Vibration Injury Network

Research Network on Detection and Prevention of Injuries due to Occupational Vibration Exposures

Development and Analysis of a Finite Element Model of the L5/S1 Disk using a Loading Model Simulating Whole-body Vibration

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ABSTRACT

Objectives. This study, being part of a broader program that will cover the entire lumbar region of the human spine, aims at developing a fine partition, three-dimensional, finite element model of the L5/S1 intervertebral disk suitable for step-by-step calculation and video recording of the impacts of vibratory and shock loading that simulate conditions of whole body vibration, time-dependent multidirectional vibrations or continuous (sequential) shocks and permitting a. Identification of the sites of maximum displacement from the corresponding positions of equilibrium in association with the loading size and direction sequence of shocks, b. Graphic and Video Simulation (GVS) of disk motion, and c. Prediction of possible damage due to the afore mentioned conditions.

Design. The disk model has 7240 elements and 6900 nodes. Three load cases were studied: a. Constant compressive load of 1600 KPa. b. Linear variation between 200 and 1600 KPa in x-direction c. Linear variation between 800 and 1600 KPa in y-direction. In all three cases the pressure was applied on all the external elements of the model having $z >= +0.4$ cm.

Methods. The model of vibratory and shock loading was selected on the basis of known peak pressure values, and the stress analysis of the disk model was performed for two sequencial peaks, assuming that

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the disk restitution period (natural period) is larger than the time interval between the two successive loadings.

Results. The tentative results show that in both cases the impact of successive loading is additive as far as deformations are concerned and that final damage, by creation of hernia or rapture, occur by lower magnitude but successive loading rather than by higher magnitude isolated shocks without repetition. The second series of experiments tentatively shows that permanent disk damage can occur at even lower magnitude successive loading, particularly at the presence of inhomogeneities of the modulus of elasticity in the disk's annulus fibrosus.

I. INTRODUCTION

The early contributions to the knowledge of the material and dynamical properties and behaviour of the lumbar intervertebral disks, and particularly the L5/S1, were followed by further investigations and contributions using finer model definition technologies and more precise dynamical analysis methods. The former contributions of high interest refer to the dynamical properties and analysis of the intervertebral disk either alone or as part of the spine. The concern for this essential part of the spine for the well being and successful reception by the human body of occasional or repetitive excessive loading and/or of impulsive forces, led to increased concern for the problem resulting in the latter series of more recent contributions. Nor less encouraged this line of research aims on proving the generally accepted (see epidemiological studies[1,2,3,4,5,6,7]) correlation between excessive loading and vibration of human body, low back pain and pertinent disorders. Between the former contributions worth citing are the work of Gallante[8], investigating the tensile properties of the lumbar annulus fibrosus, of Heliovaara[9], referring to the occupational risk leading to lumbar disk herniation, of Hirsch[10], covering the capability of the disk to react to compression forces, of Kelsey et al[11], investigating the risk factor of motor vehicle drivers to develop disk herniation, of Koeller et al[12] , on the biomechanical properties of the disk subjected to axial compression and the effect of age and degeneration upon these properties, of Natali et al[13], referring to the non-linear dynamical analysis of the loaded intervertebral disk and of others. Due to the geometry and material constitution of the intervertebral disk, most of the research conducted, older and recent, had to be based upon the results of fundamental research on the behaviour of coupled fluid-solid (in our case fluid-fibrous) systems subjected to fixed or time-dependent loading. Much of this basic research is mentioned by Clough et al[14], but broader and detailed methodological background information has to be seeked in the recent literature of finite element analysis, e.g. in Burnett[15], Gradin[16], Kardestuncer et al[17], Mori[18], Segerlind[19], Yang[20], and Zienkiewicz et al[21].

More recent contributions to the behaviour and analysis of the impact upon the intervertebral disk under fixed and time dependent loading as well as impulse forces, once only or repetitive, and applying finite element analysis, a basic tool employed in this contribution, include the work of Laible et al[22,33], Simon et al[23,24], Shirazi-Adl [25,27,28], Shirazi-Adl et al[26], Natarajaian et al[29], Broman et al[30], Goel et al[31], and McNally et al[32]. As will be pointed later in this paper, there is fundamental difference between the impacts upon any spinal disk of a static load, or even a single impulsive force and a time-dependent, quasi-periodic load or repetitive impulse force. In the latter case resonance phenomena, absent in static loading, requiring knowledge and treatment of properties of the load or shock and of the loaded disk, such as the frequency of the load applied and the natural frequency of the disk have to be taken into account. This is so, since it is broadly accepted that failure of even the best designed structures (e.g. suspension bridges) occur when resonance conditions are present (e.g. Tacoma Bridge case). It is only a guess for the moment that a non-negligible percentage of intervertebral disk failures fall into this category. The contributions of Kitazaki et al[34,35], Matsumoto et al[36], Seidel et al[38], and Buck[37], provide essential information and results in this important aspect of the problem.

The studies mentioned and the general literature of the physiology and pathology of the lumbar vertebrae and disk lead to the following observations:

1. The geometrical parameters defining the shapes of the integral parts of the lumbar spine zone (vertebrae and disks) vary broadly among people.
2. The material properties of the same parts, such as modulus of elasticity of the bones and the fibrous constituents, viscoelastic coefficient of the nucleus pulposus, porosity of the fibrous part, also vary between subjects.
3. The loading of the entire lumbar system through whole body vibrations and/or repetitive shocks, as well as the loading of the constituent parts of the same system, is of general direction and not necessarily axial, while also general is the frequency of these time-dependent loads.
4. The failures of the lumbar zone, e.g. rapture of the fibrous part of a disk, or hernia, have often small linear dimensions in comparison to the size of the damaged part (shedding of the content of nucleus pulposus is realized through millimeter size slits).

While the broader objective of this line of research includes treatment of the entire lumbar and sacral zone, both as total and as constituent parts, by developing, loading, and analyzing refined three-dimensional finite element models, this paper refers exclusively to the treatment of the L5/S1 intervertebral disk. This strategy is described as the separate treatment of each integral constituent part of the spine considering it as a free body upon which the applied external load includes, besides the direct forces, the forces resulting from its direct physical connection to the rest parts of the spine.

As first example of application of this strategy we take the intervertebral disk and in particular the case of the L4/L5 disk. The finite element model to be developed is described by anthropometric and dynamical parameters that allow treatment of all disks of the human spine. The general example that follows and the sequence of parameter definition presented illustrates the method.

II. METHODS

The methods employed in this paper for the dynamical analysis of the lumbar section of the spine and of the intervertebral disk L5/S1, includes application of D'Alembert's principle, of the free-body principle and of the finite element principle. The combination of these three principles permits isolation of any part (in this case the intervertebral disk) of a composite body (here, of the spine) and study of the dynamical behaviour of this part (free-body principle) by transforming the problem of the moving (in this case, vibrating) part into a problem of statics (D'Alembert's principle) and, finally assuming that partitioning the part under study into finite elements and distribute the time-dependent load between the nodes of defining these elements we shall achieve presentation of the deformations of the part through the displacements of the nodes (finite element principle).

Since the first two principles are employed for the definition of the total load force (weight, plus contact force, plus inertial force caused by vibration or repetitive shocks) to be applied upon the L5/S1 disk, we proceed to discussing and presenting the finite element model suitable for studying the dynamical behaviour of this part.

II.1. The Finite Element (FE) Model of the L5/S1 Disk.

The body to be analyzed by the FE method has the shape of a hollow bean. The horizontal section of the object is given in Figure 1. The hollow part is filled with a jelly-like liquid of material properties to be discussed later. The basic average human physical dimensions of the hollow bean-shaped disk are given in this Figure as well as in Figures 2 and 3. In Figure 3 we give the vertical middle section of the disk and of its nucleus with their average physical size.

The Cartesian co-ordinate system adopted for presenting the internal and external surfaces of the disk has its origin at the geometrical center of the lower external surface, the z-axis normal to the upper surface of the S1, the Ox axis is perpendicular to the Oz axis and pointing to the human anterior, and the Oy axis is perpendicular to the other two axes and pointing to the right. The numerical tables defining the two surfaces defining the fibrous part and nucleus pulposus by a dense set of points, were computed using a standard autocad computer software upon sections of the disk normal to the Oz axis. These Tables are not reproduced here.

To facilitate both presentation and use of the external and internal surface co-ordinates of the fibrous part of the disk, points are selected along sections parallel to the Oxy plane and at suitable distances from it. The origin of the co-ordinate system is placed at the geometrical center of nucleus pulposus and the Ox axis is positive along the forward direction. The Oy axis is positive to the right and the Oz axis positive upward. The co-ordinate system is solidly attached to the human body and hence changes position and orientation with the posture. The distinction between the various postures is taken care by the direction cosines of the vector of gravity with respect to the co-ordinate system. The selected co-ordinate system is rotating CCW and not CW but this is unavoidable for compatibility reasons with already made assumptions in previous work.

II.2. Physical Properties of the Disk and the Nucleus Pulposus.

The disk modulus of elasticity of the fibrous part was given the standard value 3000 N/cm^2 (a value used in many similar studies). However, the model was tested not only for various alternative values, smaller and larger, but also for locally varying values during the same run. As already mentioned, the usually assumed homogeneity throughout the fibrous part of the same intervertebral disk has been seriously questioned since its validity is highly improbable. In fact, some properties of the disk, such as symmetry of the fibrous part, are known to be unrealistic, although for any finite analysis software it is inessential, as it is unrelated to its basic performance when loaded. To the contrary, other properties, e.g. uniformity of the modulus of elasticity and uniformity in the thickness of the fibrous part are very important. If any of the two properties is position-dependent, then at places where minimum values occur, particularly, but not exclusively, at the unbounded equatorial regions of the disk can appear hernia or rupture.

For the nucleus pulposus model a 'bulk modulus' is required by ANSYS code. For fluids of viscosity in the range of water this value is suggested by ANSYS users' manual to 210000 N/cm^2. This value was used in the present analysis. A sensitivity study on this value (half and double of the 210000 N/cm^2) showed no considerable variation in the displacements and stress results of the disk. Therefore this value is considered as unimportant.

II.3. The Finite Element Model

The commercial Finite Element code ANSYS is used for the numerical analysis. Two different volumes are created, using the ANSYS pre-processing module. One volume represents the disk and one the nucleus. The model of the fibrous part comprises 6088 3D-solid elements with 3 degrees of freedom per node, the three displacements u_x, u_y and u_z. The model of the nucleus is represented by 1152 fluid elements. The whole model has 7240 elements and 6900 nodes.

II.4. Loading of the Disk and Nucleus Pulposus – Simulation of Vibrating Loading and Repetitive Shocks.

The loading of the disk, as well as all other parts of the human spine under free body study is a very delicate and complex problem considering that the final objective is to simulate the load to that of WBV conditions. The model of loading has to be broad so as to cover all dynamical conditions possible to occur, including size and direction of accelerations and postures of the human spine. Since the loading is continuously varying, while the FE analysis requires freezing of the time and selecting suitable times, and hence loading, for which the strains will be calculated, the problem of selection of the suitable loading becomes critical. This problem will be treated later. For the present the simplest possible loading model is selected in order to test the FE model already presented.

The load model selected to be applied upon the surface points of the disk, is denoted by the corresponding local weight and inertial force pressure **p**, that originates from the body weight $\mathbf{B}_d$ resting above the disk under study. This body weight has a vector of vertical direction, while the direction of **p** is that of the normal to the disk surface. The vector **p** is rigidly connected to the frame of reference Oxyz. The angle ö between the vectors **k** (unit vector along Oz-axis) and $\mathbf{B}_d$ depends upon the body posture.

In order to perform the initial tests of the FE model and its performance three types of loading were treated:

$$p = [800+(x-a)400]\text{KPa}. \qquad (1)$$

a. Constant pressure of 1600 KPa along the Oz-axis.

b. Linear variation between 800 and 1600 KPa in x-direction using the formula

   where a is the algebraic value of x at the far left end of the fibrous disk.

$$p = [800+(y-a)400]\text{KPa}. \qquad (2)$$

c. Linear variation between 800 and 1600 KPa in y-direction using the formula

   where a is the algebraic value of y at the far back end of the fibrous disk.

In all three cases the pressure was applied on all the external elements of the model having z > +0.4 cm.

In a study of the dynamical behaviour of the intervertebral disk under conditions of time-dependent loading, such as the one applied under conditions of whole body vibration, the FE tool can be applied through repetitive runs of the ANSYS software. In each run the disk will be assumed to be as deformed after the previous run while the loading will receive its value that corresponds to the next time-step. This task requires performance of the analysis for a rather large number of times and of course cannot provide real-time output, particularly for model of many elements such as the one presented in this paper. Video reproduction, after the programme runs, of the stepwise deforming disk is possible. The same can be achieved by applying a sequence of loads, say from -1600KPa to 1600KPa with a step of 200KPa, to record graphically (video) and numerically the deformations of the disk. Again, the assumption will be that each new run starts from the deformed geometry of the previous run. The effect of fluid loss during compression[39], although for some cases important, is not examined here.

Conditions of loading corresponding to repetitive shocks can be created by two or more peak loads applied upon the disk with a phase difference smaller than the restitution period of the disk. This means that the second peak load, possibly, but not necessarily, of the same size as the previous, will be applied while the disk is still deformed. Experiments under such conditions are carried out in the framework of this research.

II.5. Boundary Conditions.

The imposed boundary conditions are:

a. Fixing of all external nodes of the model having z < -0.4 cm, to represent the support of the disk under analysis by the S1 upper surface.
b. Fixing of all external nodes of the model having z > -0.4 cm and z < +0.4 cm for all x and y, except the external nodes having:

x<0 and –2cm < y < -1cm,

x<0 and 1cm > y > 2cm.

This fixing represents the peripheral support of the disk under analysis by the anterior, posterior and lateral ligaments of the spine.

III. FE ANALYSIS RESULTS.

Performance of the analysis was done using the commercial Finite Element Code ANSYS. The cases treated were four, all involving calculation of the impact of the loading upon the 7240 elements to which the disk was partitioned.

Figure 6 gives a graphical example of the loading and the fixed boundary conditions imposed upon the disk. Figures 7, 8 and 9 give the deformed disk under constant compression (Figure 7), under compression rising from 800KPa to 1600KPa along Ox-axis (Figure 8), and under compression rising from 800KPa to 1600KPa along Oy-axis (Figure 9). On the other hand the Table below lists the normal and shear stresses developed at the nodes of the model by application of uniform vertical loadi

TABLE III
AN EXAMPLE OF NORMAL AND SHEAR STRESS COMPONENTS ALONG AXES AND PLANES AT NODE POSITIONS
(IN Mpa UNITS)

Node No	S_x	S_y	S_z	S_{xy}	S_{yx}	S_{xz}
34001	0,0175	-0,0439	0,0178	-0,0049	-0,0186	-0,0029
34001	0,0219	-0,0371	0,0317	-0,0088	0,0138	-0,0004
34001	0,0131	-0,0506	0,0040	-0,0001	-0,0511	-0,0053
34002	0,0340	-0,0149	0,0656	-0,0326	0,0122	0,0062
34002	0,0132	-0,0461	0,0017	-0,0288	-0,0570	-0,0256
34003	0,0165	-0,0474	0,0125	-0,0160	0,0073	0,0018
34003	-0.0005	-0,0694	-0,0356	-0,0114	-0,0559	-0,0114
34006	-0,0877	-0,2841	-0,4425	-0,0084	-0,2672	-0,0121
34008	-0,0028	-0,1136	-0,1219	-0,0105	-0,0532	0,0019
34010	0,0264	-0,0379	0,0094	-0,0104	0,0062	-0,0020
34010	0,0264	-0,0379	0,0094	-0,0104	0,0062	-0,0020
34011	0,0527	0,0024	0,0676	-0,0372	0,0062	0,0025
34012	0,0274	-0,0348	0,0008	-0,0184	0,0060	-0,0001
34015	-0,0195	-0,2965	-0,5038	-0,0192	-0,2460	-0,0301
34017	0,0257	-0,0756	-0,1012	-0,0077	-0,0225	0,0069
34019	0,0680	0,0436	0,0259	-0,0102	0,0391	0,0199
34019	0,0680	0,0436	0,0259	-0,0102	0,0391	0,0199
34020	0,1088	0,0897	0,1052	-0,0215	0,0356	0,0193
34021	0,0605	0,0437	0,0136	-0,0127	0,0402	0,0023
34024	-0,4686	-0,7443	-1,7491	-0,0295	-0,4581	0,0046
34026	-0,0464	-0,1544	-0,3504	-0,0059	-0,0939	-0,0027
34028	-0,0810	-0,3784	-0,7309	-0,0174	-0,3661	-0,0421
34029	0,0041	-0,0696	-0,0200	-0,0117	0,0251	0,0019
34029	0,0041	-0,0696	-0,0200	-0,0117	0,0251	0,0019
34030	0,0472	-0,0407	-0,1183	-0,0036	-0,0599	0,0057
34032	0,0545	0,0073	0,0796	-0,0415	0,0210	0,0090
34034	0,0116	-0,0532	-0,0157	-0,0230	0,0267	0,0038
34037	0,0681	0,0256	0,0248	-0,0086	0,0312	0,0291
34037	0,0681	0,0256	0,0248	-0,0086	0,0312	0,0291
34038	0,0716	0,0342	0,0686	-0,0299	0,0346	0,0232
34039	0,0597	0,0192	0,0142	-0,0157	0,0365	-0,0023
34042	-0,3898	-0,4698	-1,9304	-0,0094	-0,3399	0,0219
34044	0,0764	-0,0785	-0,0120	-0,0102	-0,1378	-0,0156
34046	0,1482	0,2165	0,0036	0,0028	0,0295	0,0099
34046	0,1482	0,2165	0,0036	0,0028	0,0295	0,0099
Mean Absolute	0,0697	0,1147	0,1932	0,0146	0,0780	0,0110
Std. Dev.	0,0977	0,1551	0,4427	0,0101	0,1135	0,0108

IV. DISCUSSION AND CONCLUSIONS.

A numerical analysis of the intervertebral disk with its nucleus has been performed using the FE Analysis Code ANSYS. The FE model developed partitioned the intervertebral disk, both the fibrous and liquid parts, into 7240 elements by using 6900 nodes. This model, although basically equivalent to already published models[22,23,24,27,28,33] has the detail required and can represent all deformations possible to occur in a physical body of less than 5cm in width (Oy-axis), 4 cm in length (Ox-axis) and a little more than 1cm in height (Oz-axis). The side of individual element is about 1.2mm and its volume 1.7mm^3. The effect of lumbar spine curvature on stress distribution in intervertebral disks (Rohlmann et al[40]) is automatically taken care in this case by the choice of local co-ordinate system.

The loading model that has to simulate WBV conditions presents the most complicated part of the general problem under treatment. The experiments conducted were based upon data (anthropometric parameters and loading) taken from the contribution of "Stresses in the lumbar spine due to whole-body vibration containing shocks" by H. Seidel[38] et al, (translation from German by A. W. Bednall). The selected stresses were the maximum listed in this reference but it cannot be claimed, with any certainty at all at the moment, that the real circumstances, where the load is time-dependent, are approximated by the maximum load exerted. Further clarification of the necessity to consider the size of impact of the poroelastic characteristic of annulus fibrosus (Wu et al[41]) and make adjustments of the model studied here, is useful. It is obvious, nevertheless, that escape of liquid from a compressed cavity reduces the internal pressure.

The natural frequency of oscillation of the elasto-plastic fibrous part may be bigger than the loading frequency and as a result new extreme compression load may appear while the disk is still in its deformed condition due to the application of the previous loading.

The experimental loading used for the cases presented above lead to deformations that can be considered compatible to the assumptions made. Uniform stress caused uniform deformations in the Oz direction, linear increase in the stress from left to right along the Oy direction resulted to increase in the deformation along the same direction, and finally increase in the stress from back to front along the Ox direction resulted to increase in the deformation along this direction.

The assumption of non-homogeneous elastic property of the fibrous part of the disk leads to damage of the weak side when stressed.

The conclusions reached at this project phase are:

(a) A FE model created to disclose distortion and failure (rapture) positions require partitioning to elements of linear dimensions equal or less to a fraction of the linear dimension of the distortion or failure. Thus a 1cm linear size disk herniation can be detected by a FE model of elements with a maximum of 2.5 mm linear size. Also a 1mm linear size disk rapture can be detected by a FE model of elements with a maximum of 0.25 mm linear dimension.

(b) The FE model presented here can simulate raptures of 0.4 mm linear size since the distance from node to node is less than 0.1mm. Having such a fine partitioning of the disk FE model, one can disclose practically all sizes of substantial herniations and ruptures even during the first stage of their appearance.

(c) The presence of inhomogeneities in the modulus of elasticy of annulus fibrosus appears to be one serious reason for disk failure, without precluding failures caused by degeneration.

(d) The FE model presented in this paper, on account of its fine partitioning, showed during many tests its suitability for studying even slight deformations caused by time dependent loading applied and, furthermore, covered also cases of small raptures caused by local poorer elastic properties of the annulus fibrosus.

(e) The impact of repetitive shocks during whole body vibration periods can be studied by assuming that the FE model is loaded by successive maxima of the time dependent loading and that each new run with the new maximum load is applied to the model at the deformed geometry left by the previous loading.

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REFERENCES

1. Fairley, TE, Griffen MJ. The apparent mass of the seated human body: Vertical vibration. Journal of Biomechanics, 221989; 22:81-94.
2. Kligenstierna U, Pope MH. Body height changes from vibration. Spine, 1987; 12:566-568.
3. Panjabi M, Anderson GBJ, Jorneus L, Hult E, Mattson L. In vivo measurements of spinal column vibrations. Journal of Bone and Joint Surgery 1986;68A:695-702.
4. Pope MH, Broman H, Hansson T. Factors affecting the dynamic response of the seated subject. Journal of Spinal Diseases 1990; 3:135-142.
5. Pope MH, Wilder DG, Jorneus L, Broman H, Svensson M, Anderson G. The response of the seated human to sinusoidal vibration and impacts. Journal of Biomechanical Engineering 1987;109:279-284.
6. Seroussi RE, Wilder DG, Pope MH. Trunk muscle electromyography and whole body vibration. Journal of Biomechanics 1989;22:219-229.
7. Wilder DG, Woodworth BB, Frymoyer JW, Pope MH. Vibration of the human spine. Spine 1982;7:243-254.
8. Gallante JO. Tensile properties of the human lumbar annulus fibrosus. Acta Orthopaedica Scandinavica (Supplement) 1967;100.

9. Heliovaara M. Occupation and risk of herniated lumbar intervertebral disk leading to hospitalization. Journal of Chronic Diseases 1987;40:259-264.
10. Hirsch C. The reaction of the intervertebral disks to compression forces. Journal of Bone and Joint Surgery 1955;37A:1188-1196.
11. Kelsey JL, Hardy RJ. Driving of motor vehicle as a risk factor for acute herniated lumbar intervertebral disk. American Journal of Epidemiology 1975;102:63-73.
12. Koeller W, Muhlhaus S, Meier W, Hartman F. Biomechanical properties of the human intervertebral disks subjected to axial dynamic compression – influence of age and degeneration. Journal of Biomechanics 1986;19:807-816.
13. Natali A, Meroi E. Non-linear analysis of intervertebral disk under dynamic load. Journal of Engineering 1990;112:358-363.
14. Roaf R. A study of mechanics of spinal injury. Journal of Bone and Joint Surgery 1960;42B:810-823.
15. Moaveni, S. Finite Element Analysis**:** Theory and Application with Ansys, Prentice Hall, 1998: 437-505.
16. H. Gradin, Jr., Fundamentals of the Finite Element Method, 1986; Waveland Press, 1991, 238-255.
17 Ciarlet, PG, and Lions JL. Handbook of Numerical Analysis**:** Finite Element Methods (Part 2), Numerical Methods for Solids (Part 2), Elsevier Science, 1990: 312-326.
18 Masatake . M. The Finite Element Method & Its Applications**,** Macmillan,
19 L. J. Segerlind, Applied Finite Element Analysis, 2d ed. NY: John Wiley & Sons, 1984; 1986:335-343.
20 Shames IH, Dym CL. Energy and Finite Element Methods in Structural Mechanics: SI Units Edition, Taylor & Francis, Inc., 1991: 643-658.
21 O. C. Zienkiewicz, R. Taylor, The Finite Element Method, 4th ed., McGraw-Hill Companies, 1989: 214-225.
22 Laible JP, Pflaster D, Simon BR, Krag MH, Pope MH, Haugh LD. A dynamic material estimation procedure for soft tissue using a poroelastic finite element model. Journal of Biomechanical Engineering 1994;116:19-29.
23 Simon BR, Laible JP, Pflaster D, Yuan Y, Krag MH. A poroelastic finite element formulation including transport and swelling in soft tissue structures. . Journal of Biomechanical Engineering 1994;118:35-43.
24 Simon BR, Vu JSS, Carlton MW, Evans JH, Kazarian LE. Structural models for human spinal motion based on poroelastic view of the intervertebral disk. Journal of Biomechanical Engineering 1985;107:327-343.

25 Shirazi-Adl A. An interface continuous stress penalty formulation for the finite element analysis of composite media. Computers & Structures 1988;33:951-956.

26 Shirazi-Adl A, Shrivastava SC. Large deformation finite element treatment of changes in the volume of fluid-filled cavities enclosed in a structure. Computers & Structures 1989;34:225-230.

27 Shirazi-Adl A. On the fibre composite material models of the disk annulus – comparison of predicted stresses. Journal of Biomechanics 1989;22:357-365.

28 Shirazi-Adl A. Finite-element simulation of the changes in the fluid content of human lumbar disks, mechanical and clinical implications. Spine 1992;17:256-263.

29 Natarajan RN, Ke JH, Andersson GBJ. A model to study the disk degeneration process. Spine 1994;19:259-265.

30 Broman H, Pope MH, Hansson T. A mathematical model of the impact response of the seated sumbect. Medical Engineering & Physics 1996;18:410-419.

31 Goel VK, Monroe BT, Gilbertson LG, Brinckman P. Interlaminar shear stresses and laminae separation in a disk, Finite element analysis of the L3-L4 motion segment subjected to axial compression loads. Spine 1995; 20:689-876.

32 McNally DS, Arridge RGC. An analytical model of intervertebral disk mechanics. Journal of Biomechanics 1995;28:53-68.

33 Laible JP, Pflaster D, Krag MH, Simon BR, Haugh LD. A poroelastic-swelling finite element model with application to the intervertebral disk. Spine 1993;18:659-670.

34 Kitazaki S, Griffin MJ. Resonance behaviour of the seated human body and effects of posture. Journal of Biomechanics, 1998; 31:143-149.

35 Kitazaki S, Griffin MJ. A modal analysis of whole-body vertical vibration, using a finite element model of the human body. Journal of Sound and Vibration 1997; 200:83-103.

36 Matsumoto Y, Griffin MJ. Moving of the upper-body of seated subjects exposed to vertical whole-body vibration at the principal resonance frequency. Journal of Sound and Vibration 1998; 215:743-762.

37 Buck B. Ein Modell für das Schwingungsverhalten des sitzenden Menschen mit detaillierter Abbildung der Wirbelsäule und Muskulatur im Lendenbereich. Doctoral Dissertation, Technischen Hochschule Darmstaadt, 1997.

38 Seidel H, Blüthner R, Hinz B, Schust M. Stresses in the lumbar spine due to whole-body vibration containing shocks, Publisher: Bundesanstalt für Arbeitsschutz und Arbeitsmedizin Fachbereich Arbeitsmedizin, Sitz Berlin Nöldnerstraße 40-42, 10317 Berlin Postfach 5, 10266 Berlin 1998.

39 Lu YM, Hutton WC, and Gharpuray VM. The effect of fluid loss on the viscoelastic behavior of the lumbar intervertebral disk in compression. Journal of Biomechanical Engineering 1998; Volume 120: 48-54.

40 Rohlmann, A, Zander, Th, Calisse, J, and Bergmann, G**.** Influence of lumbar spine curvature on stress distribution in intervertebral disks. Journal of Biomechanics 1998; Volume 31, Supplement 1: 72-82.

41 Wu JSS, Chen JH. Clarification of the mechanical behaviour of spinal motion segments through a three-dimensional poroelastic mixed finite element model. Medical Engineering Physics 1996; Volume 18: 215-224.

42. Nicholson PHF, Haddaway MJ, Davie MWJ, Evans SF. A computerized technique for vertebral morphometry. Physiol. Meas. **14** No 2 (May 1993) 195-204.

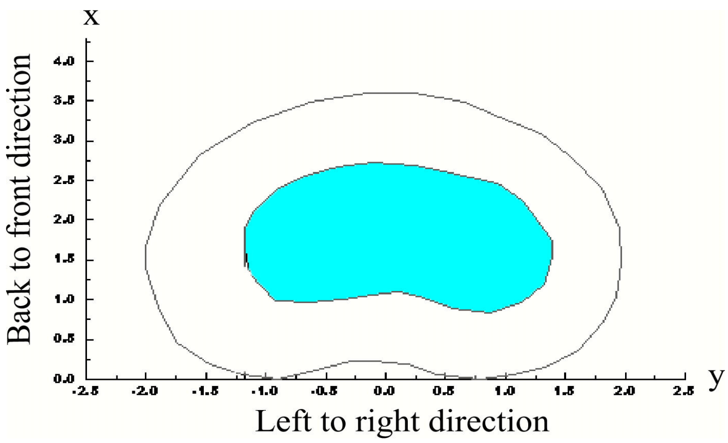

Figure 1. Projection on the Oxy plane of the mid-section of an average L5/S1 intervertebral disc with the hollow part shaded.

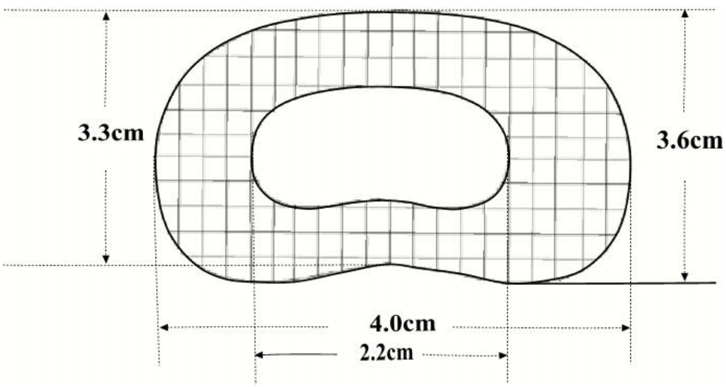

Figure 2. Dimensions of the in the plane of rough symmetry of the fibrous and the fluid part of the disc.

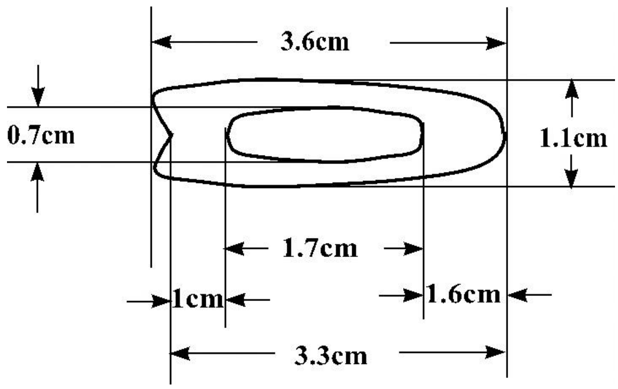

Figure 3. Vertical back to front section and physical dimensions of the disk and its nucleus pulposus.

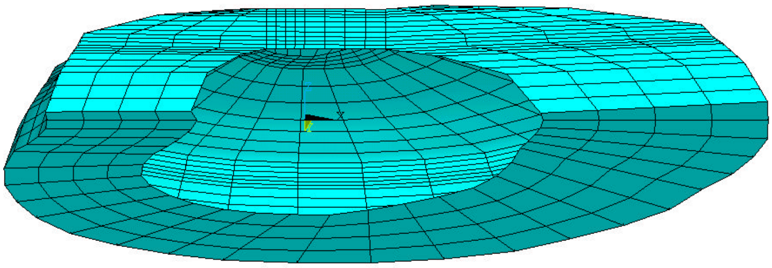

Figure 4. Model of the lower left quarter of the intervertebral disk without its nucleus.

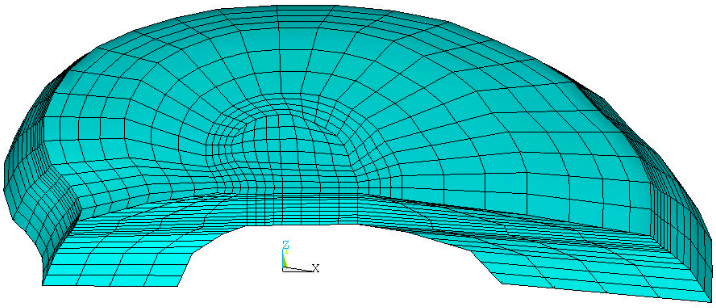

Figure 5. Model of the unstressed upper left quarter of the intervertebral disk without its nucleus. The annulus fibrosus and the nucleus are partitioned into a total of 7240 elements using 6900 nodes.

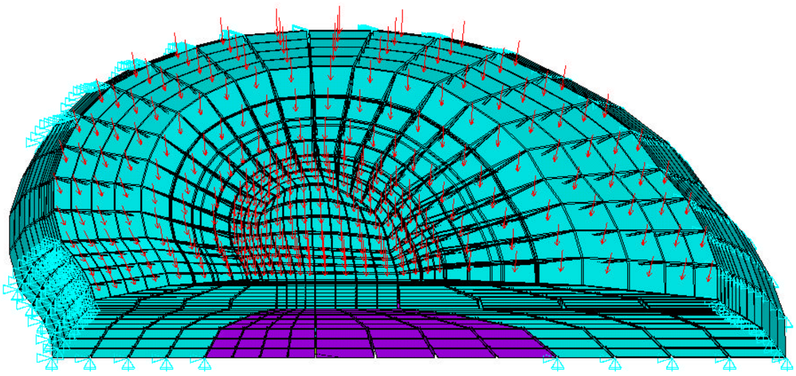

Figure 6. The pressure loading imposed to the model shown by arrows and its boundary conditions (unmovable nodes) shown by small triangles.

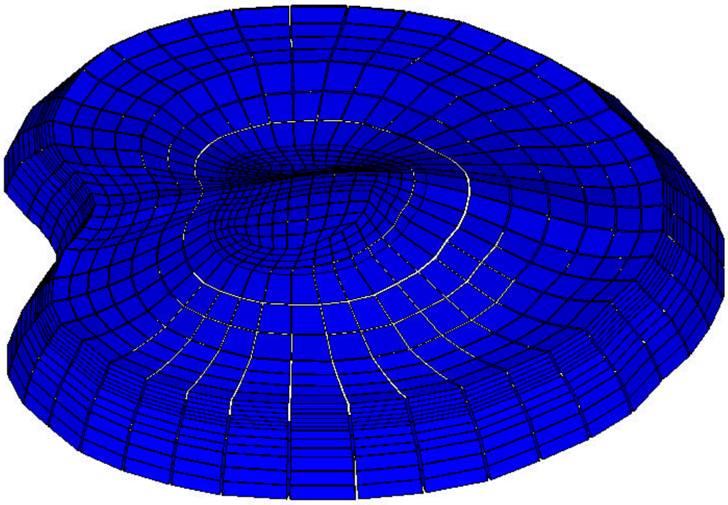

Figure 7. The deformed configuration when the disk is loaded uniformly.

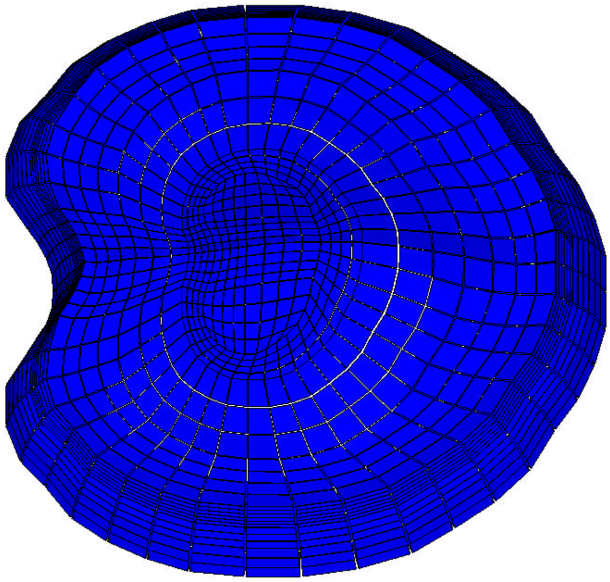

Figure 8. The deformed configuration when the disk is loaded with pressure loading with variation in Ox-axis (bent-forward posture during vibration).

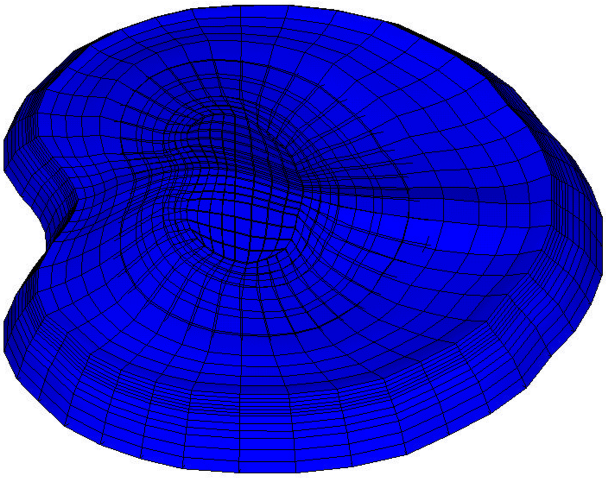

Figure 9. The deformed configuration when the disk is loaded with pressure loading with variation in y-axis (lean-to-left position during vibration).

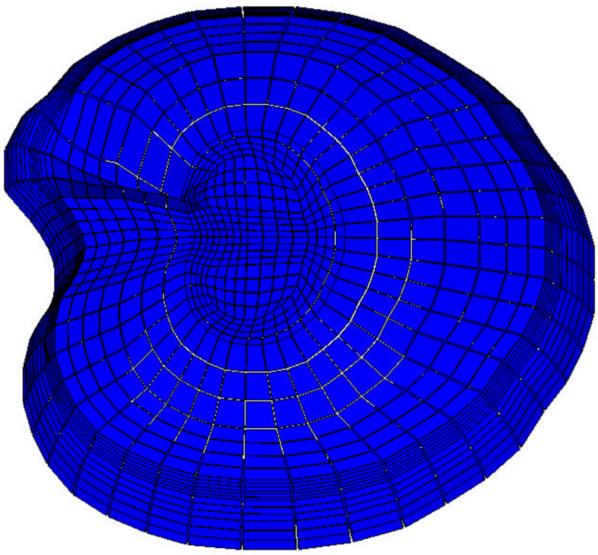

Figure 10. The deformed configuration when the disk is loaded uniformly but with its left half having modulus of elasticity equal to 1500N/cm^2, while the other half has modulus 3000N/cm^2. The weak half developed hernia and finally rapture.