Parametric Patient-Specific Finite Element Modeling of Lumbar Spine Based on Anatomical Parameters

Abstract

Background: Low back pain is one of the most common problems that force individuals to seek medical care. Since surgery is the last treatment strategy, predicting the process of conducting surgical procedures seems beneficial and somehow crucial. In this regard, the first step is having a validated biomechanical model based on the anatomical parameters of patients’ lumbar spine. Despite the impressive progress in this field, there is still a need to designing a model that could include important anatomical parameters and be applicable in terms of clinical applications.

Methods: This study aimed to develop the personalized spinal finite element model with 23 anatomical parameters. The initial data was extracted from the radiology picture of the average healthy volunteers and was designed in Catia software. Afterwards, the finite element model was analyzed in Abaqus, and results of the range of motion of motion segments in movements of flexion, extension and left and right lateral bending were verified based on the results of experimental studies present in the literature.

Results: In order to observe the application of the patient-specific spinal parametric model, a model of a patient after spinal fusion was presented. Moreover, results of the range of motion of the motion segments and intradiscal pressure were compared to the healthy model.

Conclusion: Since acceptable results were obtained at each step, it is possible to predict the result of spinal fusion and compare the biomechanical results in case of decreased or increased fusion level by developing a parametric patient-specific model for each patient, which can be an effective achievement for clinical fields.

Keywords: Spine, Finite Element Analysis, Mechanics, Spinal Fusion, Patient-Specific Modeling

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Introduction

Low back pain is one of the most common conditions that require medical interventions. Due to the high prevalence of low back pain, especially among the active population, this disorder has a significant impact on the economic burden of the community and health care systems. Studies show that the incidence of low back pain in the general population varies between 20%-30% and its risk increases in the age range of 45-65 years. Among the American population, back pain is among the top 50 reasons a person needs health care, and low-back pain costs Americans $30-50 billion in healthcare costs. After the initial treatment for low back pain, spinal fusion is one of the costly therapeutic interventions for this condition.

Although there is still controversy over the best treatment for degenerative lumbar spine disease, reducing the pressure in the posterior compartment, fixation and spinal fusion are among the most important surgical treatments for this disease. Spinal fusion using screws have been applied for many years and have yielded relatively acceptable clinical results.
The number of spinal fusion has significantly increased in recent decades. The complications of this procedure have increased over time in a way that there has been an increased need for reoperation. The failure rate of the surgery to treat one segment of the lumbar spine is 9%-36% and higher for several spinal segments. Over 65% of the reoperations are due to complications from instruments used in the surgery or pseudarthrosis\(^3\). Today, most surgeons perform spinal fusion without predicting the outcomes of the surgery in terms of biomechanics. Prediction of the range of motion is difficult for surgeons to be evaluated before surgery. In modeling of the lumbar spine, presenting a patient-specific model based on the anatomic parameters of patients is significantly important in clinical applications. Finite element analysis is one of the most common methods used in the biomechanical study of the human lumbar spine. This type of analysis has a lower cost and higher efficiency, compared to in-vivo and in-vitro techniques\(^4\). The models presented so far are often based on a mean of dimensions, and the direct use of the model in the clinic is not possible considering the wide range of patients’ anatomy and its significant impact on the outcomes\(^5\).

In addition, in line with the application of the model in clinical studies, a model must be designed that is able to cover clinical details and be personalized for each patient. This type of model is used in clinical discussions and provides the opportunity for surgeons to predict the surgical outcomes in the shortest time possible by updating the model based on the information (e.g., intradiscal pressure and intervertebral rotation) of each patient before surgery. Application of patient-specific models helps us assessing the outcomes of various surgeries and clinically evaluating the results before planning for the operation. With this background in mind, this study aimed to present and validate a personalized spinal model based on important anatomical parameters. Another objective of the study was analyzing the model for the patient before and after spinal fusion to present the application of the model. It should be noted that the dimensions of the model were extracted from the radiology picture based on the important anatomical parameters determined for each patient.

### Methods

**Parametric Modeling and Validating the Healthy Model**

In order to model the healthy lumbar spine parametrically, 23 parameters of important anatomical dimensions of a 20-year-old healthy man were determined, as presented in Table 1. Numerical values of parameters were extracted from the radiology picture of the healthy subject and were entered into an Excel file in the form of a table as the input of Catia software\(^6\). In the design of the lumbar spine and the sacrum region in Catia software, two distinct vertebral segments (cortical and cancellous) and two separate disc segments (annulus fibrosus and nucleus pulposus) were considered for distinguishing properties and more precise modeling (Figure 1). Finally, 22 pieces were placed next to each other in the assembly environment of the model.

<table>
<thead>
<tr>
<th>Table 1. Number of Anatomical Parameters in the Design of the Lumbar Spine</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Region</strong></td>
</tr>
<tr>
<td>Body</td>
</tr>
<tr>
<td>Articular process</td>
</tr>
<tr>
<td>Spinous process</td>
</tr>
<tr>
<td>Transverse process</td>
</tr>
<tr>
<td>Lamina, pedicle and foramen</td>
</tr>
<tr>
<td>Intervertebral disc</td>
</tr>
<tr>
<td>Sacrum region</td>
</tr>
</tbody>
</table>

The 3D geometric model extracted from the Catia model was entered into the Hypermesh software for structured meshing and adding the anterior longitudinal, posterior longitudinal, flavum, intraspinous, supraspinous, and intertransverse ligaments. In addition, collagen fibers were entered into the same software for meshing the model\(^7\).
The meshed model was entered into Abaqus software for defining properties and geometric and contact constraints, as well as loading and final analysis[8]. The properties and cross section area related to the ligaments are presented in Table 3. In addition, the collagen properties were considered according to previous studies[9]. In order to model the annulus fibers, one plane of the annulus elements was separated per each 35 and -35 degrees to the end disc edge. The collagen fibers were considered on the planes with one-millimeter space between them, and the cross section area of each fiber was estimated based on the annulus volume and disc height[10].

The contact constraints between the cortical and cancellous parts of the body, the annulus fibrosus and nucleus pulposus of the disc, and the surfaces between the vertebrae and intervertebral disc were defined. The upper and lower facet joints plane in tangential direction was considered as frictionless surfaces, and the exponential method was used in the normal direction, in a way that the pressure was at 100 MPa at a distance of 0.5 mm between the facet joints. In order to evaluate the accuracy of the model, 10 Nm moment was applied to the upper plane of the first vertebra for modeling the lateral bending, flexion, and extension movements. Moreover, the follower load technique was used for the simulation of human weight and muscle forces in the axial direction of the spine with 500 N for extension and lateral bending and 1175 N for flexion movements[20]. To complete the boundary condition, the degree of freedom of the sacrum region was considered to be zero at all angles. The final model was calculated by static analysis in different loadings, and the results were considered as output in Excel software separately in the form of a set of numbers for intervertebral rotation in each analysis. At the end of the modifications, the final model in each software was equal to Figure 1.

![Figure 1](image_url)

**Application of the Parametric Model in Modeling after Spinal Fusion**

As one of the applications of the validated model, one of the patients evaluated in Chang Gung University, Taoyuan, Taiwan was selected at this step. In this regard, the model was prepared for the patient before and after spinal fusion. It should be noted that the geometrical dimensions of the model were extracted from the radiology picture of the patient after the surgery. In our case, spinal fusion was required in L2-L3 intervertebral level in the lumbar spine due to reduced height in these regions. Similar to the modeling process for a healthy model, only the pre-determined anatomical parameters were changed and updated for the patient and the model preparation process was repeated for analysis similar to the previous round. In order to model spinal fusion, screws, rods, and cages were designed separately based on the dimensions extracted from the radiology picture of the patient, followed by their assembly in the Catia software.
According to Table 3, elastic rods, screws, and cages were considered for the post-operative model with the mentioned values. In spinal fusion, mostly the anterior ligaments are maintained, and the posterior ligaments are removed. Therefore, the posterior longitudinal and flavum ligaments were eliminated. The next steps included the modeling of the collagen fibers of the intervertebral discs, completing the contact constraints, loading, and boundary conditions, which were carried out completely similar to the healthy model. Moreover, the modeling of the lumbar spine of the patient was completed after spinal fusion.

### Results

#### Healthy Model Validation Results

In order to validate the 3D model of the lumbar spine with a laboratory model and verify the model based on the article by Panjabi et al.\(^{(21)}\), results related to the range of

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**Table 3. Mechanical Properties Related to the Modeling of Various Components of the Lumbar Spine in Abaqus software**

<table>
<thead>
<tr>
<th>Lumbar spine components</th>
<th>Mechanical behavior</th>
<th>Mechanical properties*</th>
<th>Cross section*</th>
<th>Number of elements</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical bone</td>
<td>Elastic</td>
<td>E=12000, v=0.3</td>
<td></td>
<td>60252</td>
<td>(11) (12)</td>
</tr>
<tr>
<td>Cancellous bone</td>
<td>Elastic</td>
<td>E=200, v=0.25</td>
<td></td>
<td>10846</td>
<td>(13) (14)</td>
</tr>
<tr>
<td>Annulus fibrosus</td>
<td>Hyperelastic (Mooney Rivlin)</td>
<td>C10=0.18, C01=0.045</td>
<td></td>
<td>7680</td>
<td>(15)</td>
</tr>
<tr>
<td>Nucleus pulposus</td>
<td>Hyperelastic (Mooney Rivlin)</td>
<td>C10=0.12, C01=0.030</td>
<td></td>
<td>776</td>
<td>(13)</td>
</tr>
<tr>
<td>Screw and rod</td>
<td>Elastic</td>
<td>E=110000, v=0.3</td>
<td></td>
<td>6151</td>
<td>(16)</td>
</tr>
<tr>
<td>Cage</td>
<td>Elastic</td>
<td>E=3600, v=0.4</td>
<td></td>
<td>2504</td>
<td>(17)</td>
</tr>
<tr>
<td>Anterior longitudinal ligament</td>
<td>Elastic</td>
<td>E=55.765, v=0.4</td>
<td>32.4</td>
<td></td>
<td>(18)</td>
</tr>
<tr>
<td>Posterior longitudinal ligament</td>
<td>Elastic</td>
<td>E=54.433, v=0.4</td>
<td>5.2</td>
<td></td>
<td>(18)</td>
</tr>
<tr>
<td>Flavum ligament</td>
<td>Elastic</td>
<td>E=3.2494, v=0.4</td>
<td>84.2</td>
<td></td>
<td>(18)</td>
</tr>
<tr>
<td>Intraspinous ligament</td>
<td>Elastic</td>
<td>E=2.2336, v=0.4</td>
<td>35.1</td>
<td></td>
<td>(18)</td>
</tr>
<tr>
<td>Supraspinous ligament</td>
<td>Elastic</td>
<td>E=12.8, v=0.4</td>
<td>25.2</td>
<td></td>
<td>(18)</td>
</tr>
<tr>
<td>Intertransverse ligament</td>
<td>Elastic</td>
<td>E=11.5, v=0.4</td>
<td>12</td>
<td></td>
<td>(19)</td>
</tr>
<tr>
<td>Capsular ligament</td>
<td>Elastic</td>
<td>E=8.6868, v=0.4</td>
<td>43.8</td>
<td></td>
<td>(18)</td>
</tr>
</tbody>
</table>

*The Elastic Modulus Units (E) and the cross section area were expressed in MPa and \(\text{mm}^2\), respectively.*
motion of motion segments of the designed healthy model were compared to the experimental results of the present study for flexion, extension, and left and right lateral bending movements, results of which are presented in Figure 3. Furthermore, results are shown in Table 3 for the validation of the intradiscal pressure along with the numerical and experimental results. According to these findings, the model prediction had a good agreement with the laboratory tests and former numerical studies.

Model Results after Spinal Fusion

In order to observe the application of the patient-specific parametric model of the lumbar spine and the sacrum region, a model of the patient was presented after spinal fusion to be compared to the validated healthy parametrical model. In addition, the results of intervertebral rotation and intradiscal pressure were compared to the healthy model (Figure 3).

Figure 3. Comparison of range of motion of the motion segments of the healthy finite element model with experimental and numerical data at various movement levels: a) extension, b) flexion, c) left lateral bending, d) right lateral bending and comparison of intradiscal pressure for L3-L4 intervertebral disc core in various movements: e) extension, f) flexion, g) left lateral bending, h) right lateral bending; and comparison of the range of motion of the motor ligaments of the healthy finite element model with the model after spinal fusion at various movements: i) extension, j) flexion, k) left lateral bending, l) right lateral bending, and comparison of the intradiscal pressure of the healthy finite element model with the model after spinal fusion for the L4-L5 intervertebral disc core in various movements: m) extension, n) flexion, o) left lateral bending, p) right lateral bending.

(IVR: intervertebral rotation, IDP: intradiscal pressure)
The personalized 3D model for each patient was validated and presented based on the results. According to the results related to the first section of validating the healthy model, it could be observed that the model behavior was within the acceptable range of experimental works for the range of motion of motion segments in flexion, extension, and left and right lateral bending movements. As observed, the diagram is at the lowest standard level for flexion movement, especially at the L4-L5 intervertebral level, which might be due to the considering of the degree of freedom of the total sacrum region at zero. The accuracy of this model can be confirmed by comparing the intradiscal pressure for L4-L5 intervertebral nucleus disc using the numerical and experimental works in this area. According to the results of the second section and after the evaluation of the biomechanical behavior of the lumbar spine after spinal fusion and based on the diagrams presented in this part, it was observed that the results related to the range of motion of the motion segments were within an acceptable range. As expected for the level determined in spinal fusion, the range of motion of motion segments was near zero for all movements and below the healthy model for other levels. These results might be due to an overall decline in the intervertebral rotation range. Comparison of the intradiscal pressure after spinal fusion revealed that in addition to the expected decrease in the intradiscal pressure, a reduction was observed in values of the diagram in all movements, and they were still within the standard range.

Discussion

According to the results of previous studies in this field, we can describe the similarities and differences between the model presented and previous models. Similar to the models presented for each patient, this modeling can be upgraded in relevant software based on a number of anatomical parameters entered into the software as input. In the latest modeling carried out in 2018 (5), all modeling process was performed in MATLAB (24), and the input parameters were often regarded as a mean of the dimensions. Meanwhile, all anatomical dimensions were directly extracted from the radiology picture of each patient in the current research to increase the precision of the modeling. In addition, separate software was used in the design and analysis of each section. Another difference between our modeling and previous works was the use of radiology pictures in lateral and AP views since most precise anatomical modeling in previous works were based on the CT scan picture of patients.

The main objective of the present study was extracting a patient-specific finite element model for patients to biomechanically predict the lumbar spine of the patients before and after the surgery. Therefore, in the first step, the personalized parametrical geometric model was constructed based on 23 anatomical parameters, which can be easily measured using two radiology pictures of patients (AP and Lateral images). Afterwards, the personalized finite element model was developed and validated based on previous experimental and numerical studies.

In order to show the application of the personalized model, the information of a patient was extracted from the radiology pictures, and the results of the model were evaluated after the surgery. This model can have a proper clinical application and can enable surgeons to predict the results of the surgery in the shortest time possible by upgrading the model based on the information of the patient before the surgery by considering information such as intradiscal pressure and range of motion of the motion segments. Such patient-specific models can be used to evaluate the results of various surgeries and clinically assess the results before planning an operation.

Among the limitations of modeling of the model simulated in the present study at
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this step, which emphasized the anatomical changes of the lumbar spine before and after the surgery, were reduced disc and vertebrae height, adding of implants, removal of the components eliminated in surgery, and changes in the curvature of the spine. If specific mechanical changes of tissues, such as disc degeneration in various grades or osteoporosis, are considered, the model must be developed by combining other clinical information. Future works by authors of this paper is focused on this subjects. Another limitation of the present study was lack of accurate simulation of muscle strength that affects the lumbar spine. By adding these details, we will have more accurate results, compared to the real human models. In addition, the geometric dimensions extracted from the radiology pictures of patients are manually entered into the Mimics software, which is slightly time consuming. By adding the image processing module as the next stage, this process will be automatically carried out to facilitate the use of the model in the clinic.

Development of a parametric patient-specific model for each patient will lead to the proper prediction of results of spinal fusion surgery. This modeling can act as a guide to predict the process and result of the surgery and can limit the surgery process from experience to an accurate scientific framework by defining some indicators. This applied model can provide more accurate planning for surgery by comparing the results related to decreasing or increasing the level of fusion before the surgery. In addition, the model might increase patient satisfaction and trust in the procedure. Following the needs and studies performed to develop the model and achieve a more precise and functional model, development of the model will be considered as the next step for predicting biomechanical results of spinal fusion surgeries at various levels and assessing the risk of adjacent level degeneration.

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