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Development of Computer Aided 3D Model From Computed Tomography Images and its Finite Element Analysis for Lumbar Interbody Fusion with Instrumentation

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Abstract – The purpose of this study is to clarify the mechanical behavior of human lumbar vertebrae (L3/L4) with and without fusion bone under physiological axial compression. The author has developed the program code to build the patient specific three-dimensional geometric model from the computed tomography (CT) images. The developed three-dimensional model provides the necessary information to the physicians and surgeons to visually interact with the model and if needed, plan the way of surgery in advance. The processed data of the model is versatile and compatible with the commercial computer aided design (CAD), finite element analysis (FEA) software and rapid prototyping technology. The actual physical model is manufactured using rapid prototyping technique to confirm the executable competence of the processed data from the developed program code. The patient specific model of L3/L4 vertebrae is analyzed under compressive loading condition by the FEA approach. By varying the spacer position and fusion bone with and without pedicle instrumentation, simulations were carried out to find the increasing axial stiffness so as to ensure the success of fusion technique. The finding was helpful in positioning the fusion bone graft and to predict the mechanical stress and deformation of body organ indicating the critical section.

Keywords: **Intervertebral disc, Threshold, Segmentation, Stiffness**

1. Introduction

Inter-body fusion is a major surgical treatment for inter-vertebral disc degeneration and instability. A degenerative change of the lumbar motion segment often leads to the decrease in disc height that accompanies motion segment [6]. Artificial disc prostheses have been used as one of the treatment to protect the vertebrae from the excessive changes in mechanical stress. To minimize contact stress, researchers tend to design the device to cover the entire cross-sectional area of the vertebrae so that the load can be spread over a large surface area. Although this makes mechanical sense, it renders surgical insertion more difficult and requires almost every artificial disc to be implanted via an open anterior approach with a large incision, and also entails prolonged operative time. The method preserves the mobility between the two vertebrae but due to the complexity of the structure and functions of the disc, it has proven difficult to design an artificial total disc that mimics all the mechanical properties of a natural disc while retaining the required durability [1].

Inter-vertebral fusion is a surgical technique to restore the height of the degenerated disc space by inserting a spacer and bone graft or artificial fusion element between vertebrae. The advantages of using inter-vertebral fusion include the mechanical stability and maintained the original disc height. However, clinical observations have shown that, using fusion method confines the mobility between the motion segment and the restored disc height immediately after surgery tends to return even below the preoperative level [3,11]. To avoid this effect, pedicle fixation is used. But the early failure of instrumentation fixation could result into bone graft collapse, recurrence of spinal deformity and accelerating deterioration of the neighboring disc. Hence, for the successful fusion process, the parameters need to be thoroughly analyzed are, bone mineral density (BMD), contact areas between devices / grafts and its location, the extent to which the endplates are removed, material properties and position of instrumentation and spacer, and loading conditions [23].

There are many published reports describing the cage technique and effectiveness. However, little research information is available for the biomechanical behavior of inter-body fusion coupled with inter-body spacer for the patient specific model. Most of the researchers capture the anatomical data from the cadaver specimen by using the digitizing technique or else, the information was used from the previous literature to generate the FE model [10]. Such a simplified model reduces the computational cost and time, but the results so obtained are erroneous and do not match with the actual and may contribute to failure of the fusion technique. The objective of the present study is to investigate the biomechanical

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behavior of patient specific model for the inter-body fusion coupled with inter-body spacer, fusion bone and instrumentation using finite element method.

2. Materials and Methods

2.1 Generation of CAD model from CT scan data

The CT scan data have been most successfully used to generate the geometry of the bony anatomy model. The author has developed a programming code to converts the complex 2-D CT scan data to the volumetric data and stores it in .stl (stereo lithography file) form. Volume segmentation for the 3D volumetric model which is developed from consecutive images stacked together is an important part of computer based medical applications for diagnosis and analysis of anatomical data. In this technique a single value called threshold is used to create a binary partition of voxel (volume element) intensities. All voxels with intensities greater than the threshold are grouped together into one class and that with intensities below the threshold is grouped together into another class. This technique is very effective in getting segmentation done in volumes with a very good contrast between regions. The main drawback of this technique is that, the results are to tightly couple with the thresholds used. Any change in the threshold values can give a different segmented region [7].

First, the CT scan images of the required part are taken on different frame planes. Such planes are stacked together to provide a volumetric image of the body organ. The developed computer code converts the complex data stored in the data element to the pictorial form by using the isosurface function. Isosurfaces is defined by connecting voxels with intensities equal to the isovalue in a 3D volume [17]. The main graphical user interface and program is developed in MATLAB version 7.3. The different functions like dicomread, smooth3, isosurface, isonormals, trimesh and trisurf, from the MATLAB toolbox, were used to develop the computer program. The flowchart is shown in Fig. 1.

The salient features of the developed software are- it shows the 3D anatomical geometric model, distinguishes soft tissue and bone tissue clearly. It also rotates, animates views of CT scan image and detects the boundary and planes of CT images from all the sides. The utility tool to measure the distances between the pixels is available which helps surgeons and physicians for the preliminary diagnoses. It may help the surgeons to decide exact shape and size of the implants. The required slices of CT images can be processed at different threshold value for 3-D visualization and data conversion. The aspect ratio of the obtained figure can be modified as per the distance of incrimination of CT slices are taken for visualization. Further the capability can be enhanced for FE analysis as the mesh geometric model can be prepared easily. The developed software has an edge over the other available commercial CT data conversion software such as RapidForm® (INUS Technology Inc., Korea) and Mimics ® (Materialise Inc.). The developed software is very economical, easy to operate and the work environment required is only Mat-lab. Most of the CT data conversion software produces scalar data in the 2D form having the surface geometric features. The developed surface model hence required more preprocessing work when it is analyzed with commercial FEA software.

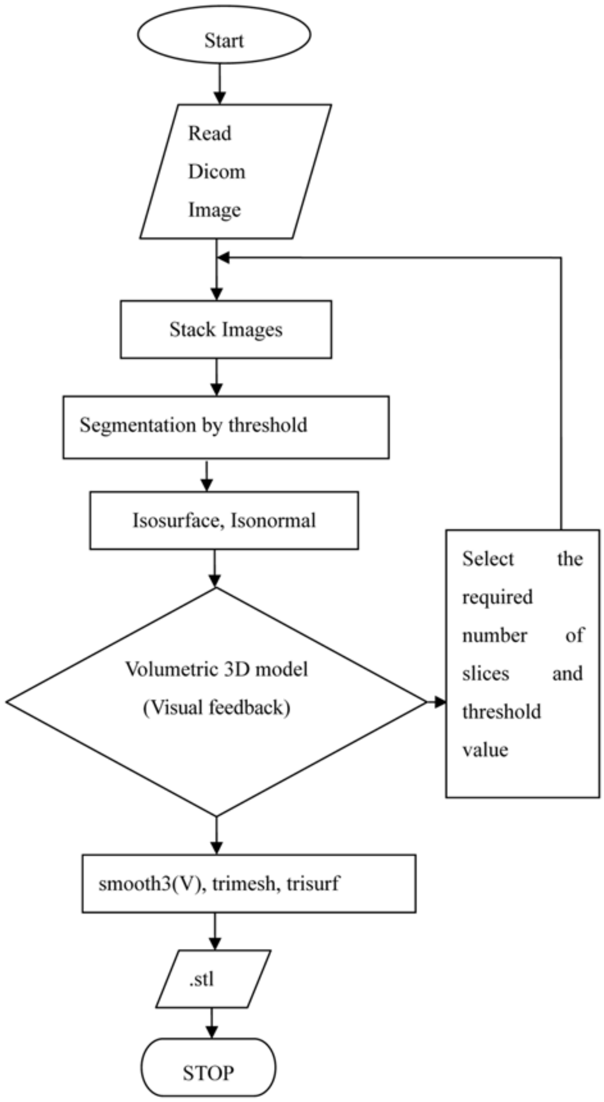

Fig. 1. Flowchart indicating steps used for developing the computer program which is used for visualization of 3D model and converting data into. stl file format.

To verify the capability of the developed software the CT images of lumbar vertebrae is processed to generate a 3D mode and also used in FE analysis. First, all the 236 axial slices of CT images of a 39-year-old female patient which is stored in Digital Image Communication in Medicine (DICOM) files are loaded into the developed software in sequential order. Out of 236 CT data, images from 150 to 236 are processed at 1250 threshold value to generate the 3D model of lumbar vertebrae shown in Fig. 2.

The physical model is manufactured for lumbar vertebrae using rapid-prototyping technique to verify the executable capability of .stl file generated by the developed program code shown in Fig. 3. The physicians or surgeons can make a rehearsal or conduct a mock surgery as a preparation with obtained model and if needed, decided the suitable implant. The .stl file format data can also be used in computer aided manufacturing (CAM) technique.

The developed software was very sensitive to noise and intensity in-homogeneities. Thus it cannot be easily applied to magnetic resonance imaging (MRI) and ultrasound volumes.

The CT scan data conversion procedure for developing the FE and RP models used in this study is shown in Fig. 4.

Multi-segmentation and volumetric meshing features have not incorporated in the developed software, the generated .stl data of lumbar vertebrae is imported to the 3-D data processing software. Geo-Magic Studio9.0® software is used for the minor repair of the model, such as removal of the artifacts and unwanted part and performs smoothing operation. The geometric data is converted into a universal graphics data format, initial graphics exchange specification (.iges). The entire finished model of required L3-L4 vertebrae is imported to UGS NX 5.0.2.2 ® for developing computer aided model. The vertebral body is assumed to have a cancellous core with the periphery composed of cortical bone. The geometrical non-linearity in the shape of the cortical bone and model cancellous bone was maintained by performing Boolean operation [18]. Even though the thickness of cortical bone varies depending on each vertebral body and on the location (anterior and posterior), an average value of 1 mm was used for the modeling in this study [20]. Many times soft tissue of inter-vertebral discs and endplates could not be captured by CT scan images correctly which make it very difficult to distinguish it appropriately and model [14]. Considering the drawback of the present technique, a modeling process accompanied by a manual technique has been used to redefine such complex geometry and soft tissue to a certain extent and to connect them to the bony anatomy. The inter-vertebral disc consists of an annulus and nucleus pulposus bounded by the endplates. The volumes for the discs and the endplates are generated by extruding the surface patches on the bottom and top surface of vertebrae. Endplates is 0.5 mm thick assumed at the superior and inferior surfaces of the inter-vertebral disc [5, 19]. The nucleus pulposus, modeled as an incompressible material, with 43% of the total disc volume and located slightly posterior to the center of the disc consistent with anatomic measurements [25]. The dimensional features

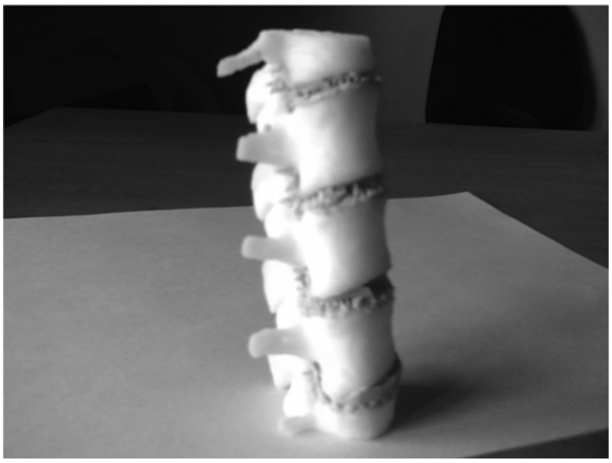

Fig. 3. Scaled physical rapid-prototyping model of lumbar vertebrae.

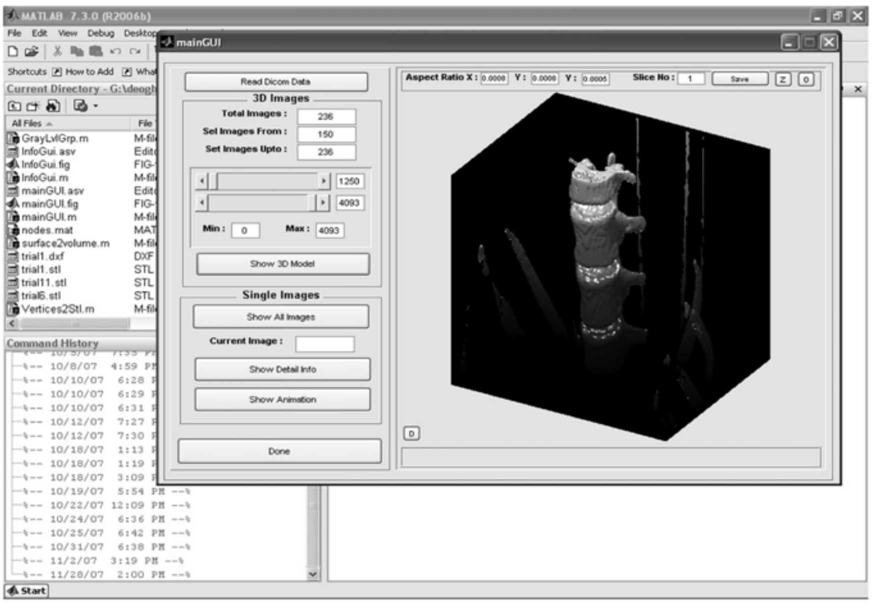

Fig. 2. Three Dimensional Lumbar model developed from CT scan data.

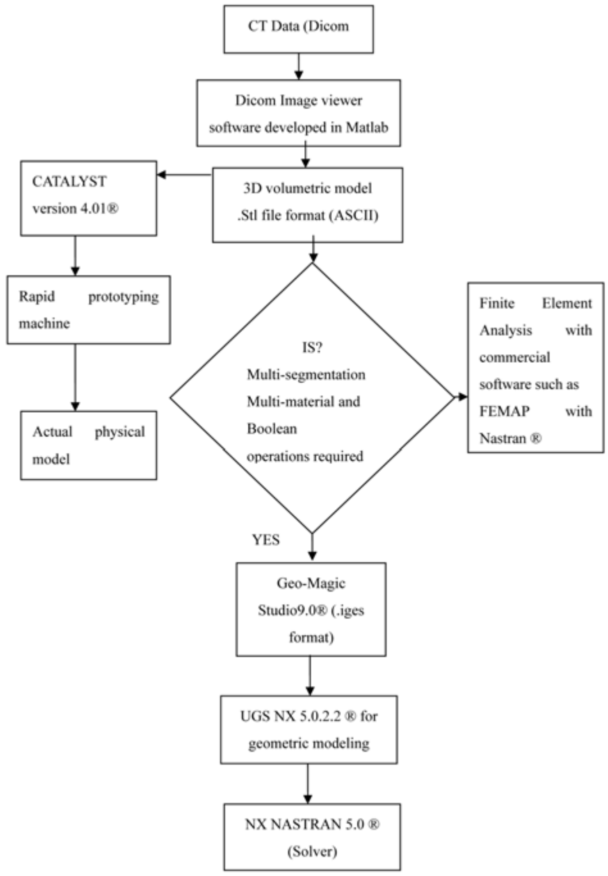

Fig. 4. A data conversion procedure for developing FE and rapid prototyping (RP) medical model.

obtained in geometric modeling software is confirmed by measuring the actual dimensions of similar age cadaver specimen to validate the modeling accuracy with vernier caliper shown in Fig. 5.

3. Results

3.1 Finite element model and its analysis

FE modeling and analysis is an important method of assessing complex motions and forces where measurement and experiments are impractical. The three-dimensional FE mesh model is generated using higher-order 10 noded tetrahedral solid elements. The model has 110321 nodes and 69616 total numbers of elements. The glue coincident mesh mating condition is used for finite element analysis with constrained to zero displacement at the bottom of L4 vertebrae. Since the average mass of adult humans fall within the range of 40 kg to 100 kg, corresponding compressive static equitant load taking into account the body weight, including trunk, head and arms is 400 N and 1000N. The biomechanical behavior at compressive loading conditions 450 N and 720 N are simulated with the assumption that the total load considered was distributed as 30% of total load on facet, 30% of total load shared by cancellous bone and remaining 40% of total load on cortical bone [2, 13]. The model

Fig. 5. Dimensions measurement with vernier caliper.

with boundary condition is then imported into NX NASTRAN 5.0®, FE analysis software. Material properties for different vertebral components like cortical bone, cancellous bone, endplates, annulus matrix and nucleus are taken from the literature [6, 4, 20, 25]. The FE results obtained are shown in Fig. 6. (a) and (b).

The results for different components are verified by comparing with previously published experimental and numerical results. Comparison between maximum stresses predicted in the healthy configuration and the corresponding expected failure stresses found in the literature is shown in Table 1. The results for displacement and Von-Mises stresses obtained in the different components show close agreement with literature results. Table 1. Summarizes, the comparison of simulated result and the corresponding literature results.

This validation of results ensures the correctness of the geometric model, reliability of FE model and its analysis. This model was utilized in the subsequent analysis for other simulation.

3.2 FE analysis with the inter-body spacer

The fusion bone and disc replacement are two general methods adopted to overcome the disc degeneration problem. The success rate of the fusion method can be improved if the information for the specific patient implant model such as of maximum permissible displacement, maximum allowable stress and stress distribution pattern under the different loading conditions is available with surgeon before the surgery. Similarly simulation results of proper positioning of bone graft material between the vertebrae and adequate stiffness requirement if available in advance before the surgery with the surgeons the probable bone graft collapse can be avoided by implementing proper safety measure [6].

The FE model developed is used for analyzing the effect of disc degeneration. The superior endplate, annulus and nucleus are removed and the space is packed by the Mic-Space inter-body spacer [19] and fusion bone graft materials. The simplified model of spacer considered in the analysis of titanium material (Ti-6Al-4 V). Three different configurations with and without fusion bone

Table.1

Components	Current FE model and it's analysis results	Results obtained from earlier research paper
Inferior Endplate deflection	Endplate deflection under compressive load : 0.02197 mm under 450 N and 0.0355 mm under 720 N	Range of End Plate deflection under Compressive Load from 0.03 to 0.1 mm under approx. 720 N [15,16]
Cortical bone	Maximum Von Mises stress in the cortical bone is 98 Mpa under 450 N and 150.7 Mpa under 720 N	Average Compressive Strength of Human cortical bone167~215 MPa [12].
Cancellous bone	Maximum Von Mises stress in the Cancellous bone is 2.35 Mpa under 450N and 3.75 Mpa under 720 N	Average Compressive Yield Stress 8.2 Mpa [8].
Annulus	Maximum Von Mises stress in the Annulus is 1.50 Mpa under 450 N and 2.54 Mpa under 720 N	Expected Failure stress 46 Mpa [22].

are analyzed for three different location of spacer on inferior endplate as a part of inter-vertebral disc. The fusion bone is assumed having a material property of 12GPa elastic modulus and 0.3 as Poisson's ratio, which is the same as that of cortical bone in this study. The position of spacer is located 5 mm, 10 mm and 15 mm from the dorsal side of the spacer and rear edge of the vertebral body. Accordingly, the volume of the fusion bone calculated are 4,990 mm^3, 3,360 mm^3 and 1,652 mm^3 and the volume of spacer was maintained constant as 1,512 mm^3. A total of six FE implant models are created for parametric studies by varying the spacer position with and without fusion bone to simulate feasible configuration. The contact between the spacer and fusion bone and the endplate was assumed to be bonded mechanically to simulate the high friction coefficient of plasmapore coating and teeth on the spacer. To characterize the importance of compressive loading and main function of spacer position, a physiological compressive displacement of 1 mm is applied in five equal increments on top surface of L3 vertebral body with fully constraint conditions on the bottom surface of L4 vertebral body. For the implant models with spacer alone, and spacer with fusion bone, the axial stiffness is determined by extracting reaction force at the inferior surface of L4 in each increment with corresponding compressive displacement applied onto the implant model [10]. The simulation results and the effect of positing spacer in the inter-vertebral disc with and without fusion bone material on the axial

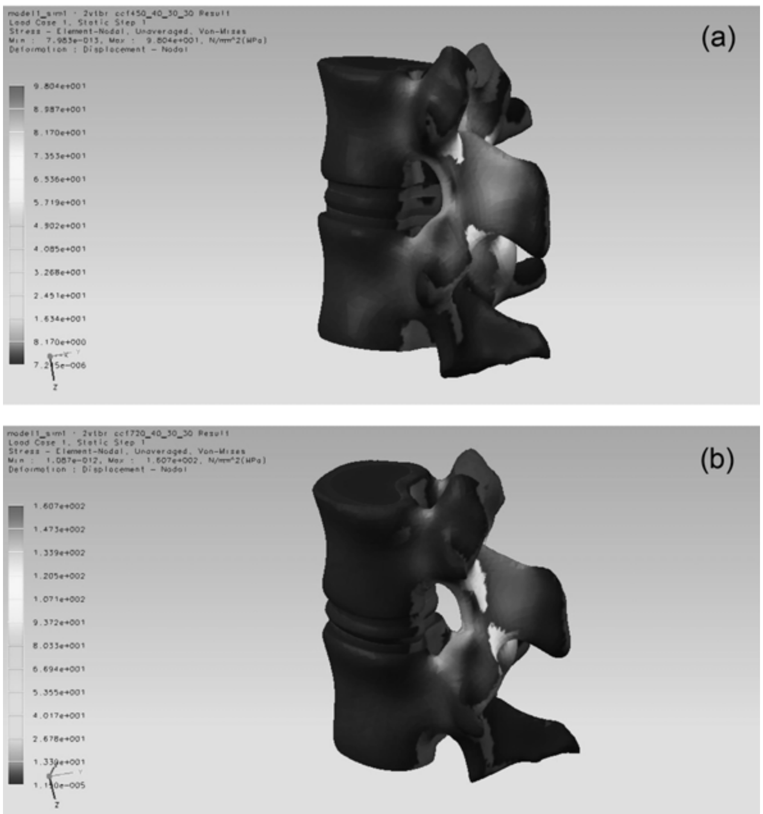

Fig. 6. (a) Von-Mises stresses at 450 N, (b) Von-Mises stresses at 720 N.

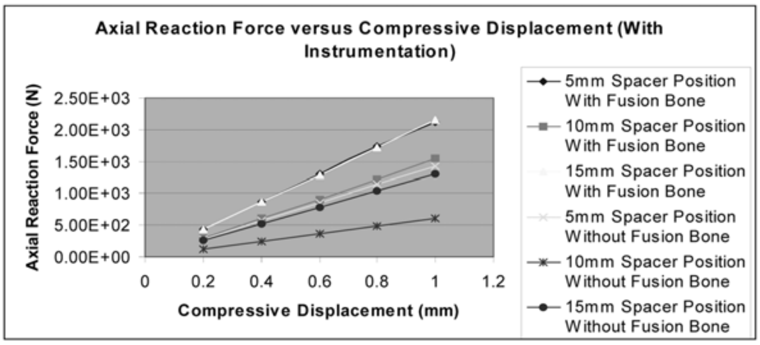

Fig. 7. (a) Axial reaction force versus compressive displacement without pedicle instrumentation for inter-body spacer with and without fusion bone.

reaction force under 0.2 mm incremental axial displacement is shown in Fig. 7 (a).

The positioning of the spacer alone at 5, 10, and 15 mm from the rear edge of the vertebral body did not show significant variation in the axial reaction force. However, the implant model with spacer and fusion bone predicted radical change in axial reaction force. The results of these simulations are very significant in order to locate the appropriate position of fusion bone graft and spacer.

The pedicel fixation increases compressive stiffness which is helpful to reduce prolonged healing time of fusion bone. Hence, percentage increase in the axial compressive stiffness of implant FE model due to different positions of spacer, size of fusion bone with pedicle instrumentation is carried out. The simplified model of pedicle instrumentation is considered in the present analysis [21]. Again six FE models are analyzed by changing the position of spacer and fusion bone with pedicle instrumentation. The pedicle instrumentation was considered as of Titanium (Ti-6Al-4 V). The area configuration of the pedicle instrumentation is 1 mm × 4 mm consider. To avoid the incidence of early screw failure, the insertion of pedicle screw through pedicle to the cancellous bone is maintained as 85% of the total extreme length between pedicle and cortical bone in all six simulations [24]. Fig. 7(b). indicate integral FE mesh model with spacer, fusion bone, simplified pedicle instrumentation and boundary conditions.

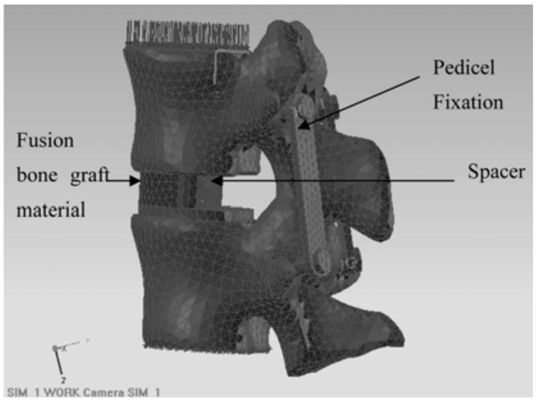

Fig. 7. (b) Integral FE mesh model with spacer, fusion bone, simplified pedicle instrumentation and boundary conditions.

The lumbar interbody fusion with instrumentation helps to prevent fusion bone graft disarticulation, maintain the preferred height of the disc and provide the stability to promote bony union between adjacent vertebral bodies. The use of instrumentation under axial compressive force increases stiffness which reduces prolonged healing time of fusion bone. However, over stiffness results into fusion bone collapse. Hence it is essential to analyze the increase in axial stiffness with instrumentation in presence of spacer, fusion bone graft combination in order to avoid over stiffness. Simulations are carried out in the presence of pedicle instrumentation, for a compressive displacement of 1 mm applied in five equal

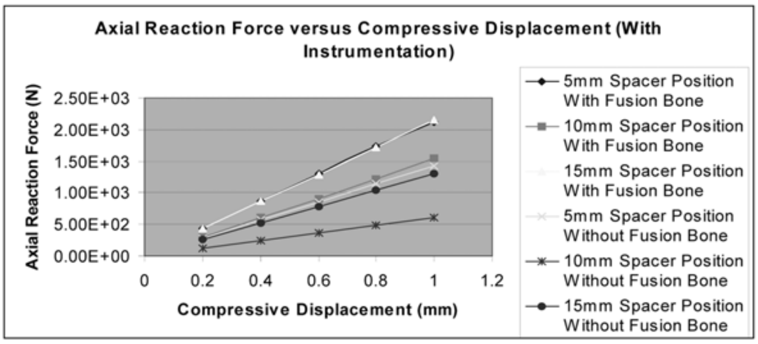

Fig. 7. (c) Axial reaction force versus compressive displacement with pedicle instrumentation for inter-body spacer with and without fusion bone.

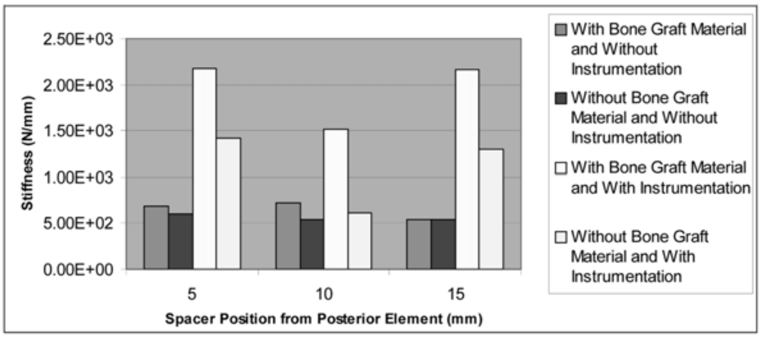

Fig .8. Bar chart representation of stiffness value due to different positions of spacer with and without fusion bone in presence of with and without pedicle instrumentation.

increments of 0.2 mm. The simulation results for different input displacements vs axial reaction force in presence of pedicle instrumentation used is shown in Fig. 7 (c).

The bar chart shown in Fig. 8. clearly indicates that use of pedicle instrumentation increases the stiffness of the integral model with bone graft and the spacer. It is observed that for 1 mm input displacement, presence of pedicel instrumentation increases stiffness by 217.78%, 109.72% and 296.33% for position 5 mm, 10 mm and 15 mm. Similar chart could be plotted for the other input displacements to find the increase in stiffness. The information is vital for the surgeons during the surgery and for clinical study to select optimum stiffness value for better and fast fusion.

4. Discussion

In the present research, a program code was developed to generate patient specific 3D model of human organ clearly defining all the vital anatomic features from CT data. The software code not only reconstructs the CT data but also makes the base for the CAD model or rapid prototyping model. Most significantly, the FE model based on the CAD data, once developed, can be easily modified and extended to add new implants. Considering the drawback of artificial disc replacement treatment, the work is focused to predict the effects of fusion bone along with spacer position with and without pedicle instrumentation. Although the current FE model is build to closely to match anatomic geometry and material properties, there are some limitations. Mainly the components of vertebral body was never so uniform in actual with isotropic properties. The current study considers only the compression loading without ligaments, but it might be interesting to study the implant model under other loadings to simulate the biomechanical behavior of lumbar model coupled with implants and fusion bone. However, the FE simulations of the patient specific model from CT scan data can provide reasonable predictions of displacements and stresses in lumbar vertebrae. The surgeons can locate the appropriate position of spacer implant and instrumentation fixation so that the harvesting of the fusion bone is proper and faster. Also, over stiffness effect can be predicated in advance to avoid fusion bone collapse. The optimum stiffness can be determined and recommended to reduce the prolonged healing time of the graft bone. The results show that varying the spacer position in surgical situations does not significantly affect mechanical behavior in terms of axial stiffness for the implant model, whereas use of pedicle instrumentation drastically increases the axial stiffness. The stress distribution results obtained during the simulations is helpful to know the critical section of the model, so as to suggest proper prevention and precaution to the patient. The work can be extended to design the proper instrumentation so as to set it for optimum stiffness value under proper stress distribution condition to ensure success of the fusion bone treatment.

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