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Inclusion of Regional Poroelastic Material Properties Better Predicts Biomechanical Behavior of Lumbar Discs Subjected to Dynamic Loading

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Abstract

Understanding the relationship between repetitive lifting and the breakdown of disc tissue over several years of exposure is difficult to study in vivo and in vitro. The aim of this investigation was to develop a three-dimensional poroelastic finite element model of a lumbar motion segment that reflects the biological properties and behaviors of in vivo disc tissues including swelling pressure due to the proteoglycans and strain dependent permeability and porosity. It was hypothesized that when modeling the annulus, prescribing tissue specific material properties will not be adequate for studying the in vivo loading and unloading behavior of the disc. Rather, regional variations of these properties, which are known to exist within the annulus, must also be included. Finite element predictions were compared to in vivo measurements published by Tyrrell et al., (Tyrrell et al., 1985) of percent change in total stature for two loading protocols, short-term creep loading and standing recovery and short-term cyclic loading with standing recovery. The model in which the regional variations of material properties in the annulus had been included provided an overall better prediction of the in vivo behavior as compared to the model in which the annulus properties were assumed to be homogenous. This model will now be used to study the relationship between repetitive lifting and disc degeneration.

Keywords

Finite element model; cyclic loading; lumbar motion segment; circadian variation in height; poroelasticity

1. Introduction

Low back disorders are common among individuals who participate in manual material handling (MMH) occupations. Studies have demonstrated this population's predisposition to early onset of disc degeneration. However, understanding the relationship between repetitive lifting and the breakdown of disc tissue over several years of exposure is difficult to study in vivo and in vitro; hence suggesting the use of finite element analysis.

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It is the unique interaction between the solid matrix and interstitial fluid, which provides the disc with the strength and flexibility necessary to withstand the large motions the spine undergoes during even normal daily activities. Fluid flow characteristics are regulated mechanically by parameters including the permeability and porosity of the solid matrix and biochemically by the proteoglycan content of the disc tissues. During degeneration, the disc undergoes morphological and biochemical changes that alter tissue hydration, permeability and ultimately the load bearing capacity of the disc. Therefore, a finite element model developed for studying the relationship between repetitive lifting and disc degeneration must be able to account for both the morphological and biochemical changes that accompany injury and degeneration and be able to withstand the large, complex loading conditions known to occur *in vivo*.

This objective can be accomplished by developing a three dimensional, poroelastic finite element model of a lumbar motion segment that includes physiological parameters such as swelling pressure and strain dependent permeability and porosity of the disc tissues. Studies using poroelastic finite element models have suggested a relationship between relative fluid flow and the mechanical response of the entire motion segment due to sustained loads (Argoubi et al., 1996; Laible et al., 1993; Lee et al., 2000; Riches et al., 2002; Simon et al., 1985). However, these models have not included parameters related to the biochemical constituents of the tissues and hence have been limited to only studying creep loading and not unloading.

Recently, Iatridis et al., developed the PEACE model which includes fluid flow as well as mechanical and electrochemical effects (Iatridis et al., 2003). While the authors elegantly demonstrated the importance of using a formulation which accounts for the mechanical as well as electrochemical influences, the formulation was based loosely on the actual biochemical content of the disc tissues. As a result, modeling injury and disc degeneration based on *in vivo* histological or biochemical data would be extremely difficult. Also, this model does not account for strain dependent permeability and porosity.

Therefore, the short-term objective of this study was to develop a three-dimensional poroelastic finite element model of an L4-L5 motion segment that reflects the biological properties and behaviors of *in vivo* disc tissues including swelling pressure due to the proteoglycans and strain dependent permeability and porosity. Biochemical and biomechanical studies have indicated the distribution of the proteoglycans, water content and the initial permeability and porosity varies regionally within the annulus (Best et al., 1994; Ebara et al., 1996; Gu et al., 1999; Iatridis et al., 1996; Urban et al., 1981; Urban et al., 1988); hence suggesting that in the radial direction, the annulus could be divided into two regions, the inner and outer annulus. But whether the gross mechanical response of the disc over short periods of creep and cyclic loading would be sensitive to these regional variations within the annulus remains unclear. Hence, we hypothesized that when modeling the annulus, prescribing tissue specific material properties (i.e. annulus and nucleus) will not be adequate for studying cyclic loading. Rather, regional variations of these properties, which are known to exist within the annulus, must also be included. Once developed, the future purpose of this model will be studying the relationship between repetitive lifting and disc injury.

2. Methods

Poroelastic Finite Element Model

The poroelastic three-dimensional finite element model (Natarajan et al., 2003) was based on the geometric shape of a lumbar motion segment (L4-L5) generated from a serial computed axial tomographic scan (CT). The model includes both anterior and posterior segments, Figure 1. The salient features of the model are provided in Table 1. In the original poroelastic finite element model, the annulus is defined as a single structure with homogeneously distributed

material properties (Tissue Specific Model, TSM). Drained elastic material properties as well as poroelastic properties of the components for the annulus and nucleus were taken from literature (Table 2). Analyses were conducted using a commercially available software package ADINA (ADINA R&D Inc., Watertown, Massachusetts).

To test the hypothesis, the TSM was modified such that the annulus was subdivided into an inner and outer annulus in which unique elastic and poroelastic material properties were defined for each region. The elastic and poroelastic material properties of the annulus were varied regionally based on data available in the literature (Regional Variation Model, RVM), Table 3.

Incorporation of Physiological Parameters

Several physiological parameters were included in the formulation in order for the model to mimic, as accurately as possible, the *in vivo* biomechanical behavior of the motion segment under loading and unloading. The initial nucleus pressure P_f for an externally applied compressive load of F , was calculated using an equation developed by Broberg (Broberg, 1993):

$$P_f = a + b \left(\frac{F}{F_0} \right) + c \left(\frac{F}{F_0} \right)^2 + d \left(\frac{F}{F_0} \right)^3 \quad (1)$$

where a , b , c and d are experimentally determined constants equal to 0.0042 MPa, 0.913 MPa, -0.032 MPa and 0.0031 MPa respectively, and F_0 is equal to 1000 N (Broberg, 1993). This initial nucleus pressure based on a compressive load of 400 N, was applied as a pore pressure at all nodes in the elements representing the nucleus.

Proteoglycans contained within the disc tissues are responsible for the inflow of fluid during recovery or unloading and resisting the outward flow of fluid during loading. The interaction between the proteoglycans and water results in a fixed charge density which dictates the initial hydration of the tissues and varies inversely to the volume of fluid within the disc at any point in time; hence, directly relating the electrochemical effect to the mechanical loading and unloading of the disc (Urban et al., 1981). Broberg defined such a relationship between fixed charge density and fluid volume as (Broberg, 1993):

$$f_i = \frac{m}{WC * V_i} \quad (2)$$

where m is a reference volume, WC is the water content of the disc and V_i is the volume of the disc as defined by the geometry of the disc in the finite element model at time t_i .

Using this expression, the swelling pressure can then be expressed as a function of the fixed charge density by (Broberg, 1993):

$$P_{\text{swell}_i} = P_f \frac{f_i^2 + 1}{\alpha f_i^2 + 1} \quad (3)$$

where P and α are constants equal to 0.66 MPa and 0.15 respectively. Prescribing an initial water content values within the TSM for the nucleus (82%) and annulus (72%), and in the case of the RVM inner annulus (72%), outer annulus (65%) and nucleus (82%), reflects the collagen and proteoglycan contents of these individual tissues. By prescribing unique WC values for the inner annulus, outer annulus and nucleus within the RVM, as the volume of fluid within these tissues changes in response to loading and unloading, the corresponding fixed charge density and swelling pressures within the tissues will also fluctuate independently.

As the solid matrix deforms in response to the externally applied load, the pores through which the fluid flows also deform and collapse. This dilatation of the pores results in an increased resistance to the outward flow of fluid (DiSilvestro et al., 2001; Holmes et al., 1985; Kwan et al., 1990). To account for the decreasing permeability with increasing disc strain, another pressure p_{strain_i} (Iatridis et al., 1998) was introduced in the disc.

$$P_{strain_i} = \frac{(Ee_i) - \left(\frac{1}{2} H_A \frac{(e_i+1)^2 - 1}{(e_i+1)^{2\beta+1}} \exp \left[\beta \left((e_i+1)^2 - 1 \right) \right] \right)}{-\varphi_i} \quad (4)$$

where E is the drained elastic modulus, e_i is the axial strain, H_A is the aggregate modulus, β is the non-linear stiffness coefficient and φ_i is the porosity at time t_i .

To accommodate non-uniform deformation of the disc, the annulus and nucleus were divided into quadrants. The quadrants, which divided the disc into anterior, posterior and right and left lateral portions, were chosen based on the loading conditions involving only compression and flexion. Calculating the volume of each quadrant in each region separately provides more accurate values for volume change that in turn allow swelling and strain pressures to vary regionally within a tissue.

A FORTRAN subroutine was written to calculate the swelling pressure and strain dependent pressure, at the end of every time step, which was then included in the “user-supplied load” routine available in ADINA. The inclusion of the pressure calculations within the subroutine and the interaction between the subroutine and the finite element model (RVM) is illustrated in Figure 2. Pressures were applied at disc-endplate interface.

Convergence Study

A convergence study was conducted by refining the original coarse finite element mesh (TSM) of the annulus, nucleus and endplates. Analyses using three mesh densities were conducted (1) with original coarse mesh, (2) with mesh obtained by subdividing each element in the original mesh into four new elements (intermediate mesh) and (3) with finer mesh obtained by subdividing each element in the original mesh into sixteen new elements (fine mesh). The optimal mesh density was determined by comparing the model predictions with each mesh density with results of circadian variation measured *in vivo* by Tyrrell et al. (Tyrrell et al., 1985). To predict the change in stature over 24 hours, the lumbar motion segment models were subjected to a compressive creep load of 850 N for 16 hours, simulating normal daily activities without participating in physical work or exercise, followed by a compressive creep load of 400 N, simulating the loading during sleep for 8 hours.

Disc height change predicted by the finite element models was compared with percent change in total stature reported by Tyrrell et al. (Tyrrell et al., 1985). In order to make this comparison it was assumed that the lumbar, thoracic and cervical spine will undergo the same deformation (Krag et al., 1990). If we assume that one third of the total stature loss is in the lumbar spine

and that each lumbar disc deforms equally, then each lumbar disc will see one fifth of one third or one fifteenth of the total stature loss. Hence, the disc height change predicted by the finite element models were multiplied by 15 (Krag et al., 1990) and compared with stature change reported by Tyrrell et al. (Tyrrell et al., 1985). This method was previously used to compare the ability of the TSM to predict circadian variation in total stature to *in vivo* measurements reported by Tyrrell et al. (Tyrrell et al., 1985). The model results were consistent with the *in vivo* measurements (Natarajan et al., 2004).

Short-term Creep and Cyclic Loading

Using the intermediate mesh density, analyses were conducted using the TSM and RVM in order predict changes in stature in response to (1) short-term creep load and standing recovery and (2) short-term cyclic loading and standing recovery. Model predictions were compared with those observed *in vivo* (Tyrrell et al., 1985). In the short-term creep study conducted by Tyrrell et al., a normal subject held a 40 kg barbell across his shoulders for 20 minutes, at which time the barbell was removed for a 10-minute recovery period. The loading profile used in the model analyses is shown in Figure 4a, in which a pressure equivalent to 400 N was applied as a preload followed by an additional 400 N for 20 minutes on the superior surface of L4, after which the load was reduced to 400 N for 10 minutes for recovery. Comparisons of percent change in total stature were made at 2 minutes and 22 minutes after the start of creep loading in order to see how the finite element model compares with the *in vivo* results during the initial loading and unloading and at the end of loading (20 minutes) and unloading (30 minutes).

In the same study, a normal subject lifted a 40kg dumbbell from floor to shoulder height at a rate of 12 lifts per minute for 20 minutes followed by a 10 minute standing recovery. This protocol was simulated in the finite element models by applying a preload of 400 N in conjunction with a peak-to-peak compressive force of 400 N and a 5 Nm peak-to-peak flexion moment at a rate of 12 lifts per minute for 20 minutes. After 20 minutes, the disc was allowed to relax for 10 minutes with just the preload (400 N) present. The loading profile for a single cycle is shown in Figure 4b. The resulting percent change in total stature was compared with those observed *in vivo* at the same time points mentioned previously (Tyrrell et al., 1985).

3. Results

Convergence and validation of the finite element model

The circadian variation in disc height as predicted by the TSM with three different mesh densities were compared with the *in vivo* data (Tyrrell et al., 1985) (Figure 3). Due to the insignificant differences between the results of the intermediate mesh as compared to the fine mesh, the intermediate mesh density was used for the remaining analyses.

Short-term Creep and Cyclic Loading

The distribution of the percent loss of total stature predicted by the RVM more closely compared to the *in vivo* results throughout loading as well as unloading (Figure 5). The abrupt change in stature at the end of 2 minutes after creep loading as well as unloading was within 5% of the corresponding *in vivo* results. RVM predictions of total stature change at the end of loading (20 minutes) and unloading (30 minutes) were also within 4% of the corresponding *in vivo* measurements and the total stature loss at the end of creep unloading also matched the *in vivo* results (Tyrrell et al., 1985).

Loss of total stature obtained from the RVM during short-term cyclic loading compared very well with *in vivo* (Tyrrell et al., 1985) results as compared to the TSM (Figure 6). Comparisons of percent change in total stature between the RVM predictions and the *in vivo* results (Tyrrell et al., 1985) were made at 2 minutes, 20 minutes, 22 minutes and 30 minutes. At each time

point, the finite element model predictions were within 3% of *in vivo* results including at the end of 10 minute recovery period.

4. Discussion

A three-dimensional poroelastic finite element model of a lumbar motion segment that includes biomechanical and biochemical parameters, which dictate the fluid flow characteristics of the disc during loading and unloading, was developed. Convergence of the finite element model results was accomplished by refining the mesh and comparing the model predictions with results of circadian variation measured *in vivo* (Tyrrell et al., 1985).

The results presented here indicate that while inclusion of physiological parameters is necessary for modeling the interaction between the fluid and the solid matrix, including regional variations in the elastic and poroelastic material properties of the disc tissues yields biomechanical predictions which best mimic those observed *in vivo* thus validating the current finite element formulation.

While the validation method is not the most rigorous, we were limited by the availability of data in the literature by which our model could be compared with. The study conducted by Tyrrell et al. was the most comprehensive and provided adequate detail of the testing method that we could model a similar situation. We acknowledge that the comparison of disc height loss, or percent change in stature, is a one-dimensional comparison and may not yield the most comprehensive validation for a model of such complexity. It is important to note that this one-dimensional behavior is dependent on the interaction of the solid matrix and the fluid in all three dimensions. Furthermore, a three-dimensional model is essential if we are going to use it to study the relationship between repetitive lifting and disc degeneration due to the highly complex and multi-directional loading conditions experienced by the spine.

Another limitation of this model is the definition of the quadrants. In the current study the authors chose to perform the swelling pressure and strain pressure calculations for each region (inner and outer annulus and nucleus) for the anterior, posterior, right lateral and left lateral volumes. Based on the relative simplicity of the loading conditions used (compression and compression with flexion moment) the authors felt that this approximation was sufficient. In the future, the calculations will be conducted on an element by element basis so as to be able to accommodate asymmetric loading known to occur in lifts associated with manual material handling.

A three-dimensional poroelastic finite element model of an L4-L5 motion segment that reflects the biological properties and behaviors of *in vivo* disc tissues has been developed. The results of this study support the hypothesis that simply prescribing tissue specific material properties are inadequate for predicting the biomechanical response of the intervertebral disc during short-term creep and short-term cyclic loading and unloading. Rather, regional variations of these properties, which are known to exist within the annulus, must also be included.

This model can now be used to study failure initiation and propagation as well as various pathological conditions. Studies are currently underway investigating the effects of different lifting modes, influence of isolated changes in disc health and the sensitivity of disc mechanics to mechanical (annular tears, changes in collagen content) and biochemical (decreased PG content and hydration) changes known to be part of the degeneration process.

Acknowledgments

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Figure 1.

Figure illustrates the finite element model of an L4-L5 motion segment with the intermediate mesh density, which was used as the basis for this investigation.

Figure 2.

Flow chart illustrating the methodology used to calculate the swelling pressure and strain pressure in the RVM. Nodal displacements are calculated within the finite element software ADINA and passed to a FORTRAN subroutine.

Figure 3.

Comparison of the percent change in total stature over the 24-hour period predicted by the finite element models (Tissue Specific Model –TSM model) based on three different mesh densities with *in vivo* results published by Tyrrell et al. (Tyrrell et al., 1985).

Figure 4.

(a) Loading profile for short-term creep loading and unloading. (b) Loading profile for a single cycle for the short-term cyclic loading. Both the short-term creep and short-term cyclic loading protocols were based on the lifting conditions used in vivo and described in Tyrrell et al. (Tyrrell et al., 1985).

Figure 5.

Comparison of the percent change in total stature over the 20-minutes of creep loading and 10-minute recovery predicted by the finite element models (TSM and RVM with intermediate mesh) with *in vivo* results published by Tyrrell et al. (Tyrrell et al., 1985).

Figure 6.

Comparison of the percent change in total stature over the 20-minute period of cyclic loading followed by 10 minutes of standing recovery as predicted by the finite element models (TSM and RVM with intermediate mesh) with *in vivo* results published by Tyrrell et al. (Tyrrell et al., 1985).

Element and material model information for all components modeled in the Tissue Specific Model (TSM). All material properties were taken from the literature (Goel et al., 1988; Koeller et al., 1986; Panjabi et al., 1984; Sanjeevi et al., 1982; Sharma et al., 1995)

Table 1

Structure	Type of Element	No. of Elements	Material Model	Elastic Material Properties	
				Elastic Modulus	Poisson's Ratio
Cortical Bone	3-D Solid (8 node)	1759	Elastic	12 GPa	0.3
Cancellous Bone	3-D Solid (8 node)	3112	Poroelastic	100 MPa	0.2
Posterior Elements	3-D Solid (8 node)	2112	Elastic	3.5 GPa	0.25
Nucleus	3-D Solid (8 node)	720	Poroelastic	1.0 MPa	0.45
Annulus	3-D Solid (8 node)	1920	Poroelastic	2.5 MPa	0.4
Annular Fibers	Truss	1760	Nonlinear elastic	-	-
Ligaments	Truss	32	Nonlinear elastic	-	-
Facet Cartilage	3-D Solid (8 node)	192	Elastic	11 MPa	0.4
Facet Contacts	Contact	24	-	-	-
Endplate	3-D Solid (8 node)	264	Poroelastic	20 MPa	0.4

Within the TSM tissue specific elastic and poroelastic properties were defined for the annulus and nucleus. Values for permeability, porosity and aggregate modulus and nonlinear stiffness coefficient were taken from the literature (Best et al., 1994; Ebara et al., 1996; Gu et al., 1999; Iatridis et al., 1996; Urban et al., 1981; Urban et al., 1988).

Table 2

Structure	Permeability			Porosity	Water Content	Young's Modulus	Aggregate Modulus	Density	Nonlinear Stiffness Coefficient	
	k_x	k_y	k_z						ρ	β
	$m^4 N^{-1} s^{-1}$	$m^4 N^{-1} s^{-1}$	$m^4 N^{-1} s^{-1}$	ϕ	WC	E	H_A			
Annulus	1.87E-15	1.87E-15	1.56E-15	0.78	72	2.56	MPa	1060		2.7
Nucleus	2.13E-15	2.13E-15	1.45E-15	0.83	82	1.56	0.27	1000		2.7

Within the RVM, the annulus was subdivided into and inner and outer annulus. Unique elastic and poroelastic properties were defined for each region of the annulus as well as for the nucleus. Values for permeability, porosity and aggregate modulus and nonlinear stiffness coefficient were taken from the literature (Best et al., 1994; Ebara et al., 1996; Gu et al., 1999; Iatridis et al., 1996; Urban et al., 1981; Urban et al., 1988).

Table 3

Structure	Permeability			Porosity	Water Content	Young's Modulus	Aggregate Modulus	Density	Nonlinear Stiffness Coefficient
	k_x	k_y	k_z						
	$m^4 N^{-1} s^{-1}$	$m^4 N^{-1} s^{-1}$	$m^4 N^{-1} s^{-1}$	ϕ	WC	E	H_A	ρ	β
					%	MPa	MPa		
Outer Annulus	1.68E-15	1.68E-15	1.64E-15	0.73	65	4.2	0.44	1060	2.7
Inner Annulus	1.87E-15	1.87E-15	1.56E-15	0.78	72	2.56	0.44	1060	2.7
Nucleus	2.13E-15	2.13E-15	1.45E-15	0.83	82	1.56	0.27	1000	2.7