# Finite Element Application in Prediction of Suitable Material Properties for an Arthroplastic Lumbar Disc

Azadeh Ghouchani, Mohammad Ravari, and Farid Mahmoudi

*Abstract*—**Diseases of lumbar spine and associated diseases of the intervertebral disc are a major focus of contemporary spinal care. Low back pain, in fact, is becoming in the recent years one of the most diffuse chronic pathologies. Ascribed to the prevalence of low back pain- mostly caused due to disc degeneration, and limitations of current treatments, arthroplasty has been propounded for replacing the degenerated disc. A three-dimensional finite element model (FEM) of the L3-L4 motion segment using ABAQUS v 6.9 has been developed. The annulus fibrosus of the model is idealized as an inhomogeneous composite of an isotropic ground substance, reinforced by helically oriented collagen fibers. The model took into account the material nonlinearities and is imposed different loading conditions. In this study, the model is validated by comparison of its predictions with several sets of experimental data. Disc deformation under compression and segmental rotational motions under moment loads for the normal disc model agreed well with the corresponding in vivo studies. We determined the optimal Young's modulus as well as the Poisson's ratio for the artificial disc under different physiologic loading conditions by linking ABAQUS with MATLAB 2010.a. The results of the present study suggest that a well-designed elastic arthroplastic disc preferably has an annulus modulus of 19.1 MPa and 1.24 MPa for nucleus section and Poisson ratio of 0.41 and 0.47 respectively. Elastic artificial disc with such properties can then achieve the goal of restoring the disc height and mechanical function of intact disc under different loading conditions and so can reduce low back pain.**

*Index Terms*—**Finite element, intervertebral disc, modeling, optimization**

## I. Introduction

Disc degeneration is a natural procedure of ageing and is characterized by changes in the morphology and biochemistry of the disc. Degenerative changes in the disc may progress with aging in a roughly linear fashion. After disc degeneration (DD), changes in segmental flexibility are noted. These changes, in turn, affect the functioning of various other segmental structures [1].

Surgical treatments for disc degeneration can be roughly grouped as fusion, disc replacement and dynamic stabilization. For a relative slight degeneration patient, dynamic stabilization device can be considered, while fusion and disc replacement will be used for severe cases [1]. Currently, spine procedures are aimed primarily at arthrodesis (fusion). The lumbar underbody fusion procedure is an effective and popular surgical technique for treating low back pain related to degenerative disc disease [2]. This procedure restores disc height, enlarges the steno tic foramen, stabilizes the spine, and provides mechanical strength between vertebrae [3]. However, it has been argued that various spinal fusions restrain motion at the surgical level. This local environmental change at the surgical level results in high stress at the adjacent disc levels and accelerates degeneration. Patients may need to undergo another surgery for extended fusion at the adjacent levels. Clinical studies have reported incidence rates ranging from 6% to 58% [3, 4]. Several reports have established that spinal fusion with pedicle fixation accelerates the degeneration of adjacent motion segments because the relative immobility of fused spinal segments transfers stress to adjacent segments [2]. Due to the limitations of current treatments for degenerative disc disease; arthroplastic methods to repair the diseased disc have been proposed [5]. Therefore, a non-fusion artificial disc was developed to solve the adjacent segment problems. Briefly, a typical arthroplastic approach includes replacement of the intervertebral disc with an implant an artificial disc.

The artificial disc is a mobile implant for degenerative disc replacement that attempts to lessen the degeneration of the adjacent elements following interbody fusion procedures. Currently, two types of artificial disc, ball-and-socket and mobile core, are in the market. Prospective randomized clinical trials for the ball-and-socket type ProDisc II (Synthes, Inc., Paoli, PA/Spine Solutions, New York, NY) comparing the fusion device and disc arthroplasty under the Food and Drug Administration Investigational Device Exemption showed that this dynamic stabilizer was safe for use and had a good outcome [6].

Today, a number of finite element (FE) analyses and cadaver studies have attempted to evaluate the adjacent effects for artificial discs or to compare artificial discs with fusion. Goel et al. [7] found that the Charité slightly increased motion and facet loading at the implanted level compared to adjacent segments, while loadings at the adjacent levels decreased with use of a hybrid method. In the study of Grauer et al. [8] a finite element model of a L3–S1 segment was used to compare the biomechanical effects of the Charite' artificial disc (ChD) placed at two lower levels (2LChD model) with L5–S1 fusion using a cage (CG) and a pedicle screw system plus L4–L5 level ChD placement combination (CGChD). The changes at L3–L4 level for both of the cases were of similar magnitude (approximately 25%), although in the CGChD model it increased and in the 2LChD model it decreased. The changes in motion at the L4–L5 level were

Manuscript received November 9, 2011; revised December 12, 2011.

A. Ghouchani is with AmirKabir University of Technology (Tehran Polytechnic), PhD student at the department of biomedical engineering department, specialization in biomechanics, Tehran, Iran. She is the corresponding author. (E-mail: Azadeh_gh_e83@ yahoo.com).

M. Ravari is with the department of biomedical engineering, Mashhad branch, Islamic Azad University, Mashhad, Iran as an instructor (E-mail: m1ravari@yahoo.com).

F. Mahmoudi was with the Mechanical Engineering Department, Ferdowsi University of Mashhad, Iran. But is now with knowledge and technology management department, AmirKabir University of Technology, Tehran, Iran (E-mail: Farid_m@aut.ac.ir).

large for the CGChD model as compared with the 2LChD model predictions (approximately 70% increase vs. 10% increase). The biomechanics of the lumbar spine treated either by fusion or total disc replacement (TDR) under severe loading conditions was compared by Denoziere and Ku [9] using a L3–L5 model. Rohlmann et al. [10] also found the importance of artificial disc position. Using a L1–L5 model, they examined how the mechanical behavior of the lumbar spine is affected by the height and position of a ProDisc prosthesis. Chen et al [11] compared inserting the ProDisc II (ADR) with bilateral posterior lumbar interbody fusion (PLIF) cages with a pedicle screw fixation system in L3-L4 level. The success of this arthroplastic approach depends on the restoration of the mechanical function of intervertebral disc. Thus, it is valuable to study the mechanical performance of the artificial disc after implanting into the spine [5]. This is not a simple undertaking as it is difficult to directly measure mechanical behavior of the disc and the implanted device in vivo. Although in vitro studies mimicking in vivo situations could be helpful, it is difficult to control key variables of specimen properties, and thus the application of in vitro experimental methods is limited [12].

For this reason, finite element modeling of in vivo clinical situations is particularly useful. Furthermore, experimental methods have typically been limited to assaying gross vertebral body motions, annular bulges, and intradiscal pressure increases. Alternatively, model analyses of the disc can provide details of behavior not easily measured, such as entire stress and strain fields. In principle, they also lend themselves readily to parametric studies, so that the influences of geometry and material property variations on behavior can be gauged [5]. In the previous studies, the focus was on the investigation and examination of in-market artificial discs. While, in this study, the focus is on designing an optimized artificial disc capable of mimicking intact disc behavior under physiologic loading. Since the L3–L4 and L4-L5 intervertebral discs are commonly associated with low back pain, a three-dimensional (3-D) nonlinear finite element model (FEM) of the L3–L4 disc was constructed and used to determine the mechanical behavior of artificial intervertebral disc [5]. Specifically, we analyzed material properties that are required for the elastic artificial disc, such as the modulus of elasticity and Poisson's ratio and also the mechanical behavior of implanted disc under different loading conditions. Optimization was done by linking MATLAB software with ABAQUS software.

## II. Methods and Materials

### A. Geometry

Recent advances in computing technologies both in terms of hardware and software have helped in the advancement of CAD in applications beyond that of traditional design and analysis. CAD is now being used extensively in biomedical engineering in applications ranging from clinical medicine, customized medical implant design to tissue engineering. This has largely been made possible due to developments made in imaging technologies and reverse engineering techniques supported equally by both hardware and software technology advancements [13].

The primary imaging modalities that are made use of in different applications include, computed tomography (CT), magnetic resonance imaging (MRI), optical microscopy, micro CT, etc. each with its own advantages and limitations. The geometrical representation of the model was obtained using thin slice (1mm) Computed Tomography (CT) images. Acquired data from the lumbar region (L3- L4) were imported into image processing software Amira 4.1 (Visage Imaging, United States) for segmentation.

Using this CT scan, a 3-D finite element model was generated via modeling software ABAQUS (ABAQUS 2009, version 7.2) consisting of vertebra-disc-vertebra unit using a CAD station (Fig 1). This existing model of the L3–L4 motion segment was then modified to include the annulus fibers and material properties. All collagen fibers were simulated by two node link elements with resistance tension only, and they were arranged in the anatomical direction given by the textbook [14].The fibers were modeled using Abaqus Command by writing Input File. Other parts were modeled using Abaqus CAE.

To model the artificial disc, the disc nucleus was replaced by an elastic material and all annulus layers were also replaced by another elastic material. Apart from the mechanical properties, the structure and shape of two parts were assumed to remain unchanged. This was done to highlight the role of mechanical properties on disc biomechanical behavior. In addition to appropriate material properties, we analyzed the mechanical behavior of implanted disc under different physiologic loads.

### B. Meshing

The vertebral bodies and the intervertebral discs were meshed using the mapped mesh approach. 3D, isotropic, 8-noded solid/brick elements were used in the modelling of the cortical shell, trabecular core, bony and cartilaginous end-plates, and the annulus matrix of the intervertebral disc. The thickness of the cortical shell varied for each vertebral body and was set to range from 1.5 to 2 mm. These values were based on the CT scan measurements.

A combination of element types was used for the intervertebral disc. The annulus fibrosus was defined by four radial layers and modeled as a composite material consisting of fibers embedded in a ground substance [15]. For the fibers, 3D truss elements were aligned in layers to form a criss-cross pattern placed at an average angle of ±30° to the horizontal plane of the disc. The nucleus pulposus was simulated as an incompressible material and represented by 8-noded hydrostatic fluid elements with a Poisson's ratio close to 0.5. The ground substance of the annulus fibrosus (the annulus matrix) was represented by 8-noded solid/brick elements. The model consists of 11,368 volume elements, about 450 hydrostatic fluid elements and 105 spring elements and has nearly 75,000 degrees of freedom.

### C. Material Properties

The nucleus pulposus (NP) is a gelatinous fluid filled cavity inside the annulus fibrosus (AF). The annulus fibrosus (AF) was modeled as a matrix of homogeneous ground substance reinforced by AF fibers. The fibers were assumed to take up 38 percent of the AF volume [5], and [16]. The material properties used in the numerical simulations are listed in Table 2. These properties were previously

determined experimentally and were shown to predict well the mechanical behavior of the disc. The model of the artificial disc was achieved by replacing NP with solid elements and all AF layers with solid elements which have elastic properties.

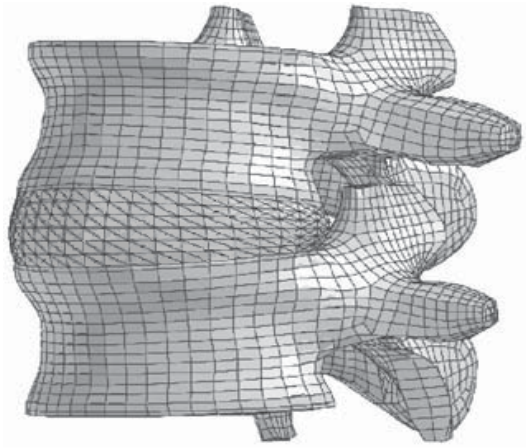

Fig. 1. A three-dimensional finite element model of a L3–L4 motion segment

### D. Boundary Conditions

The lumbar L3–L4 intervertebral disc was loaded with forces and moments, applied to the center of the mass of the L3 top surface, whereas the L4 was fixed at the bottom in all directions so that permitted free rotation in three principal axes but the translation in all three axes was constrained. Three loading conditions were considered: the axial compression, the torsional rotation in the sagittal plane and the lateral bending.

Axial compression was modeled by applying a concentrated force to a reference point above center of mass in which the top surface of L3 is coupled to. The lumbar disc was subjected to axial torque by applied rotation moment to the L3 top surface. A lateral bending moment was applied to the L3 top surface such that the moment is around longitudinal axis (y-axis). Fig 2 schematically depicts the loading conditions.

The mechanical performance of the artificial disc after implantation was evaluated and the results were compared to the response of a normal disc.

## III. Model Validation

A simulation model of the lumbar spine is a powerful tool to predict biomechanics performance, and to design orthopedic implants. This can be done without using costly and time consuming cadaveric biomechanics tests. Some examples include total disc replacements, interbody fusion devices, pedicle screw and rod systems, and interspinous spacers. The model must be validated according to actual biomechanics data to have confidence in predictability.

Biomechanics testing of the spine is currently used to evaluate the performance of spine implants such as total disc replacements and fusion systems. This is typically done by applying loads such as compression and moments to multi-segmental spine or a functional spinal unit (FSU). The intact condition is tested first, and then the spine is instrumented with implants and tested again [17].

To illustrate the capabilities of the model and validate its physiological similarity, the FEM is subjected to three physiological loads; axial compression, axial torsion, and lateral bending. After validating the finite element model, the same loading conditions are applied to the unit motion segment with implanted device to optimize the device mechanical properties (Young's modulus and Poisson's ration). Root mean square errors and the maximum absolute errors were computed between model predictions and available literature (experimental or model) data.

### A. Compression

The accuracy of the FEM was evaluated by subjecting a normal disc to a physiological compressive load of 400N. The model predicted a vertical displacement (reduction in height) of 0.49 mm under compressive load of 400N, which is in good agreement with published experimental results of 0.51mm with a standard deviation of 0.24mm. For the same applied compressive load, the model predicted an intradiscal pressure of 0.320MPa. This predictions fall within the measured range of experimental values from 0.203 to 0.430MPa. The model also predicted a radial outward bulge of 0.21mm.This value once again agrees well with that experimentally measured 0.42±0.25mm [5], [18].

### B. Torsion

A model study of the lumbar disc subjected to axial torque is also important, as it will help correlate the type and magnitude of loads with the degenerative process and failure of the disc. A torsional rotation of 5Nm was also applied. The model predicted an end rotation of 1.48o.

Since an axial rotation of 1.83o with a standard deviation of 0.44o has been reported in response to a 5Nm torque in previous experimental measurements [5], the model predictions fall within the measured range.

### C. Bending

A lateral bending moment of 5Nm around longitudinal axis (y-axis) was applied to the upper surface of L4 vertebra. In a previous study, it was reported that lumbar intervertebral discs had a mean endplate tilt of $2.8^{o}$ in response to a 5Nm moment [5], [19]. The model predicted an endplate tilt of $2.5^{o}$ for the normal disc.

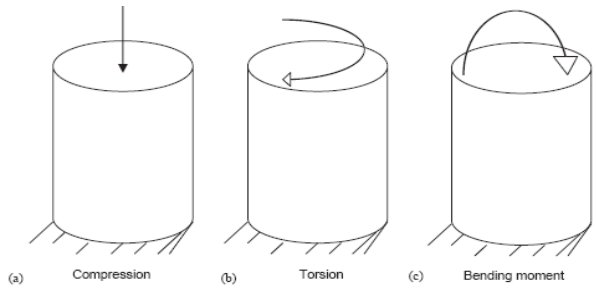


Fig . 2. Loading conditions in the model of L3–L4 intervertebral disc; a) compression, b) torsion, and c) bending moment.

## IV. Optimization of the Model

Optimization was done via MATLAB 2010.a for 4 parameters Ea, ua, En, un indicating young modulus and Poisson's ratio of central part (NP) and peripheral part (AF), respectively. ABAQUS, by giving graphical FEM in the form

of input file, was linked to MATLAB. To suggest ideal material properties for the implant, the effect of the variation of the modulus and Poisson's ratios on several parameters such as vertical displacement, axial rotation at the sagittal cross section, disc tilt, and disc bulge were evaluated.

We got the optimized mechanical properties through 1097 variations of parameters. In all these 1097 cases, we determined disc behavior [21]. According to the minimum error for mimicking intact disc behavior, the corresponding parameters were chosen as optimized ones (Table2).

TABLE 1: MATERIAL PROPERTIES USED IN THE INTACT MODEL [5], [20 ]

| Material | Elastic Coefficient | Constitutive Relation |
|---|---|---|
| Vertebral body shell (cortical bone) | $E_x$=700 $u_{xy}$=0.45<br>$E_y$=700 $u_{xz}$=0.315<br>$E_z$=1000 $u_{yz}$=0.315<br>$G_{xy}$=241.4<br>$G_{xz}$=241.4<br>$G_{yz}$=241.4 | Transversely isotropic, linear elastic |
| Vertebral body core (cancellous bone) | $E_x$=140 $u_{xy}$=0.45<br>$E_y$=140 $u_{xz}$=0.315<br>$E_z$=200 $u_{yz}$=0.315<br>$G_{xy}$=48.3<br>$G_{xz}$=48.3<br>$G_{yz}$=48.3 | Transversely isotropic, linear elastic |
| Cartilaginous endplate | E=23.8 u=0.4 | Isotropic, linear elastic |
| AF ground substance | 4.2 u=0.45 | Isotropic, linear elastic |
| AF fibers | 357.5-550 u=0.35 | Isotropic, nonlinear |
| Nucleus pulposus | κ*= 1666 | Nearly incompressible fluid |

TABLE 2: OPTIMUM PARAMETERS AND IMPLANT PERFORMANCE UNDER THREE LOADING CONDITIONS

| Loading | Compression | Torsion | Bending |
|---|---|---|---|
| En (MPa), Ea (MPa) | 1.45, 5.1 | 1.21, 30.4 | 1.06, 12.3 |
| un, ua | 0.499, 0.33 | 0.485, 0.28 | 0.44, 0.34 |

## V. RESULTS

The minimum error in each case and the corresponding elastic parameters is calculated. In compression, the optimum value for Ea was calculated 5.1MPa which is very close to AF ground substance modulus (4.2MPa), because annular fibers have no resistance in compression and in this loading condition, it seems as there were no fibers. But, in torsion and lateral bending that some of the fibers are in tension, Ea got a higher value depending on the amount of fibers being tensioned.

The variation of the axial rotation at the sagittal cross section on the top level of the artificial disc was calculated as a function of arthroplastic implant modulus: these results are shown in Fig. 3, 4. It should be noted that the model predicted an axial rotation of -2.606e-02 rad (1.49 degree) for the artificial disc, when the modulus of implanted device was selected from the range 29.5-30.5MPa for peripheral part and 1-1.3MPa for central part. Such high value for Ea, however, is not acceptable under compression as was shown before. The difference in axial rotation between intact and arthroplastic disc in the means of Poisson s ratio of both parts of artificial disc is depicted in Figs 5 and 6 [22].

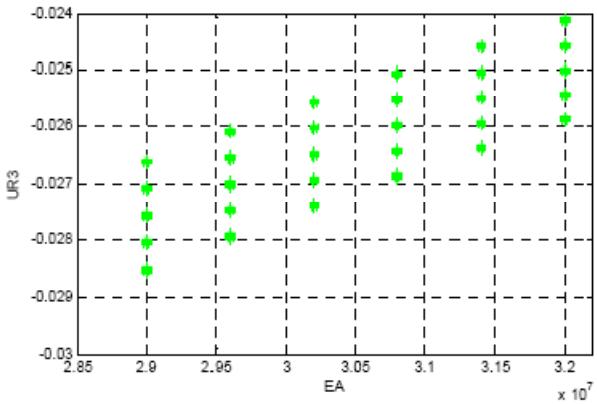


Fig. 3. Axial rotation (UR3) versus the young modulus of peripheral part (EA) of the artificial disc.

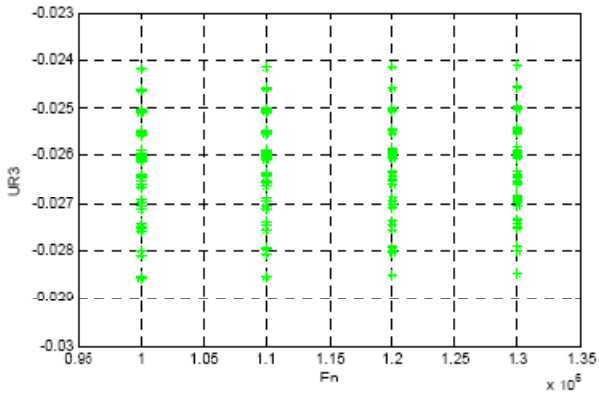


Fig. 4. Axial rotation (UR3) versus the young modulus of central part (En) of the artificial disc.

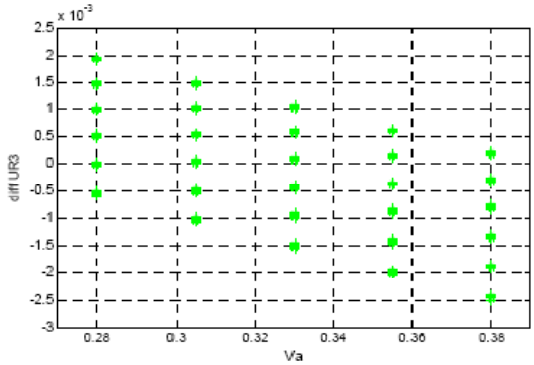


Fig. 5. Difference in model predictions of axial rotation (diff UR3) against variation of poisson s ratio of peripheral part (Va) under the axial torsion of 5Nm.

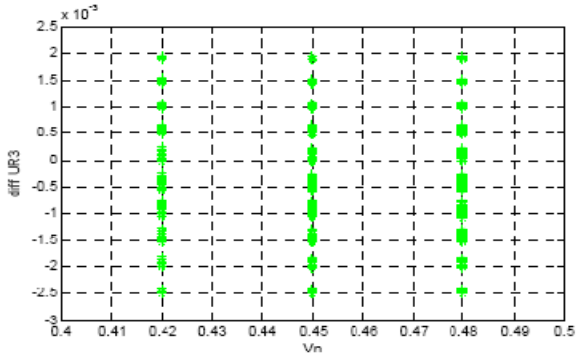


Fig. 6. Difference in Model predictions of axial rotation (diff UR3) against variation of POISSON s ratio of central part (Vn) under the axial torsion of 5Nm

These graphs and calculations are done for each parameter in each loading condition. With regard to the results of model validation and artificial disc behavior under the loading condition, and the graphs showing the difference between intact and artificial disc, appropriate values for the parameters corresponding to the least difference are determined.

Artificial disc should mimic intact disc behavior in all loading conditions such as compression, torsion, bending and even shear. To achieve this goal, the optimum parameters calculated in each case of loading are tested in other loading conditions to determine errors in other loading conditions. By RMS method optimum elastic properties for artificial disc and so the minimum error for all loading conditions are determined.

The results predicted elastic modulus and Poisson's ratio of 19.1 MPa, 0.32, 1.24 MPa and 0.475 for peripheral and central part, respectively. Table 3 shows the behavior of optimized artificial disc under three loading conditions. In compression, artificial disc predicts less reduction in height and outward bulge. If an artificial disc shows more outward bulge in compression to the same compression load, this may press the peripheral nerves and cause pain.

Because the optimum value determined for Ea is more than Young modulus of annulus ground substance (considering little effect of fibers in compression). In lateral bending, one side of the disc is in tension and the other side is in compression. The compression side has a low young modulus, although the tension side has a high one because of tensioned fibers. Axial rotation of artificial disc is more than intact disc which gives wider range of motion and Ea was determined about 30MPa. But to minimize errors in three loading conditions, Young modulus for Ea was determined 19.1MPa. By lowering the modulus, the displacement grows.

TABLE 3: BEHAVIOR OF ARTHROPLASTIC DISC UNDER PHYSIOLOGICAL LOADS

| | Compression: -400N (z) | Bending: 5Nm(y) | Torsion: 5Nm (z) |
|---|---|---|---|
| Artificial disc | U3=-3.8e-04 m, U2=3.96e-05m | UR2=2.9e-02 rad | UR3=3.208e-02 rad |

## VI. CONCLUSION

The finite element method can be a powerful tool in the field of spinal research. It allows us to repeat experiment, to change parameters, thus analyze the influence of a single component within the construct investigated. It is useful in analyzing stress patterns of lumbar, also leading to an optimal design of the surgeon [23]. It does, however, not mean that biomechanical in vitro approaches should be replaced by such a model.

The current finite element model also has limitations, even if its modeling is based on the characteristic of physiological material and the geometric shape of lumbar. The anatomic structure of spine is complicated, and such properties of the small articulation as friction coefficient were not very clear. So all the material parameters adopted for the model were simplified or based on hypothesis on some degree. For example all solid components including cortical and cancellous bone, annulus ground substance were simulated as linear, isotropic material properties. It is a simplified model. In the future study, more nonlinear material will be modeled.

A three-dimensional finite element model of L3-L4 motion segment was modeled. The model was checked for sensitivity to the input parameter values and found to give reasonable behavior. The finite element model displacement predictions under different loads were consistent with experimental results, thus validating the model. We calculated the minimum error in displacement for each loading condition and so the corresponding elastic parameters.

In this study, elastic properties of an artificial disc were such optimized that an implant with the aforementioned mechanical properties can mimic intact disc mechanical behavior with errors less than tenth of millimeter in compression, and thousand of radian in torsion and bending.

To further simulate the realistic situation, future investigations could extend current model to a full lumbar spine model, and also include ligaments and facet joints in the model. Since muscles are known to exert large forces on the spine, a model could be constructed to incorporate muscle force effects and study the role of the muscles in modifying vertebral body motion, disc bulge, intradiscal pressure, facet loading and ligament tension.

The results from this computational model could be used for the assessment of the biomechanical compatibility; to select the most suitable material of the device; and consequently cure disc degenerative disease and low back pain whose most cause is disc degenerated disease.

## ACKNOWLEDGMENT

The authors thank Engineers Elham Ghouchani and Meisam Soroush for their generous technical support with

these two softwares, and express their appreciation to Dr. Kobravi, and Dr. Tabatabai for their helpful comments and suggestions.

## REFERENCES


[1] Q. H. Zhang and E. C. Teo, "Finite element application in implant research for treatment of lumbar degenerative disc disease," *journal of Med Eng Phy*, vol 30, pp.1246-1256, 2008

[2] P. Fritzell, *et al*. "Lumbar fusion versus nonsurgical treatment for chronic low back pain—a multicenter randomized controlled trial from the Swedish lumbar spine study group," *J Spine* ,vol 26(23), pp.2521–34, 2001

[3] G. Denoziere and D. N. Ku, "Biomechanical comparison between fusion of two vertebrae and implantation of an artificial intervertebral disc," *J Biomech* , vol 39(4), pp. 766–75, 2006

[4] S. D. Kuslich, G. Danielson, J. D. Dowdle, J. Sherman, B. Fredrickson, and H. Yuan, *et al*. "Four-year follow-up results of lumbar spine arthrodesis using the Bagby and Kuslich lumbar fusion cage," *J Spine,* vol 25, pp.2656–62. 2000

[5] J. Yao, R. T. Sergio, and P. Ducheyne, "A Three-Dimentional Nonlinear Finite Element Analysis of the Mechanical Behavior of Tissue Engineered Intervertebral Discs under Complex Loads*"J. Biomaterials*, vol 27, pp. 377-387, 2006

[6] C. J. Siepe, H. M. Mayer, and K. Wiechert "Clinical results of total lumbar disc replacement with Pro Dic II: three-year results for different indications"*J Spine* , vol 31pp.1923–32, 2006

[7] V. K. Goel, J. N. Grauer, T. Ch. Patel, A. Biyani, K. Sairyo, and S. Vishnubhotla, *et al*, "Effects of Charité artificial disc on the implanted and adjacent spinal segments mechanics using a hybrid testing protocol," *J Spine* vol 30, pp. 2755–69, 2005

[8] J. N. Grauer, A. Biyani, A. Faizan, A. Kiapour, K. Sairyo, and A. Ivanov, *et al*.,"Biomechanics of two-level Charite artificial disc placement in comparison to fusion plus single-level disc placement combination," *Spine J*, vol 6(6), pp. 659–66, 2006

[9] G. Denoziere and D. N. Ku, "Biomechanical comparison between fusion of two vertebrae and implantation of an artificial intervertebral disc," *J Biomech* , vol 39(4), pp.766–75, 2006

[10] A. Rohlmann, T. Zander, and Bergmann, "Effect of total disc replacement with ProDisc on intersegmental rotation of the lumbar spine," *Spine J* , vol 30(7), pp.738–43, 2005

[11] C. Shih-Hao, Z. Zheng-Cheng, C.Chen-Sheng, C.Wen-Jer, and H. Chinghua. "Biomechanical Comparison between Disc Arthroplasty and Fusion" *J. Med Eng Phys*. Vol 31, pp. 244-253, 2001

[12] A. Ghouchani and H. Katouzian, "Optimized model of lumbar artificial disc using finite element method," in *Proc. 2011, 3rd international Conf. on software technology and engineering* (ICSTE 2011), Kuala Lumpur, Malaysia, ASME press, pp. 549-554.

[13] W. Sun, B. Strarly, J. Nam, and A. Darling. "Bio-CAD modeling and its applications in computer-aided tissue engineering". *J. Computer-Aided Design*. Vol 37, pp. 1097-1114, 2005

[14] A. Agur and M. J. Lee, Grant's atlas of anatomy. 10th ed. Philadelphia: Lippincott Williams & Wilkins; 1999.

[15] M. A. Tyndyk, V. Barron, P. E. Mchugh, and D. O. Mahoney. "Generation of a finite element model of the thoracolumbar spine". *J. Acta of Bioengineering and Biomechanics,* Vol. 9, No. 1, pp. 35-46, 2007

[16] N. Campbell-K, M. Jorgensen, D. Burr, and W. Marras, "The Prediction of Lumbar Spine Geometry; Method Development and Validation". *J.Clinical Biomechanics*. Vol 20, pp. 455-464, 2005

[17] D. Coombs, M. Rao, M. Bushelow, J. Deacy, P. Laz, and P. Rullkoetter. "Simulation of lumbar spine biomechanics using Abaqus". *2011 SIMULIA Customer Conf.*

[18] A. Shirazi-Adl and A. Ahmed. "Mechanical Response of A Lumbar Motion Segment in Axial Torque Alone and Combined With Compression" *J. Spine*. Vol. 11. ,pp. 914-927, 1986

[19] P. Pollintine, A. S. Przybyla, P. Dolan, and M. A. Adams, "Neural arch load-bearingin old and degenerated spines," *J Biomech*, vol 37(2), pp. 197–204, 2004

[20] S. A. Shirazi-Adl, S. C. Shrivastava, and A. M. Ahmed, 'Stress analysis of the lumbar disc-body unit in compression. A three-dimensional nonlinear finite element study*," J Spine* , vol 9(2), pp.120–34, 1984

[21] A. Ghouchani, M. Ravari, and F. Mahmoudi, "Modeling and optimization af an elastic arthroplastic disc for degenerative disc disease," presented at International Conference on Modeling and Optimization, Cairo, Egypt, October 1-3, 2011

[22] A. Ghouchani, "Geometry and material optimization of intervertebral arthroplastic disc," MSc. dissertation, Dept. Biomedical. Eng., AmirKabir University of Technology, Tehran, Iran, 2010.

[23] H. Li and Z. Wang, "intervertebral disc biomechanical analysis using finite element modeling based on medical images," *J Computerized medical imaging and graphics*, vol 30, pp. 363-370, 2006