INTRODUCTION

Structural alterations of the disc are accompanied by changes in disc tissue structure and composition during degeneration processes. Fissures and cracks appear in the disc sometimes extending from the outer annulus to the nucleus. Human cadaver studies have shown that the probability of presence of annular tear is 0.6 in the age group 30 to 34 which increases to 0.85 in the 60 year age group. Among different phenomena that occur during disc degeneration, annular radial tears and concentric tears are commonly associated with disc herniations. Radial tears are most often present in the posterior annulus and located in the inner 1/3rd of the upper region of the annulus occupying 2.5% of the total volume of the annulus. Concentric tears are most often located in the right anterior quadrant of the annulus in the upper outer annulus region and have been found to occupy 1.7% of the total volume of the annulus. Structural alterations of the disc due to annular tears change the flexibility of the disc and thus contribute to the progression of these failures. Change in biomechanics of the disc occur not only due to increase in size of existing cracks and fissures but also depend on the integrity of the surrounding annular fibers. Understanding how pre-existing cracks and fissures affect the biomechanics of the motion segment is very difficult to study using either in vivo or in vitro models. Finite element model technique is ideally suited for this type of problem. It is hypothesized that (1) the biomechanical response of a lumbar disc with annular tears behave closer to the normal disc if the entire fiber length near the tear location can carry part of its normal load carrying capacity and (2) in whatever way the fibers near the tear location share the load, largest increase in motion was predicted under torsion.

METHODS

A previously validated three-dimensional poro-elastic finite element model of the L4-L5 motion segment that includes biological parameters (swelling pressure and strain dependent permeability and porosity) was used. Progression of annular tears was modeled in the following way: Each finite element representing annulus is characterized by eight “integrating” points distributed in a symmetric fashion within the element. Material property associated with these eight points determines the overall stiffness of the element. Using the “user-defined-material” routine available in the FE package (ADINA), material property associated with these “integrating” points in each element can be modified. Creation of annular tears from inside surface to the outer annulus surface was modeled by progressively changing material properties at integrating points in elements along the assumed direction of tear progression from annular ground substance material to nucleus material property. Progression of annular tears was assumed to occur in elements located in the posterior left lateral quadrant of the disc for radial tear and anterior left lateral quadrant for concentric tear. Finite element models representing radial tears occupying 0.0% to 0.32%, of the total annulus volume were created in the left lateral quadrants of the annulus. Finite element models representing concentric tears occupying 0.0% to 0.37%, of the total annulus volume were created in the left lateral anterior quadrants of the annulus. Three different ways how the fibers around the tear location sustain load was studied (figure 1): (1) Entire length of fibers around the tear location cannot sustain load (MODEL: EL0), (2) Entire length of fibers around the tear location can withstand 50% of its normal load (MODEL: EL50%) and (3) Partial length of fibers around the tear location alone cannot sustain load (MODEL: PL0). The models were pre-loaded with a 1200 N compression followed by moments of 7.5 Nm in all the three principal planes in both directions. The principal motions obtained from these models were compared with the corresponding results from an intact motion segment to determine the effect of increasing tear volume on the flexibility of the lumbar disc.

RESULTS

If the entire length of fibers around the tear location (Model EL 0) was not able to carry any load, an exponential increase in disc biomechanics was observed as radial tear volume increased in all three loading modes. Concentric tear produced such an exponential increase under lateral bending mode only. This model also predicted that under torsion loading mode (Figure 2) a fivefold increase in motion (0.3% to 1.5%) occurred due to radial tear while concentric tear produced an increase of less than twofold (0.3% to 0.6%) as compared to the other two models.

If the entire length of fibers around the tear location was able to carry load partially (Model EL50%) a linear biomechanical change was observed as the tear volume increased (except concentric tear under lateral bending mode). In this model both radial and concentric tears produced least increase in motion (0.2% to 0.3%) as compared to the other two models under all loading modes.

If the fiber length around the tear location alone was not able to carry load (Model PL0) a linear biomechanical increase was observed as tear volume increased (except under lateral bending with concentric tear). In all loading modes, increase in motion was nearly same as those observed when the fibers around the tear location were able to carry load partially.

Comparing the effect of loading modes on the increase in flexibility of the disc, torsion (Figure 2) produced largest increase in motion due to both radial (1.6% as compared to 0.5% under flexion/extension and 0.8% under lateral bending) and concentric tears (0.6% as compared to 0.25% under both flexion/extension and lateral bending).

CONCLUSIONS

Largest increase in motion as tear volume increased was calculated when the entire length of fibers near the tear location was not able to carry any load under all the three loading modes. Least increase in motion as tear volume increased was seen when the entire fiber length near the annular tear carries part of its normal load capacity. The analyses thus showed that integrity of annular fibers near the tear location is important and affects the change in flexibility of the disc due to annular tears. The biomechanical changes in a lumbar disc due to annular tears combined with in-effectiveness of the surrounding annular fibers were largest under torsion.

Figure 1. Representation of load sharing of the annular fibers

Figure 2: Percent increase in motion as annular tear volume increases under torsion loading mode.

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