remaining 53 experimental data ranges and corresponding simulations include data on rotation, facet force, and disc pressure (Figure 7). The simulation results for *in vitro* loading showed a wide variation in all of the results due to geometric variation alone; often similar to the ranges reported in the experiments. However, the simulated *in vivo* loading produced a much narrower range than the experiments.

4. Discussion

The current study involved a mesh convergence study, a semi-direct validation of one automatically generated subject-specific FE model, and an indirect validation of 18 automatically generated FE models with subject-specific geometries. The mesh convergence study showed that the original mesh density was sufficient based on differences under 5% (most under 2%) in flexion, extension, lateral bending, and axial rotation for key metrics of rotation, disc pressure, and facet force. A limitation of the mesh convergence is that only one FSU (L4-L5) from one subject was used. While this one FSU represented the mesh topology used in all other FSUs and specimens, changes in geometry could produce different mesh convergence results in other areas.

1 Ideally mesh convergence would be performed on every model produced, but that is usually not
2 practical. Choosing a representative model for convergence is consistent with the methods of
3 other automated modeling techniques in the literature (Bischoff et al., 2014; Galloway et al.,
4 2013).

5 The semi-direct validation results show that the automated method was able to produce
6 results comparable to manual FE modeling methods. With prior calibrated material properties the
7 model produced torque-rotation curves in good agreement with experimental measurements.
8 Results were in good agreement at both the full lumbar level (L1-S1) and for individual FSUs,
9 providing a modest level of hierarchical validation – that is, we observed good agreement with
10 experiment for both constituent parts (FSUs) as well as the full lumbar model. Improvements for
11 semi-direct validation of subject-specific models could be made with modifications to the
12 automated FE algorithm to better treat soft tissues. The facet surfaces were based on the
13 geometry of the subchondral bone. Linked CT and MRI datasets could be used to include
14 subject-specific facet cartilage thicknesses and to include subject-specific geometry and material
15 properties for the intervertebral discs.

16 The lumbar spine FE models used in this study were created considering our context of use.
17 A relatively simple and efficient FE model setup was employed using rigid bones, rigid
18 endplates, and ligaments meshed with 1-D structural elements. These choices improved
19 simulation run times which was important considering the large number of models and
20 simulations completed. While these models are certainly not appropriate for all applications,
21 similar FE models have been published and validated for situations with relatively low loads
22 (Cegoñino et al., 2014; Dreischarf et al., 2014; Little et al., 2007; Moramarco et al., 2010; Rao,

2012). Validation for higher loads, as studied by others (Shirazi-Adl et al., 2006), or other key variables of interest would likely need to employ more complex modeling methods.

The goal of our indirect validation was to evaluate a set of subject-specific model geometries all using the same general material properties. In the true human population material properties in the lumbar spine vary considerably and affect biomechanics (Lee and Teo, 2005). Therefore, it makes sense that our models do not match the full range of results from the general population because we only included geometric variation in 18 subjects and no material property variation. Similarly, it is reasonable that not all of our results remain within the bounds of the experimental data because often there were fewer subjects in the experimental tests than there were in the simulations. The relatively wide distribution of our results, all using the same material properties, is an indication that the 18 specimens modeled exhibit a relatively wide range of geometric variation, and this variation was responsible for substantial variability in the biomechanical outcome measures.

Overall we found that our models compared reasonably well to the experimental data. The results suggest that our general material properties may be stiffer than average for flexion and lateral bending and may be softer than average for extension. The response of the models appeared close to the average for axial rotation. However, our goal was not to determine average material properties for the population but to evaluate generic FE models with different geometries. Facet force results showed similar trends to the experimental data in different bending modes. Facet forces were generally higher in the simulations than the experiments for the modes of primary interest: extension and axial rotation. These differences may be due to the FE models and/or reliability of the experimental data. While these experiments are currently the best available data for facet force, they have primarily been aimed at methods to measure facet

1 force, not establishing a range of facet forces in the population. They also have significant
2 potential limitations in terms of accuracy, calibration, and in some cases compromise of the facet
3 joint capsule.

4 Disc pressures in the simulations in general compared favorably to the experimental data for
5 compression and bending modes at multiple spinal levels. The indirect validation results for the
6 automated FE models are comparable to other full lumbar spine FE models in the literature
7 (Ayturk and Puttlitz, 2011; Dreischarf et al., 2014). Dreischarf et al. (2014) published a paper
8 similar to the current study by comparing eight FE models to similar experimental data.
9 However, all eight models were constructed by different institutions, with different methods,
10 using different FE solvers, and different material properties. To our knowledge this is the first
11 study to create and evaluate many full lumbar FE models of different specimens using the same
12 modeling methods.

13 We attempted to validate the models against a broad and thorough dataset. We feel that these
14 results indicate the FE models and automated methods can reasonably be used within the range
15 of validation tests performed (i.e. non-coupled bending modes with and without follower loads)
16 for the metrics tested (i.e. rotation, facet force, and disc pressure). Facet force should be used
17 with caution because the forces predicted were generally higher than the experiments, but the
18 trends appeared to be reliable. Therefore, the model can be considered validated for studies that
19 look at how variations in geometry, material properties, or implants affect trends in the validated
20 output variables within the loads studied.

21 Additional direct or semi-direct validation would be necessary to determine if the models
22 could be used to predict results from a specific subject. We expect that additional model

1 complexity may be necessary to accurately predict subject-specific outputs, even with known or
2 calibrated material properties.

3 Future work will involve creating and evaluating a statistical shape model (SSM) of the
4 lumbar spine and employing the model for probabilistic analysis. One advantage of the method
5 for generating automated lumbar spine FE models is that the geometry can be parameterized
6 through the use of an SSM, something that has not been possible previously (Dreischarf et al.,
7 2014; Sarkalkan et al., 2014). With the SSM and automated methods the primary modes of shape
8 variation in the lumbar spine can be correlated with biomechanical outputs from FE models.
9 Probabilistic methods can also be used to evaluate how variations in shape and soft tissue
10 material properties affect normal biomechanics and how implants work in spines with anatomical
11 shape variations.

12 The current study evaluated lumbar spine FE models created through automated methods.
13 Mesh convergence, semi-direct validation, and indirect validation studies were performed. To the
14 authors' knowledge this is the first study to create and evaluate many full lumbar FE models of
15 different specimens using the same modeling methods. A large range of biomechanical results
16 were produced solely due to geometric variation in the 18 specimens simulated. The studies
17 showed that the automated models can be used to reliably evaluate lumbar spine biomechanics,
18 specifically within our intended context of use: in pure bending modes, under relatively low non-
19 injurious *in vivo* simulated loads, to predict torque rotation response, disc pressures, and facet
20 forces.

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Fig. 1. The three modeling tasks used to support V&V of the automated lumbar spine FE modeling algorithm.

Table 1

Material properties used in the mesh convergence and indirect validation simulations. Studies referenced in table 1: a) Cegoñino et al., 2014; b) Dreischarf et al., 2014; c) Little et al., 2007; d) Moramarco et al., 2010; e) Rao, 2012; f) Ayturk and Puttlitz, 2011; g) Nolte et al., 1990; h) Rohlmann et al., 2006; i) Eberlein et al. 2004, 2001; j) Malandrino et al., 2013; k) Holzapfel et al., 2005.

Fig 2. Views of the L1-LS1 FE model in flexion/extension, lateral bending and axial rotation.

Fig 3. Images of the 18 automatically generated full lumbar FE models highlighting diversity of size, shape, and alignment.

Fig 4. Mesh convergence results showing the percent change in output variables rotation, disc pressure, and left/right facet force when the model was subjected to left/right axial rotation, left/right lateral bending, flexion, and extension. Comparisons were made between the original template mesh adopted from Rao (2012) and a second mesh in which element density was doubled.

Fig 5. Semi-direct validation torque rotation curves for L1-S1. Experimental data (Experiment), FE results from the current study (Automated-FE), FE results from Rao (Rao2012).

Table 2

RMS error results of the semi-direct validation study (in degrees) for L1-S1 and FSU (average RMS error over all five FSUs).

Fig 6. Torque rotation curves for flexion, extension, and right/left lateral bending for all 18 FE models (black) and the experimental results from Guan et al. (2007), mean (solid red), ± 1 standard deviation (dotted red).

Fig 7. Indirect validation results. Modes: range of motion flexion-extension (FE), lateral bending (LB), axial rotation (AR); one-direction bending flexion (F), extension (E), lateral bending (L), axial rotation (A). Moment (M), Follower Load (F), number of subjects (N), simulation (Sim), experimental (Exp). Authors: a) (Rohlmann et al., 2001), b) (Percy and Tibrewal, 1984; Percy, 1985; Percy et al., 1984), c) (Wilson et al., 2006), d) (Sawa and Crawford, 2008), e) (Niosi et al., 2008), f) (Wilke et al., 2001), g) (Brinckmann and Grootenboer, 1991), h) (Ayturk, 2007; Ayturk and Puttlitz, 2011), i) (Heuer et al., 2007). Simulation data (black) shown with box plots with mean, standard deviation, and all data points. Experimental data (red) with box plots (mean and standard deviation) and/or whisker plots (median and range).

Figure2

Figure3

Figure4

Figure5

Figure6

Figure 7

Table1

Bones / Endplates	Rigid ^{a,b,c,d,e}					
Ligaments	Tension Only, Exponential Force-Displacement ^{f,g,h}					
Annulus Fibrosis	Matrix		Fibers	Anterior	Lateral	Posterior
	C10 (MPa)	0.25 ⁱ	k1 (MPa)	2 ^j	3 ⁱ	5 ^j
			k2	190 ^j	45 ⁱ	10 ^j
	D (1/MPa)	0 ⁱ	Angle (deg)	26 ^k	34.9 ^k	43 ^k
			Kappa	0	0	0
Nucleus Pulposus	Fluid-Cavity K=2200 MPa ^e					
Facet Cartilage	Frictionless, Softened Contact, Linear Pressure-Overclosure, k=100 MPa/mm					

Table2

RMS Error				
Automated FE	vs Experiment		vs FE Rao 2012	
	FSU	L1-S1	FSU	L1-S1
Flexion-Extension	0.6	3.3	0.2	1.1
Lateral Bending	0.3	1.7	0.2	0.8
Axial Rotation	0.3	2.1	0.1	0.6