Biomechanics of the Intervertebral Disc in Compression: Experimental and Numerical Study

Ramiro Arturo González Gutiérrez, Vladimir Rangel Alonso and José Gustavo Zambrano Rosas

Department of Metalworking, Instituto Tecnológico de Tijuana, Tijuana 22414, Mexico

Received: November 04, 2014 / Accepted: November 17, 2014 / Published: January 25, 2015.

Abstract: Loading history and age are factors for disc degeneration and disc biomechanics; however, their relationship is unclear. To evaluate disc biomechanics, we conducted an experimental, anatomical and numerical approach to distinguish discs with mild and severe degeneration. In the experimental procedure, 10 cadaveric lumbar discs are tested to static and dynamic compression, and the elastic and viscous moduli and the dynamic parameters are reported. The morphology of disc degeneration is gained with MRI (magnetic resonance imaging) and used to generate a nonlinear finite element model of a degenerated disc, and assisted with the experimental results in order to numerically investigate the distribution of stresses and strains within the disc. The results show a promising methodology for the study of intervertebral disc biomechanics and in general other tissues, organs and medical devices.

Key words: Intervertebral disc, viscoelasticity, testing, magnetic resonance imaging, finite element model.

1. Introduction

The intervertebral disc is a fibrocartilage structure composed of a nucleus pulposus, annulus fibrosis, and cartilage endplate (Fig. 1). The disc along with the facet joints carries all the loading to which the trunk is subjected. This includes compression, shear, bending and torsion, acting together statically or dynamically [1].

The biomechanics of disc loading changes with age and degeneration [1], and its study have been the subject of numerous investigations: experimentally, numerically, and clinically [2-5]. Testing discs in compression serves to evaluate disc stiffness, Young's modulus, and storage and loss moduli [2-4]. From a viscoelastic approach, degenerated discs have shown a reduction in disc stiffness, movement, stress relaxation and damping capacity [5]. Such results represent valuable data as inputs into numerical models, such as the FEM (finite element method), for

Fig. 1 Intervertebral disc components.

investigating mechanical properties of the disc tissues and the distribution of stress and strain inside the disc. Thus, we developed one FEM of a degenerated disc based on magnetic resonance imaging and investigated the stress and strain distribution using also testing results from a group of degenerated discs. The paper is organized as follows: Sections 2-5 give the materials and methodologies; Sections 6-8 present the results from the MRI (magnetic resonance imaging) inspection, testing and the FEM simulation, respectively; Section 9 addresses the implications of the results; and Section 10 gives the main conclusions.

2. The Specimens

Ten lumbar intervertebral discs from elderly donors

Corresponding author: Ramiro Arturo González Gutiérrez, professor, Ph.D., research fields: mechanical design, biomechanics, finite element modeling. E-mail: ramiro.gonzalez@tectijuana.edu.mx.

obtained from Hospital Clinic of Barcelona, Spain were used in this study. After dissection, all specimens were examined to verify absence of any disease, and measurements of the disc height and cross section were recorded before the inspection with MRI. The anatomical data are given in Table 1.

3. MRI Inspection

A Siemens Harmony™ system was used with a spin-echo T_1 and echo-gradient T_2 sequences in accordance with Ref. [6]. The tomography resolution for all discs were 256×256 pixels, field of view of $10.4 \text{ mm} \times 10.4 \text{ mm}$; slide thickness of 0.4 mm , and slide interspace of 0.4 mm . One tomography was used for developing the FEM of disc degeneration, using additionally MIMICS™ (MATERIALISE™, Leuven, Belgium) and MARC MENTAT™ (MSC Software™, Santa Ana CA, USA) software for treatment of medical imaging, and stress and strain analyses, respectively. The procedure used in this study is outlined in Fig. 2.

4. Experimental Protocol

Biomechanical testing to all intervertebral discs was done at the biomaterials laboratory of the CMEM

Table 1 Mean and standard deviation of the disc age, size and height.

Discs	Lumbar level	Sex F = Female M = Male	Age (years)	Cross section (mm ²)	Height (mm)
$n = 5$	L23	F = 1, M = 4	70 (3.5)	1,739 (184)	9.8 (1.8)
$n = 5$	L45	F = 1, M = 4	70 (3.5)	1,951 (223)	10 (1.6)

Fig. 2 Procedure used.

(Departament of Materials Science and Metallurgy) of the Polytechnic University of Catalonia. The experimental protocol was based on previous studies [2, 4] and included axial compression and dynamic compression. The compressive modulus and stress relaxation were determined using static loading and adjusting data to a Kelvin-Voigt model as follows:

$$\sigma(t) = E\varepsilon(t) + \eta \frac{d\varepsilon(t)}{dt} \quad (1)$$

$$E = \frac{\sigma_0}{\varepsilon_0} \quad (2)$$

$$E_R(t) = \frac{\sigma_0}{\varepsilon_0} \exp\left(-\frac{E}{\eta}t\right) \quad (3)$$

$$E_R(t) = \frac{\sigma_0}{\varepsilon_0} \exp\left(-\frac{t}{t_R}\right) \quad (4)$$

where, E is the compressive modulus, $\varepsilon(t)$ is time dependent strain, η is viscosity, σ_0 is stress, ε_0 is strain, E_R is the relaxation modulus, t_R is time decay from σ_0 to a stress equal to $(1/\exp) \cdot \sigma_0$.

The disc storage E' and loss E'' moduli and hysteresis H were determined using cyclic compression and adjusting data as follows:

$$E' = \frac{\sigma_0}{\varepsilon_0} \cos\beta \quad (5)$$

$$E'' = \frac{\sigma_0}{\varepsilon_0} \sin\beta \quad (6)$$

$$\tan\delta = \frac{E''}{E'} \quad (7)$$

$$H = \left[\int_{r_1}^{r_2} \sigma d\varepsilon\right]_{\text{load}} - \left[\int_{r_1}^{r_2} \sigma d\varepsilon\right]_{\text{unload}} \quad (8)$$

where, β is the phase shift between stress and strain, and $\tan\delta$ and H represents the ratio of storage to loss moduli and the hysteresis, respectively.

5. Numerical Protocol

Thresholding of the disc and vertebrae was the first step in creating the FEM. This task included creation of masks for the various segmentations through Boolean algebra, and identification of cortical and cancellous bone from the vertebrae, and nucleus, annulus, endplate and cavities from the disc. The

procedure in MIMICS™ is outlined in Fig. 3, and a 3D view of the regions of segmentation is shown in Fig. 4.

The finite element mesh was created in MARC™ and was based on tetrahedral element # 157 with four nodes, ideally suitable for incompressible materials. Five models were developed base on the triangle size: 1 mm, 1.5 mm, 2 mm, 3 mm and 5 mm, which were in accordance with the size of most degenerative features. Material formulation included orthotropy for cancellous and cortical bone, with values of the Young's moduli in the range of $140 \text{ MPa} < E < 22,000 \text{ MPa}$, Shear modulus $48 \text{ MPa} < G < 5,400 \text{ MPa}$, Poisson ratio $0.20 < \nu < 0.48$; disc materials were treated as incompressible and elastomeric with a formulation based on the Mooney-Rivlin solid model [7, 8]:

Fig. 3 Segmentation procedure for bone and disc.

Fig. 4 Disc model showing segmentation.

$$W = C_1(I_1 - 3) + C_2(I_2 - 3) \quad (9)$$

where, I_1 and I_2 are the first and second invariant of the Cauchy strain tensor, C_1 and C_2 are empiric constants of the material with values shown in Table 2.

The constants for the annulus fibrosus were kept fix ($C_1 = 0.10$ and $C_2 = 0.025$) and only the constants for the nucleus pulposus were adjusted, on the basis that under compression, the main deformation occurs in the nucleus [9]. The criterion used for the adjustment of the Mooney-Rivlin constants were based on the reaction F.D. (force difference) between those from the simulation (F_s) and the testing (1,000 N), as follows:

$$F.D. = \sqrt{(F_s - 1,000)^2} \quad (10)$$

All disc models were submitted to the same boundary conditions corresponding to the compression testing. The type of analysis used was a nonlinear static and the displacements, reaction force, stress and strain for the intervertebral disc and vertebrae materials were investigated.

6. Degenerative Features with MRI

The MRI inspection revealed that the lumbar level was sensitive with degeneration scoring (student T-test, $P < 0.05$). The main degenerative features identified in the intervertebral discs are shown in Fig. 5.

In general, cavity formation appeared in the nucleus region and in the annulus of all discs to a lesser extent. However, the L45 discs showed more cavities with a clear collapse of the disc height, abundance of schmorl's nodes in the disc-vertebrae boundary, bone reactive sclerosis and bone formation at the annulus anterior and posterior sides. Thus, L23 discs with maintenance of the disc height were assigned a mild

Table 2 Mooney-Rivlin C_1 and C_2 constants [8].

Material	Formulation	C_1 and C_2 value range
Annulus	Elastomeric	$0.10 < C_1 < 0.20$
Fibrosis	Mooney-Rivlin	$0.025 < C_2 < 0.05$
Nucleus	Elastomeric	$0.03 < C_1 < 0.12$
pulpous	Mooney-Rivlin	$0.01 < C_2 < 0.03$
Endplate	Isotropic	$E = 20 \text{ MPa}$, $\nu = 0.30$

Fig. 5 Degenerative features for mild and severe scoring.

degeneration scoring, and discs with a collapse of the disc space typically from the lower L4/5 level were assigned a severe degeneration scoring.

7. Disc under Static and Dynamic Compression

Under static and dynamic loading, there were no significant differences among mechanical properties and lumbar level (student T-test, $P > 0.05$) due to the large scattering. However, under static compression, the discs showed a nonlinear stress-strain relationship with large deformations axially and radially. Stress relaxation was faster in highly degenerated discs which showed a lower relaxation modulus (Table 3).

Increase in axial stiffness and dissipation was observed with increase in frequency loading. Storage

modulus E' decreases up to 20% less in discs with severe degeneration, although the loss modulus was about even between degeneration scorings. In general, the mean storage modulus E' was around 20 times larger than the mean loss modulus E'' (Table 4).

8. FEM of Disc Degeneration and Compression Simulation

Based on the convergence test, the most convenient model was that of an element size of 1.5 mm. The selected model included 187,746 elements and 43,172 nodes spread in six regional material masks. Next, the initial selection of the Mooney-Rivlin constants for the nucleus pulposus were $C_1 = 0.12$ and $C_2 = 0.03$, these values were modified as a function of the reaction force. A 1% difference in F.D. was used as the tolerance. The resulting values were $C_1 = 0.07$ and $C_2 = 0.02$. The algorithm for the adjustment of the Mooney-Rivlin constants is shown in Fig. 6.

The strain distribution showed that the nucleus bears most of the highest deformation, reaching -55%, the annulus distribution showed strains from +10% to -40%. In contrast, the vertebrae deformed less than 5%, with 0.25% at cortical region (Fig. 7).

Table 3 Mean values and standard deviation for the motion response under static compression.

Parameter	Degeneration scoring	
	Mild	Severe
Stiffness (N/mm)	1,277 (622)	1,206 (562)
Deformation (%)	9.55 (3.44)	10.27 (4.27)
Radial bulging (mm)	0.33 (0.35)	0.87 (0.62)
Young's modulus (MPa)	14.80 (12.30)	10.63 (2.80)
Relaxation time (s)	1,193 (307)	962 (293)
Relaxation modulus (MPa)	3.80 (3.52)	2.36 (0.61)

Table 4 Mean values and standard deviation for the motion response under dynamic compression.

Frequency (Hz)	Illness scoring	E' modulus (MPa)	E'' modulus (MPa)	Hysteresis (J)
0.1	Mild	4.5 (2.5)	0.25 (0.22)	0.11 (0.08)
	Severe	4.0 (2.0)	0.25 (0.20)	0.11 (0.08)
1	Mild	4.8 (1.9)	0.30 (0.25)	1.02 (0.87)
	Severe	4.5 (2.3)	0.75 (0.33)	0.60 (0.25)
5	Mild	8.0 (3.4)	2.1 (1.1)	3.15 (1.50)
	Severe	6.5 (2.7)	3.0 (1.3)	2.70 (1.72)

Fig. 6 Algorithm used for adjustment of properties.

Fig. 7 Stress and strain distributions in the FEM.

The stress distribution showed that the nucleus is subjected to a state of compression, reaching -1 MPa, whereas the annulus is put to tensile and compressive states of stress ranging from +0.25 MPa to -1.25 MPa (Fig. 7). The principal stresses in the vertebral regions were from 4 MPa to -8 MPa in tensile and compression, respectively. The cortical bone was the most stressed part due to its high rigidity.

9. Discussion

The MRI findings indicate that discs from the lower lumbar level L45 are more likely to exhibit disc

collapse and disruption, which implies a severe loading history in accordance with the clinical experience. The schmorl's nodes and osteophyte formation imply severe degeneration, and with disc disruption limited the identification of the nucleus, annulus and endplate. In contrast, void formation around the nucleus was a reliable indicator of early stages of degeneration, as described elsewhere [10, 11]. Thus, the MRI from the L23 disc was more appropriate for identification of regional materials and developing a FEM of disc degeneration. Segmentation between vertebrae and disc was straightforward, and between annulus, nucleus and endplate depended on contrast.

The biomechanical testing indicates that degeneration diminished their viscoelastic properties, stress relaxation, damping and dissipation capacities, in accordance with previous studies [12-14]. However, the large scatter in the results was normal which caused a poor correlation. In general, individual differences in lumbar disc mechanical behavior usually overshadow any class differences with respect to age, sex, disc level, or degree of degeneration [2].

Other disc models based on medical images have been developed in Refs. [15, 16], but only few have included experimental results. The disc model presented here is suitable for the study of stress and strains. The adjustment of the Mooney-Rivlin constants C_1 and C_2 for the nucleus correspond to a mild degeneration state which is consistent with the MRI and previous numerical studies [8]. The large deformations in the annulus and nucleus are due to the incompressible and elastomeric formulations, which takes into account the high content of water in the disc. The model also predicts bulging in the annulus and nucleus in accordance with Ref. [17] being larger in the anterior sides, where more water is present. The stress distribution indicates that the nucleus is subjected to a compressive state and suggests that the nucleus is the main carrier of the load as described elsewhere [8, 9].

10. Conclusions

The proposed methodology is novel, versatile, functional and economic. Therefore, its application has potential for the medical sciences, clinical field, design and innovation of medical devices, implants, prosthesis and its manufacture.

Acknowledgements

The authors will like to express their gratitude to the following researchers: M. D. Manel Llusà, M. D. Jaume Pomes, Dr. Josep Planell and Dr. Damien Lacroix and our alma mater Instituto Tecnológico de Tijuana.

References

- [1] Adams, M. A., and Roughley, P. J. 2006. "What Is Intervertebral Disc Degeneration, and What Causes It?" *Spine* 31: 2151-61.
- [2] Nachemson, A., Schultz, A. B., and Berkson, M. H. 1979. "Mechanical Properties of Human Lumbar Spine Motion Segments. Influence of Age, Sex, Disc Level and Degeneration." *Spine* 4 (1): 1-8.
- [3] Schultz, A. B., Berkson, M. H., and Nachemson, A. L. 1979. "Mechanical Properties of Lumbar Human Spine Motion Segments—Part II: Responses in Compression and Shear; Influence of Gross Morphology." *J. Biomech. Eng.* 101 (1): 53-7.
- [4] Kasra, M., Shirazi, A. A., and Drouin, G. 1992. "Dynamics of Human Lumbar Intervertebral Joints: Experimental and Finite Element Investigations." *Spine* 17 (1): 93-102.
- [5] Kazarian, L. E. 1975. "Creep Characteristics the Human Spinal Column." *Orthop. Clin. North A.* 6 (1): 3-18.
- [6] Brinckmann, P., Frobin, P. W., Kramer, M., and Hartwig, E. 2001. "Height of Lumbar Discs Measured from Radiographs Compared with Degeneration and Height Classified from MR Images." *European Radiology* 11: 263-9.
- [7] Lu, Y. M., Hutton, W. C., and Gharpuray, V. M., 1996. "Do Bending, Twisting, and Diurnal Fluid Changes in the Disc Affect the Propensity to Prolapse? A Viscoelastic Finite Element Model." *Spine* 21: 2570-9.
- [8] Schmidt H., Kettler A., Heuer, F., Simon, U., Claes, L., and Wilke, H. J. 2007. "Intradiscal Pressure, Shear Strain, and Fiber Strain in the Intervertebral Disc under Combine Loading." *Spine* 32: 748-55.
- [9] Noailly, J., Wilke, H. J., Planell, J., and Lacroix, D. 2007. "How Does the Geometry Affect the Internal Biomechanics of a Lumbar Spine Bi-segment Finite Element Model? Consequences on the Validation Process." *Journal of Biomechanics* 40 (11): 2414-25.
- [10] Thompson, J. P., Pearce, R. H., Schechter, M. T., Adams, M. E., Tsang, I. K., and Bishop, P. B. 1990. "Preliminary Evaluation of a Scheme for Grading the Gross Morphology of the Human Intervertebral Disc." *Spine* 15 (5): 411-5.
- [11] Pfirrmann, C., Metzdorf, A., Zanetti, M., Hodler, J., and Boos, N. 2001. "Magnetic Resonance Classification of Lumbar Intervertebral Disc Degeneration." *Spine* 26 (17): 1873-8.
- [12] Keller, T. S., Spengler, D. M., and Hansson, T. H. 1987. "Mechanical Behavior of the Human Lumbar Spine. I. Creep Analysis during Static Compressive Loading." *Journal of Orthopedic Research* 5 (4): 467-78.
- [13] Allen, K., and Athanasiou, K. 2006. "Viscoelastic Characterization of the Porcine Temporomandibular Joint Disc under Unconfined Compression." *J. Biomechanics* 39: 312-22.
- [14] Tanaka, E., Aoyama, J, Tanaka, M., Murata, H., Hamada, T., and Tanne, K. 2002. "Dynamic Properties of Bovine Temporomandibular Joint Disks Change with Age". *Journal of Dentistry Research* 81 (9): 618-22.
- [15] Li, H., and Wang, Z. 2006. Intervertebral Disc Biomechanical Analysis Using the Finite Element Modeling Based on Medical Images." *Computerized Medical Imaging and Graphics* 30: 363-70.
- [16] Wang, Z., and Li, H. 2005. "A Novel 3D Finite Element Modeling Based on Medical Image for Intervertebral Disc Biomechanical Analysis." Presented at the 27th Annual Conference of the IEEE Engineering in Medicine and Biology, Shanghai, China.
- [17] Schmidt, H., Kettler, A., Rohlmann, A., Claes, L., and Wilke, H. J. 2007. "The Risk of Disc Prolapses with Complex Loading in Different Degrees of Disc Degeneration—A Finite Element Analysis." *Clinical Biomechanics* 22 (9): 988-98.