COMPARISON OF THE BIOMECHANICAL RESPONSE BETWEEN NON-LINEAR EL.ASTIC AND POROELASTIC FINITE ELEMENT MODELS OF A HEALTHY C5-C6 MOTION SEGMENT

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Introduction: Neck pain is the most common manifestation of spondylotic radioculopathy or myelopathy which commences with cervical disc degeneration. Reduction in water content of the disc has been shown to be an initial sign of degeneration followed by a complete desiccation in the severely degenerated discs. Although a number of *in vivo* and *in vitro* biomechanical studies have attempted to predict the normal biomechanics of the healthy cervical spine, but these studies are incapable of quantifying the relative contribution on the biomechanical response from individual spinal components. Numerous mathematical and finite element techniques have attempted to fill this void, but to the best of author's knowledge, none of the above studies have included the fluid flow into and out of the disc. The objective of this study is to develop and validate a poroelastic finite element model of C5-C6 motion segment under diurnal compressive load of 350 N for 24 hours. Comparison of the biomechanical response between the non-linear elastic model and the poroelastic mode was studied for flexion, extension, axial rotation, and lateral bending motions of 1.00 Nm with a compressive preload of 73.6 N.

Methods: A three dimensional non-linear elastic finite element model of an intact C5-C6 cervical spine motion segment was developed using CT scans of 38-year-old female normal subject that included cortical bone, cancellous bone, posterior elements, annulus fibrous, nucleus pulposus, and facet cartilage. Additionally, anterior longitudinal ligament, posterior longitudinal ligament, intraspinous fiber, ligamentum flavum fiber, and capsular ligament were also included. Geometries of all spinal components agreed well within the standard deviation of literature. A poroelastic model was developed from non-linear elastic model by adding the effects of disc pressure due to change in the proteoglycan concentration with load and strain-dependent permeability. An initial pore pressure of 0.041 MPa was assumed to act at all nodes of the disc in the poroelastic model (1). The effect of pressure due to the proteoglycan concentration p_i^{swell} is given by (1).

$$p_i^{swell} = Pf_i \frac{f_i^2+1}{\alpha f_i^2+1}$$

where f is the proteoglycan concentration, P=0.66 MPa, α=0.15
The effect of strain-dependent permeability, p_i^{strain} is given by (2,3,4,5),

$$-p_i^{strain} = \frac{E\varepsilon_i - \frac{1}{2}H_A \frac{\frac{2}{M}\ln\left[\left(\frac{k_i}{k_0}\right)\left(\frac{e_0}{e_i}\right)\right]}{\left[1+\frac{2}{M}\ln\left[\left(\frac{k_i}{k_0}\right)\left(\frac{e_0}{e_i}\right)\right]\right]^{\beta+\frac{1}{2}}} \exp\left\{\beta\frac{2}{M}\ln\left[\left(\frac{k_i}{k_0}\right)\left(\frac{e_0}{e_i}\right)\right]\right\}}{1-\left[\varepsilon_i+\phi_0^f\left(1-\varepsilon_i\right)\right]}$$

where E is the elastic modulus, ε is the axial strain, H_A is the aggregate modulus, M is the strain-dependent permeability coefficient, k is the permeability, e is the void ratio, β is the nonlinear stiffening coefficient, φ^f is the porosity. Subscript *0* corresponds to initial condition, while subscript *i* corresponds to the instantaneous condition. The effects of both pressures were incorporated into the model by applying equivalent loads at the nodes on the superior & inferior interface of the endplates and disc. Due to the paucity of poroelastic material data in cervical spine, most of these material properties were taken from the lumbar spine region. To validate the poro-elastic model a diurnal compressive load of 350 N for 16.5 hours represented day time activities and 40 N load for 7.5 hours to model night time rest period was applied on the superior surface of C5 keeping the inferior surface of C6 fixed (6). The disc height loss during 24 hours of compressive loading was compared with the *in vivo* study to validate the poroelastic model (7). Comparison of the biomechanical response between non-linear elastic model and poro-elastic model was done for the following moment loadings: flexion, extension, axial rotation, and lateral bending moments of 1.00 Nm for 5 seconds. A physiologic compressive preload of 73.6 N was also included in the analyses. Rotation of C5 with respect to C6 in the sagittal plane was also computed for the non-linear elastic model and compared with the corresponding results from the poroelastic model. The biomechanical responses of the individual components such as disc height loss/gain, annular pressure, nucleus pressure, endplate pressure and facet load were also compared between the two models.

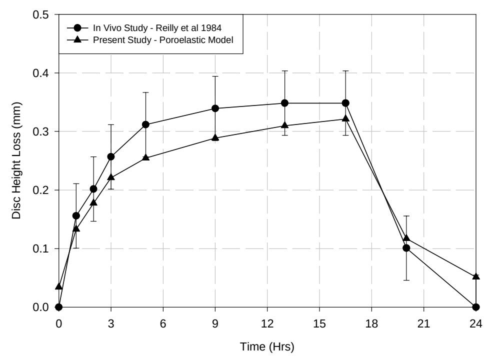

Figure 1: Validation of poroelastic model with *in vivo* study

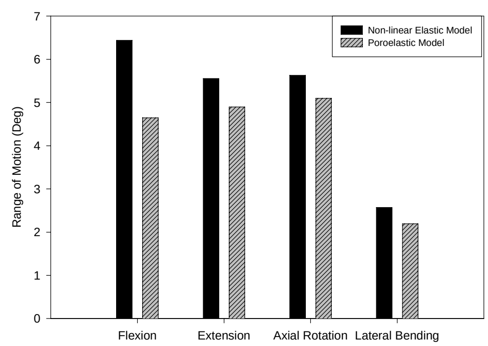

Figure 2: Comparison of range of motion for two models

Results: The disc height loss for the poroelastic model agreed well with the published *in vivo* study (Figure 1). Further, the range of motion of the poroelastic model was lower than the non-linear elastic model by 28% in flexion, 12% in extension, 9% in axial rotation, and 15% in lateral bending (Figure 2). Presence of a higher facet load in the non-linear model was of the order of 30% to 40% as compared to the facet loads predicted by the poro-elastic model.

Discussion: The results of the present study show that a poroelastic model is a better representation of physiologic compressive and moment loading. An increase in pressure in the individual components of poroelastic model makes it stiffer and hence, capable of withstanding higher physiologic loads. In conclusion, the poroelastic model was less flexible than the non-linear elastic model due to the incorporation of physiologic disc pressure.

References: (1) Broberg K. J Biomech 1993;26:501-512. (2) Argoubi M et al. J Biomech 1996;29:1331-1339. (3) Holmes MH et al. J Biomech 1990;23:1145-1156. (4) Iatridis JC et al. J Biomech 1998;31:535-544. (5) Riches PE et al. J Biomech 2002;35:1263-1271. (6) Moroney SP et al. J Orthop Res 1988;6:713-720. (7) Reilly T etal. Chronobiology Int 1984;1:121-126.