Biomechanical Response of a Lumbar Motion Segment Under Repetitive Loading Conditions – A Finite Element Study

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Introduction: Low back pain is one of the most prevalent and costly work related injuries in the United States. Although epidemiological studies have suggested possible causes, the actual mechanisms by which the lumbar spine is injured during repeated load cycles resulting in low back pain, remains unknown. It is the unique interaction between the solid and fluid components that provide the disc the strength and flexibility required to bear cyclic loading of the lumbar spine. However, the response of the disc to cyclic loading conditions that occur during repetitive lifting is difficult to measure in vivo and in vitro and has not been investigated using any model as well. Therefore, the purpose of this study was to construct a numerical model that predicts the loading and unloading behavior of a lumbar motion segment including the poro-elastic behavior of the disc and the effect of the change in proteoglycan content in the disc with respect to applied cyclic load. This model was used to predict the change in disc height due to repetitive lifting over a short period of time. The results were compared with the gross measurement of height change measured in vivo (1).

Materials and Methods: The theory of poroelasticity was applied by modifying a three-dimensional finite element model of the L3-L4 motion segment, which was validated for static loading conditions (2) to include poroelastic components. This model included the vertebral bodies, disc, posterior elements, seven spinal ligaments and facet contacts. Permeability values for the nucleus, annulus, endplate, cancellous bone and cortical bone were taken from the literature (3). The porosity of the nucleus and endplates was defined as 0.8 and of the annulus as 0.7. A much smaller value was assumed for cancellous (0.27) and cortical bone (0.02). The bulk modulus for nucleus was assumed to be 2500 MPa. The drained elastic modulus and Poisson's ratios for all the disc components were also taken from the literature (3).

The inclusion of poroelastic components in the finite element model introduced the fluid within the intervertebral disc. The initial pressure within the disc was determined knowing the applied load and using an equation used by Broberg (4); this initial pressure was applied as a pore pressure at every node defined within the disc. The effect of the change in the concentration of proteoglycans contained within the nucleus was modeled by incorporating a pressure, in this case referred to as the swelling pressure, which is dependent on the fixed charge density.

Knowing the disc's initial volume and the magnitude of the external load, as well as assuming the water content of the disc to be 70% (5), the initial fixed charge density of the disc was defined as (4):

$$f_o = \frac{m}{WC * V_o} \qquad (1)$$

Where f_o *is* the initial fixed charge density, m is a reference volume (4), WC is the water content of the disc and V_o is the initial volume of the disc as defined by the geometry of the disc in the finite element model. Once the initial conditions were defined, the fixed charge density was then defined as a function of the time over which the load is applied (4). This was done by first discretizing the time over which the load was applied where each step could be defined by the initial time t_i and the final time t_{i+1} where the load was held constant during this interval. Second, knowing the fixed charge density at time t_i, the fixed charge density at time t_{i+1} was solved using equation (2). This was done through trial and error until the solution to the integral equaled the right hand side of the equation (2).

$$\frac{m}{p}\int_{1/f_{i+1}}^{1/f_i} \frac{z\left(z^2+a\right)dz}{\left[1+\left(f_i z\right)^{4/3}\right]\left(z^3-qz^2+a\,z-q\right)} = -\frac{k_o t}{2} \qquad (2)$$

Where f_i and f_{i+1} are the fixed charge densities at time t_i and t_{i+1} respectively, p is the initial pressure of the disc, a is a constant, q is defined by a constant P (0.66 MPa) divided by the nucleus, k_0 is a term from Darcy's law which in this case was assumed to be 11×10^{-16} m^5/Nsec and t is the length of time over which the load is applied equal to, $t_{i+1} - t_i$. Once the distribution of the fixed charge density has been determined over the period of time considered in the analysis, it was then used to calculate the swelling pressure over that time period using the equation (3):

$$p_{swelli+1} = Pf_{i+1}\frac{f_{i+1}^2+1}{a\,f_{i+1}^2+1} \qquad (3)$$

Where P and α are constants (4). The fixed charge density and corresponding swelling pressure were calculated at each time point defined in the analysis. The value for p_{swell} at each time t was included in the finite element analysis as an input. Using this iterative process and including the swelling pressure as an input into the analysis, enabled the model to be used predict the response of the motion segment under cyclic loading conditions.

The model was validated by comparing the gross measurements of total stature measured in vivo by Tyrrell et al., in which a normal subject lifted a 40 kg barbell to knuckle height at a frequency of 12 lifts/min for 20 minutes followed by a 10 minute standing recovery period. This was modeled in the finite element model by applying a compressive preload of 400 N to the superior surface of the L3 vertebral body plus a 400 N peak-to-peak compressive load and 5 Nm flexion moment at a frequency 0.2 Hz for 20 minutes. The load was then reduced to 400 N for 10 minutes of recovery.

Results: The finite element model results were compared to the overall height change measurements reported by Tyrrell et al. (1). The gross change in height of the subject was calculated by multiplying the measured height loss in a single motion segment by fifteen in the finite element study. The finite element model predicted a 0.85% loss of total stature over the 20 minutes of loading as compared to 0.86% reported by Tyrrell et al. The recovery of disc height predicted by the finite element model closely mimics the in vivo results with a significant increase in disc height initially and then becoming more gradual as time progressed (Figure).

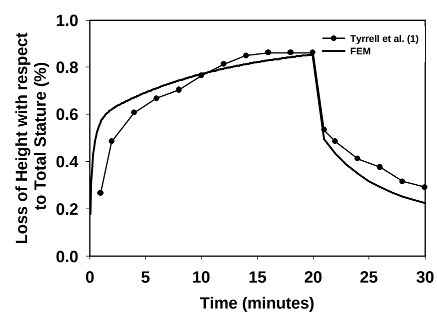

Figure Percent loss of height as compared to total stature when cyclically loaded over a short period of time as predicted using the finite element model and compared to in vivo results (1).

Discussion: A numerical model has been constructed that predicts the loading and unloading behavior of a lumbar motion segment over a short period of repetitive lifting similar to those observed in vivo. The results have been validated using gross measurement of height change in vivo (1). Although a single motion segment was modeled, the change in total stature was calculated by assuming that each disc in the spine would experience the same reduction in height.

This numerical model can now be used to study the response of the disc to various cyclic loading conditions that occur during repetitive lifting.

References:

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