An Investigation of the Role of Dynamic Axial Torque on the Disc Herniation Mechanism

By

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Leigh Marshall

ABSTRACT

Background: Disc herniations are common and have been demonstrated as one potential source of low back pain. To date epidemiological studies have found associations between lifting, lifting and twisting and twisting with increased risk in the development of disc herniations (Greenough and Fraser, 1994, Kelsey et al., 1984, Mundt et al., 1993). Subsequent, *in vitro* investigations were able to produce disc herniations through repeatitive flexion extension motions on cervical porcine functional spinal units (Callaghan and McGill, 2001). However, *in vitro* investigations on axial torque have drawn mixed conclusions and controversy remains on the role it plays with respect to disc herniations (Farfan et al., 1970, Adams et al., 1981). Therefore, the work in this thesis was to investigate the role of dynamic axial torque on the disc herniation mechanism.

Methods: Porcine cervical spines were used as they are a good approximation to the human lumbar spine (Yingling et al., 1999). The study design involved repetitive flexion extension motions of the spinal units either preceded or followed by dynamic axial torque. During axial torque the spinal units were loaded to 17.5 Nm (standard deviation = 0.5 Nm) of dynamic axial torque for either 2000 or 4000 testing cycles. These spinal units were compared to spinal units that were loaded in repetitive flexion extension motions only and axial torque only. The spinal units were tested in a servohydraulic dynamic testing machine, combined with a custom jigs which allowed loading in flexion/extension, axial torque and compression. Plane film radiographs with contrast in the nucleus were obtained at regular intervals during and following the mechanical testing. Final dissection determined the disc injury patterns.

Results and Discussion: Examination of the sectioned intervertebral discs indicated axial torque in combination with repetitive flexion extension motions, regardless of order, encouraged radial delamination. While, repetitive flexion extension motion alone encouraged posterior or posterolateral herniation patterns. Axial torque alone was unable to initiate a disc herniation. There was an increase in both rotation and stiffness of the intervertebral disc in response to repeated axial torque. There were no differences in rotation and stiffness between the groups. Both x-ray images and computed tomography scans were equally as good at identifying posterior or posterolateral herniations but were not good at detecting radial delamination.

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TABLE OF CONTENTS

LIST OF TABLES

LIST OF ILLUSTRATIONS

1 – INTRODUCTION

Low back pain is a prevalent cause of pain and disability in the adult population. There are a variety of ailments from muscular to neurological disorders which are manifested as low back pain (Hicks et al., 2002). Disc herniations have been demonstrated to be one potential source of this pain (Deyo et al., 1990). Two percent of patients with low back pain show evidence of a herniated disc (Frymoyer, 1988). Epidemiological studies have shown increased risk in the development of both low back pain and herniated discs with physically heavy work, static work postures, frequent bending and twisting, lifting, repetitive work, vibrations and asymmetrical postural demands (Greenough and Fraser, 1994, Marras et al., 1993, Mundt et al., 1993, Wilder et al, 1988, Kelsey et al., 1984, Andersson, 1981). *In vitro* investigations were able to produce intervertebral disc herniations through mimicking lifting motions by way of repeated, full range flexion extension motion on cervical porcine functional spinal units (Callaghan and McGill, 2001). However, *in vitro* investigations on twisting have drawn mixed conclusions on whether axial torque and twist plays a role in the development of disc herniations. On the one side Adams et al., (1981) hypothesized axial twist within physiological ranges of rotation was unimportant in the development of disc herniations. On the other side Farfan et al. (1970) hypothesized axial torque was important in the development of disc herniations and with repeated cycles of twist there would be a gradual communication between the nucleus pulposus and the outside of the disc. Therefore, the mechanism by which axial torque effects the development of disc herniations remains inconclusive and in this way motivated this thesis.

1.1 – Purpose

The purpose of this thesis was to evaluate the role of dynamic axial torque on disc herniation mechanisms of functional spinal units. The role of dynamic axial torque on injury mechanisms was investigated by comparing injury patterns observed in the following groups: flexion extension only; axial torque only; axial torque followed by flexion extension; and flexion extension followed by axial torque. Axial torque was controlled and applied to the functional spinal units while the resultant rotation was measured.

1.2 – Hypotheses

The following hypotheses were made with respect to the injury patterns of the functional spinal units:

1 - Functional spinal units loaded in axial torque only would have circumferential tears in the periphery of the annulus; however, the nucleus pulposus would remain unaffected.

2 – Functional spinal units loaded in repetitive flexion extension motions followed by axial torque would have a posterior herniation and radial delamination resulting in damage to the entire disc.

3 – Functional spinal units loaded in axial torque followed by repetitive flexion extension motions would result in an earlier onset of disc herniation and more serious tissue destruction.

2 – LITERATURE REVIEW

2.1 - Overview

Low back pain is a prevalent cause of pain and disability in the adult population. It has been estimated that 80% of adults will experience an episode of low back pain in their lifetime of which approximately 4-5% will experience acute episodes yearly (Marras, 2000, Plante et al., 1997, Deyo, et al., 1990). There are a wide variety of ailments ranging from muscular to neurological disorders which can be manifested as low back pain (Hicks et al., 2002). Herniated disc have been demonstrated to be one potential source of this pain (Deyo et al., 1990). Epidemiological studies have shown increased risk in the development of both low back pain and herniated discs with physically heavy work, static work postures, frequent bending and twisting, lifting, repetitive work, vibrations and asymmetrical postural demands (Greenough and Fraser, 1994, Marras et al., 1993, Mundt et al., 1993, Wilder et al, 1988, Kelsey et al., 1984, Andersson, 1981). *In vitro* studies have created herniated discs through combined hyper flexion and compression, flexion and cyclic compression as well as repetitive full range flexion-extension and compressive (Adams and Hutton, 1982, Adams and Hutton, 1985, Callaghan and McGill, 2001).

Epidemiological studies have found associations between twisting and the development of both low back pain and herniated discs (Mundt et al., 1993, Andersson, 1981). Unfortunately, it is often difficult to determine in epidemiological studies if it is the twisting motion that is detrimental to the intervertebral disc complex because many movement patterns involving twisting also result in coupled motions (Andersson, 1981). Further problems associated with some epidemiological studies involve the inability to distinguish which is more detrimental to the disc, the kinematic variable of twisting or the kinetic variable of axial torque. Large axial torques otherwise known as "twisting torques" can be generated *in vivo* without twisting of the intervertebral disc. These "twisting torques" are accomplished through the co-activation of the musculature surrounding the spine. Since the spinal musculature is not designed to create pure twisting torques, large increases in compression of the spine is also generated with axial torque (McGill, 2002). This pattern of muscular activation may also be responsible for the increased risk of low back pain and disc herniations with twisting.

The role of twisting and axial torque in the development of herniated discs has also been investigated *in vitro*; however, mixed conclusions have been drawn by those investigators (Adams and Hutton, 1981, Farfan et al., 1970). On the one side Adams et al., (1981) maintains axial rotation of the disc can play no role in the derangement of the disc within physiological ranges of twisting, while on the other side, Farfan et al., (1970) maintains repeated twisting compromises the disc within physiological ranges of motion and is important in the development of herniated discs. Therefore, this *in vitro* study will examine the role of twisting and axial torque on disc herniation mechanisms.

2.2 - Mechanisms of Disc Herniations

A herniated disc is defined as the "localized displacement of disc material beyond the limits of the intervertebral disc space" and the pathological definition is "identification of disc material forced out of normal position through an annular defect" (Fardon and Milette, 2001). A herniated disc is initiated by an increase in the hydrostatic pressure of the nucleus pulposus against the inner wall of the posterior annulus fibrosus. Tampier (2006) characterized through dissection and discogram herniated discs created in porcine cervical spines by combined loading in compression and repetitive flexion extension motion. He hypothesized that an increase in hydrostatic pressure from the nucleus pulposus produces a cleft in the inner posterior wall of the annulus through which the nucleus is pumped, creating a pocket between collagen fibers with delamination occurring after the nucleus passed through the layer. As the hydrostatic pressure increases and with repeated trauma to the intervertebral disc a new cleft is formed and this process is repeated as the herniation continues through the inner most annular layers to the outer annular layers of the intervertebral disc.

Eighty-five to ninety percent of herniated discs occur in the posterior region of the intervertebral disc (Haughton, 1988). Tsuji et al., (1993) compared the laminate structure of the anterior and posterior annulus of 24 cadaver intervertebral discs of the L3/L4 and L4/L5. The posterior region of the disc was found to have greater number of incomplete and discontinuous lamellar bundles, larger fiber-interlacing angles and greater incidences of bundle interruption which extended in the posterior direction of the intervertebral disc. Tsuji et al., (1993) speculated these factors predispose the posterior region of the intervertebral disc to herniation.

Herniated discs also tend to occur in the lower lumbar spine with more than 95% of lumbar herniated discs occurring at the L4/L5 and L5/S1 levels (Deyo et al., 1990). Individuals between the ages of 35 and 50 have the highest frequency of herniated discs (Marras, 2000). Differences in the water content and proteoglycan concentration in the nucleus of young compared to older spines may explain this increased frequency. Younger spines have a higher water content and proteoglycan concentration in the nucleus compared to older spines (Urban and McMullin, 1988). As the spine ages the water and proteoglycan content of the nucleus pulposus decreases which results in a decreased volume and hydrostatic pressure in the nucleus, as well as, decreased hydraulic behavior of the spinal segment. The decreased pressure in the nucleus pulposus and less viscous nature makes the intervertebral discs of older spines less susceptible to herniation (McGill, 2002, Adams and Dolan, 1995).

Many researchers have attempted to recreate the conditions necessary to cause an intervertebral disc to herniate in an *in vitro* setting. Adams and Hutton (1982) created herniated discs in 43% of human lumbar motion segments tested under compression and hyperflexion (flexion beyond the normal limit of flexion). The remainder of the motion segments in this study failed by compression fractures or hyperflexion fractures. Subsequent work by these same authors found six of 29 human lumber motion segments failed by a herniated disc when loaded in cyclic compression as well as flexion (Adams and Hutton, 1985). The remainder of these motion segments failed by endplate fracture and vertebral crumbles. Since herniated discs were not reliably re-created in the above noted motion patterns more work had to be done to determine what motion patterns increase the susceptibility of the intervertebral disc to herniation.

Further research using a porcine cervical model has shown disc herniations can be produced through a combination of compression and highly repetitive full range flexion-extension motion (Callaghan and McGill, 2001). Of the two spinal segments tested under 260 N of compression in position control one of the discs did not herniate and one disc showed a track initiation after an average of 83700 cycles. Of the three spinal segments tested under 1472 N of compression in position control three discs had herniated and one had failed by endplate fracture after an average of 34974 cycles. Therefore, the authors concluded the higher the compressive loading the more severe and frequent the disc herniations and the nucleus pulposus would track in the posterior and posterolateral direction. Subsequent *in vitro* work also using a porcine model had found the tracking of the nucleus pulposus can be biased toward the posterolateral of the disc by changing the directionality of the bending moment (Aultman et al., 2005). When the bending moment of the motion segments was changed from the axis running along the anterior posterior of the vertebra to an axis running anterior posterior minus 30 degrees to the left the tracking of the nucleus pulposus was to the posterolateral of the right side of the intervertebral disc in 15 of the 16 trials.

In vitro studies to date have focused on the two-axis loading of compression and flexion-extension in the etiology of disc herniations (Aultman et al., 2005, Callaghan and McGill, 2001, Adams and Hutton, 1985, Adams and Hutton, 1982). However, other *in vitro* studies research has found with forward flexion there is an increase in the available axial rotation of lumbar spinal motions segments compared to when those same segments are rotated in a neutral posture (Pearcy and Hindle, 1991). The increase in available rotation found by these authors lead them to speculate three-axis loading of compression, flexion-extension, and torsion may increase the vulnerability of the intervertebral disc to injury. Epidemiological studies have also found frequent bending and twisting is associated with increased risk of low back pain and absence from work (Marras et al., 1993, Andersson, 1981). Therefore, further studies on the effects of axial torque and twisting on the intervertebral disc are important and warrant further investigation to understand its effects of the intervertebral joint complex.

2.3 - Axial Torque and Twisting

In vitro studies on human lumbar spinal units have found that 90% of the torque strength of the spinal segment is provided by the intervertebral disc and the facet joints (Farfan, 1984, Farfan et al., 1970). The intervertebral disc consists of the nucleus pulposus which is surrounded by the annulus fibrosus. The inner annular fibers attach to the endplates and contain the nucleus pulposus while the outer annular fibers attach to the vertebral bodies. The annular fibers consist of a series of lamellae which are tilted in alternative directions between successive layers, thereby, making only half of the annular fibers able to provide reinforcement against rotation in any one direction (Hukins and Meakin, 2000, Hickey and Hukins, 1979). Furthermore, it is the outer annulus which provides most of the intervertebral discs resistance to bending and torsion and the facet joints which provide resistance against excessive shear and axial rotation (Adams and Dolan, 1995).

The articular facet joints are oriented to protect against excessive torsion of the spinal segments (Farfan, 1969). Facet joints which have a more oblique orientation are less effective at resisting rotation (Ko and Park, 1997, Duncan and Ahmed, 1991, Farfan, 1969). Duncan and Ahmed (1991) investigated the effect of facet asymmetry on the human lumbar motion segments response to axial torque by comparing the rotational response of the left and right facet joints to an applied load. The authors found increased axial rotation at the L4/L5 compression facet joint that were both more obliquely oriented and flat. Haughton et al., (2002) also found significant increases in rotation towards the sacrum compared to the upper lumbar in response to passive rotation of the lumbar spine of subjects with no back pain. On average the authors found 0.2° of rotation at the L1/L2 joint compared to 1.6° of rotation at the L5/S1 joint. The lower lumbar spine facet joints also have a more oblique facet orientation compared to the higher lumbar spine facet joint and may explain the increased incidence of disc degeneration at the L4/L5 and L5/S1 levels of the spinal column (Osti et al., 1992, Duncan and Ahmed, 1991, Butler et al., 1990, Farfan, 1969).

Oxland et al., (1991) tested porcine cervical spinal segments consisting of three vertebra pre- and post- injury to correlate injury patterns with biomechanical instabilities. The authors applied pure moments in flexion, extension, right lateral bend, left lateral bend, right axial rotation and left axial rotation in four increments to 3.6 Nm and determined the resultant neutral zone and range of motion of these motion segments. Multivariate analysis was used to correlate instabilities with the observed injuries and found axial rotation was correlated with injuries to the anterior disc, endplate and capsular ligaments.

Further studies on axial torque and twisting of functional spinal units have found rotating spinal segments to failure, as defined as a drop in torque with successive rotation, results in fractures at the base of the articular processes and circumferential rents to the peripheral annular fibers (Liu et al., 1985, Adams and Hutton, 1981, Farfan et al., 1970). These circumferential tears are found in the outer annular layers with the inner annular layers remaining unaffected (Farfan, 1969). However, disagreements remain as to whether the intervertebral disc can become damaged prior to damage to the articular processes. Adams and Hutton (1981) maintain damage to the intervertebral disc from torsional stress must be preceded by damage to the articular facet joints; meanwhile, Farfan et al. (1970) maintains that the intervertebral disc can become damaged in physiological ranges of twisting. Hence, the debate remains as to whether axial torque and twisting is important in the etiology of discogenic degeneration.

2.4 - Axial Torque and Twist and Disc Degeneration

Adams and Hutton (1981) maintain that axial rotation cannot play any role in the derangement of the intervertebral disc within physiological ranges of motion. In this study the structures of the human lumbar functional spinal units were sequentially damaged by cutting the supra/interspinous ligament, the compression facet and the tension facet and tested the spinal units under combined compression and torsion. The resultant stiffness curves were analyzed and indicated the compression facet was the structure which provided the greatest resistance to torsion. The load required to cause the facet joint to fail was approximately one third to one tenth of the load required to damage the intervertebral disc. The limit of rotation before fracture was reached at approximately 1-3° of rotation with 10-30 Nm of axial torque.

Since more rotation would be required to damage the intervertebral disc than to damage the facet joints the researchers hypothesized that axial rotation within physiological ranges of motion could not play a role in the derangement of the intervertebral disc. Only when the articular facet joints are damaged as in cases of trauma and subsequent to failure can torsion cause circumferential tears in the annulus fibrosus. Given these findings it had been concluded that axial rotation does not play any role in the etiology of disc degeneration or disc prolapse (Adams and Hutton, 1981).

Callaghan and McGill (2001) demonstrated that herniated discs are a cumulative injury which can be created under modest compressive loads with highly repetitive flexion extension motions. Adams and Hutton (1981) drew their conclusions from testing functional spinal units which were sequentially damaged and loaded in combined compression and torsion to failure. This loading makes the aforementioned study conclusions more relevant to instances of acute trauma and not a cumulative injury model. With repetitive loading the failure tolerance of a tissue decreases which may result in tissue failure at sub maximal loads (McGill, 2002). This cumulative trauma model would explain how annular fibers may become damaged from axial torque and twisting at loads lower than would be expected to cause damage to the intervertebral facet joints.

While Adams and Hutton (1981) found the compression facet provided the greatest resistance to axial rotation, an *in vitro* study by Krismer et al., (1996) found the annular fibers restricted axial rotation more than the facet joints. This study tested twelve cadaver motion segments from T11 – L5 in combined compression and torsion. Six of the spinal segments were tested following dissection of the annular fibers running in one direction and than tested again after bilateral facetomy and six of the spinal segments were tested following bilateral facetomy and then following dissection of the annular fibers acting to restrict rotation in one direction. Application of 8.5 Nm of axial torque resulted in an increase of axial rotation by 2° after dissection of the annular fibers acting in the direction of rotation compared to the pre-injured state and resulted in an increase of 1.2° after bilateral facetomy compared to the pre-injured state. The authors concluded that the annular fibers restrict more axial rotation than the facet joints and are more likely to become damaged in torsion.

Degeneration of the intervertebral disc can be found without damage to the articular processes. An *in vivo* study to investigate the prevalence of intervertebral disc degeneration and facet joint osteoarthritis was completed on 68 males and females with low back pain between the ages of 15-76 (Butler et al., 1990). Magnetic resonance imaging and computed tomography were used to determine the incidence of disc degeneration and facet joint osteoarthritis respectively in this subject pool. Disc degeneration was defined by these authors as a lack of preserved disc space, non-smooth borders between the annulus and nucleus, evidence of a herniated disc and/or a non-clear signal of the intervertebral disc. The authors found of the 108 disc characterized as degenerated, zero showed evidence of facet joint osteoarthritis. However, of the 41 facet joints diagnosed with facet joint osteoarthritis only one did not show evidence of disc degeneration. The authors concluded that disc degeneration proceeds facet joint osteoarthritis. However, which movement patterns lead to the disc degeneration observed in this study is unknown.

Farfan et al., (1970) studied how human lumbar functional spinal units responded to axial torque and twisting. The authors also investigated the response of the spinal segments to repeated axial rotation. The intervertebral joints of the functional spinal units increased in stiffness in response to increased strain rates. The recorded torque also declined in response to repeated rotation. Dissection and examination of intervertebral disc revealed circumferential tears in the posterolateral outermost annular fibers to the side opposite of rotation, with the inner annular fibers remaining intact and no communication between the nucleus and the annulus. These authors postulated the intervertebral disc may become injured by torsion in the physiological ranges of rotation and as deeper and deeper annular fibers became more and more compromised a gradual interaction would develop between the nucleus and the outside of the intervertebral disc.

One *in vivo* study which tested the hypothesis that tears to the periphery of the annulus may lead to secondary degenerative changes in the intervertebral discs was completed on sheep lumbar vertebra (Osti et al., 1990). The left anterolateral annulus of the sheep lumbar vertebra was cut and the sheep were returned to a field station where they were sacrificed one to 18 months post-operation. The authors found upon dissection of the vertebra that the longer the time elapsed between the operation and the animal sacrifice the more severe the disc degeneration. Only one of 12 sheep sacrificed one to two months post-operation had evidence of inner annular failure compared to all sheep sacrificed 18 months post-operation showed displacement of the nucleus. The study concluded peripheral tears in the annulus may play a role in degeneration of the intervertebral joint complex.

Osti et al., (1992) classified 135 lumbar discs from 27 cadavers according to the type of annular tears and the state of the nucleus pulposus. The annular tears were classified as either rim tears, circumferential tears or radial tears. The authors noted rim tears were more common in the anterior annular fibers, circumferential tears were equally distributed in the anterior and posterior annular fibers and radial tears were more common in the posterior disc. Furthermore, disc degeneration was more common in the L5/S1 discs compared to the L1/2 discs. The authors also noted the lesions found in the periphery of the intervertebral disc occurred without damage to the rest of the disc. Histochemical analysis of these lesion indicated they were likely due to mechanical trauma. These authors' postulated lesions to the periphery of the annulus precede degeneration of the nucleus pulposus and suggested annular tears may influence and accelerate degeneration of the intervertebral disc. Many researchers have found annular tears may be produced from axial torque and twisting, thereby making axial torque and twisting a potential candidate for the mechanical trauma which caused the peripheral tears observed in this study (Liu et al., 1985, Farfan, 1973, Farfan et al., 1970, Farfan, 1969).

2.5 - Axial Torque and Twist and Disc Herniation

There have been a few epidemiological studies to date which have found an association between twisting and herniated disc in the lumbar spine. One such study looked into the role of lifting both small children and inanimate objects with the development of disc herniations (Mundt et al., 1993). This was an epidemiological study of 297 cases of confirmed and unconfirmed herniated discs whose cases were also matched with control subjects who did not have low back pain. The subjects were all given a questionnaire to complete which assessed the relative risk of developing a herniated disc. The authors found the following movement patterns were associated with an increased risk of developing a herniated disc: lifting from the floor with a bent back and straight legs; starting and ending a lift at waist level; and twisting while lifting. Greenough and Fraser (1994) also used a questionnaire to examine the relationship between diagnosis of back pain and injuries. The questionnaire was completed by 300 patients from one orthopedic surgery practice and found disc prolapse was associated with twisting 3.8 times more often than with falls. However, other studies have found no association between twisting alone and increased incidences of herniated discs (Kelsey et al., 1982).

Kelsey et al., (1984) completed an epidemiological investigation into the role of frequent twisting and lifting in the etiology of lumbar disc prolapse. These authors found an increased risk in developing a disc herniation was associated with lifting and carrying objects greater than 11.3 kg, twisting while lifting an object greater than 4.5 kg or lifting objects more have 25 times in a day. These authors did not find an association between twisting alone or twisting while holding an object and disc herniation. The inconsistent associations found between twisting and increased risk of herniated discs in epidemiological studies may be due to the accuracy of questionnaires which heavily depend on people's memories which may or may not be accurate. Furthermore, it may be difficult to distinguish these motion patterns in real world tasks. Andersson (1981) noted it is particularly difficult to distinguish between lifting and twisting in *in vivo* studies as they predominately occur as coupled motions in industrial tasks. Furthermore, Mundt et al., (1993) suggest weak associations found between lifting and twisting could be due to this motion being less accurately recalled in non-occupational settings.

To further complicate matters it is difficult to distinguish between the kinematic variable of twisting and the kinetic variable of generating torque in epidemiological studies. As mentioned previously, it is possible to generate large "twisting torques" which place large compressive loads on the spine without twisting the intervertebral disc. In addition, it may be the magnitude of torque generating the twisting motion that is detrimental to the intervertebral joint rather than the twisting motion itself. This information is not available in epidemiological studies that make it difficult to determine the effect of twisting on herniated disc mechanisms.

Marras et al., (1993) performed a cross sectional study on 403 industrial jobs which involved tasks that were repetitive and did not involve job rotation. The authors monitored the thoracolumbar spine in three dimensions with a lumbar motion monitor while employees performed their job tasks. The jobs were divided into groups of low and high risk of low back disorders based on injury reports and medical records. The authors found that the higher the maximum load moment, maximum lateral velocity, average twisting velocity, lifting frequency and maximum sagittal trunk angle during the lift one higher the odds of being placed in the high risk group. However, these factors could not discriminate group placement when looked at independently of the other factors.

An *in vitro* study using a porcine model was completed to examine the influence of static axial torque on failure mechanisms of the intervertebral joint complex (Drake et al., 2005). Comparisons were made between functional spinal units exposed to repetitive full range flexion extension motions to functional spinal units exposed to 5 Nm of static axial torque and repetitive full range flexion extension motions. At 3000 cycles of flexion extension, 71% of the specimens tested in the static axial torque group had herniated compared to 29% of the specimens in the no static axial torque group. The authors concluded the addition of static axial torque resulted in earlier onset of herniation, a higher incidence of facet fractures, higher energy dissipation and stiffness following 3000 cycles of flexion extension. Therefore, static axial torque appeared to influence the subsequent disc herniation process.

Gordon et al., (1991) loaded 14 cadaver motion segments from the lower thoracic region of the spine to the sacrum. The motions segments were loaded repetitively at 1.5 Hz in a combination of compression, flexion to approximately 7° and rotation to approximately 3° for an average of 6.9 hours. Loading was terminated when the reaction force recorded leveled off for 1 hour. Magnetic resonance images and radiographs were taken pre- and post-testing and were assessed blindly by a radiologist and spinal surgeons. The authors found ten of the discs had annular protrusion defined as annular tears in the posterolateral region of the disc and four discs showed nuclear extrusion. The authors concluded loading in a combination of compression, flexion and rotation over time leads to annular separation and subsequent disc prolapse.

There remains little research completed to date of the precise mechanism by which axial torque and twisting influences the disc herniation process. In addition, there is little research into the role of dynamic axial torque and twisting and its effect on the failure mechanisms of intervertebral joints and whether or not dynamic axial torque and twisting may influence disc herniation mechanisms in particular. This leads into the study of the influence of dynamic axial torque on disc herniation mechanisms.

3 – METHODOLOGY

3.1 – Common Equipment Set-Up and Specimen Preparation

Porcine cervical spines were used for this study as they were a good approximation to human lumbar spines and most closely resemble the lumbar spine of a human adolescent (Oxland et al., 1991). In addition, the porcine cervical spine is anatomically, functionally, geometrically, macroscopically and microscopically similar to the human lumbar spine (Yingling et al., 1999, Oxland et al., 1991, Tampier, 2006). The porcine spines were obtained from a common source (mean age 6 months, weight 80 kg) and were advantageous over a human cohort because they provided control over such things as: genetic makeup; diet; exercise history; and age that may affect spinal mechanics.

The spines were stored in double polyethylene bags at -20 °C and were thawed at room temperature for 15 hours prior to testing. The spines were sectioned into functional spinal units consisting of two consecutive vertebrae and the intervening intervertebral disc. The C5/C6 functional spinal units were used in this study. The functional spinal units were stripped of their surrounding musculature leaving the osteo-ligamentous structures intact. In addition, the anterior processes overlapping the intervertebral disc were trimmed so they would not limit rotation of the disc during testing. The intervertebral discs of the sectioned vertebral endplates were graded according to Galante's criteria (Galante, 1967). Only those specimens which met Galante's grade 1 criteria of a normal disc with the annulus free of ruptures and a shiny white gelatinous nucleus were used for further testing.

The intervertebral disc of the functional spinal units were injected with a radio-opaque dye mixture consisting of 0.1 mL blue dye (Coomassie brilliant blue Gmix: 0.25% dye, 2.5% MeOH and 97.25% distilled water), 0.2 mL water and 0.2 mL of barium sulphate in order to track the progression of the nucleus pulposus in radiographs. The blue dye allowed for visual inspection of the tracking of the nucleus post dissection. The dye mixture was of a sufficient consistency that it did not diffuse into the annular fibers unless a fissure was present in the annulus (Callaghan and McGill, 2001).

The functional spinal units were secured to steel cups by inserting a screw into the center of the intervertebral disc and 18 gauge galvanized wire was looped bilaterally around both the anterior processes and lamina of the vertebral bodies. The cups were filled with non-exothermic dental stone (Denstone®, Miles, South Bend, IN, USA) to further secure the functional spinal units to the cups. The superior and inferior vertebra also had two additional screws inserted into the anterior of the vertebra close to the anterior processes and were covered with non-exothermic dental stone to further reduce rotation between the functional spinal unit and the cups during testing. To maintain hydration during the testing protocol the functional spinal units were wrapped in a saline soaked plastic backed cloth and plastic wrap.

The functional spinal units were placed in a servohydraulic dynamic testing system (model 8511, Instron Canada, Burlington, Ontario, Canada). As previously described by Callaghan and McGill (2001) flexion and extension of the functional spinal units were achieved through an electrical brushless servomotor (model BNR3018D, Cleveland Machine Controls, Billerica, MA, USA) and planetary gear head (model 34PL0400 Applied Motion Products, Watsonville, CA, USA) which was controlled by custom software interfaced with an ISA bus motion controller (model DMC1701, Galil Motion Control, Mountain View, CA, USA). The angular position data was measured using an incremental optical encoder attached to the motor shaft (model LDA-048-1000, SUMTAK Corporations of America, Piscataway, NJ, USA) and the torque data was measured using a strain gauge torque transducer (model 01190-152, Sensor Developments, Lake Orion, MI, USA).

The servohydraulic dynamic testing system was also combined with a custom jig which had been modified to incorporate a pneumatically driven torsion capacity with load-rate control (figure 3.1). The functional spinal units were loaded with a first order system at an initial load rate of 24 Nm/s and final load rate of 6.2 Nm/s. The initial load rate was taken after the functional spinal unit reached 10 Nm of axial torque due to play between the functional spinal unit and steel cups at lower levels of axial torque. The pressure gauge controlled the magnitude of load that the functional spinal units were exposed to while the flow valve controlled the rate of load application. The pneumatically driven pistons were set up in parallel and acted in opposite directions to cause rotation of the functional spinal units. The force data was measured using two strain gauge force transducers (model MLP-300-C0, Transducer Techniques, Temecula, CA, USA). The torque produced by each of the pistons was calculated by multiplying the force measured by the piston by the perpendicular distance between each piston and the center of the steel cups. The overall torque was calculated by adding the torque produced by each piston individually. Linear displacement of the piston during axial torque was measured using a linear potentiometer (Novotechnik, Southborough, MA, USA). Angular rotation was calculated from the linear displacement data using the following formula:

Angular Rotation = Linear Displacement / 0.15 m (figure 3.2).

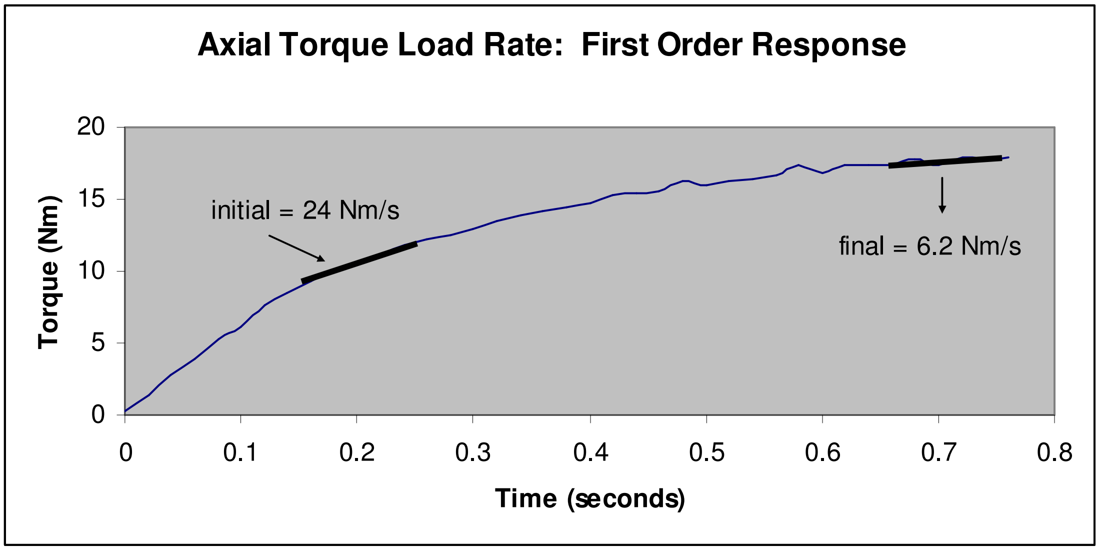


Figure 3.1 The axial torque was loaded by a first order system. The initial load rate was 24 Nm/s and the final load rate was 6.2 Nm/s.

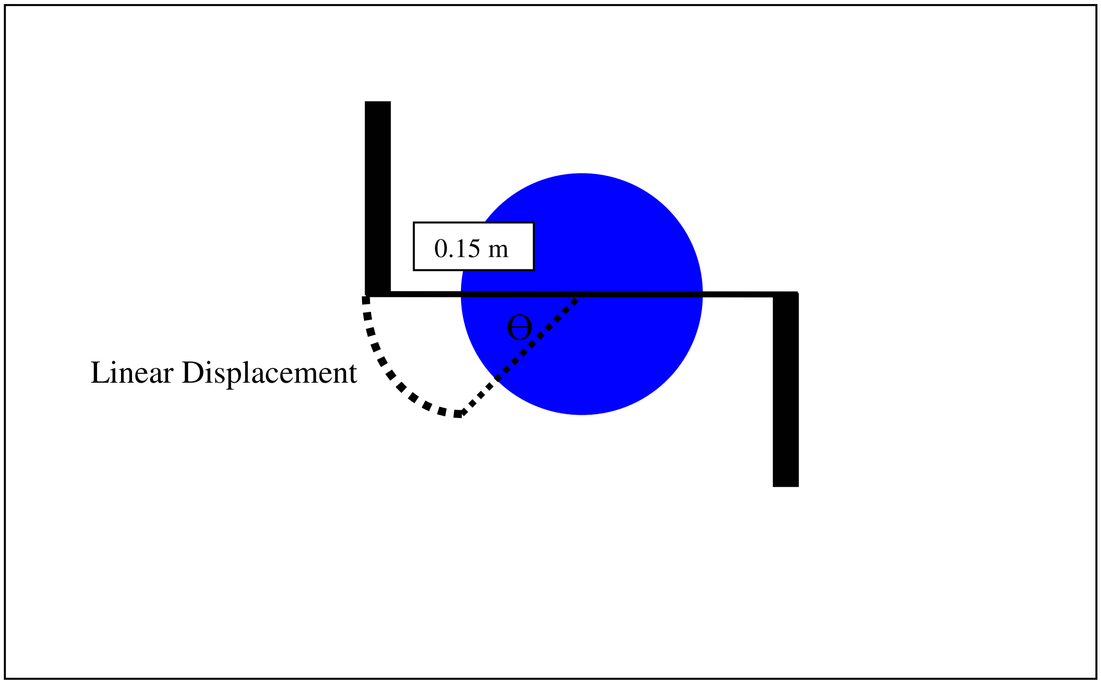


Figure 3.2 This was a diagram of how angular rotation was calculated from the displacement of the pistons. The solid lines represent the specimens in their neutral position and the dotted lines represent the resultant motion after axial torque was applied.

This set-up of the servohydraulic dynamic materials testing machine combined with the custom jigs allowed the test specimens to undergo loading in flexion/extension, compression, and axial torque (figures 3.3).

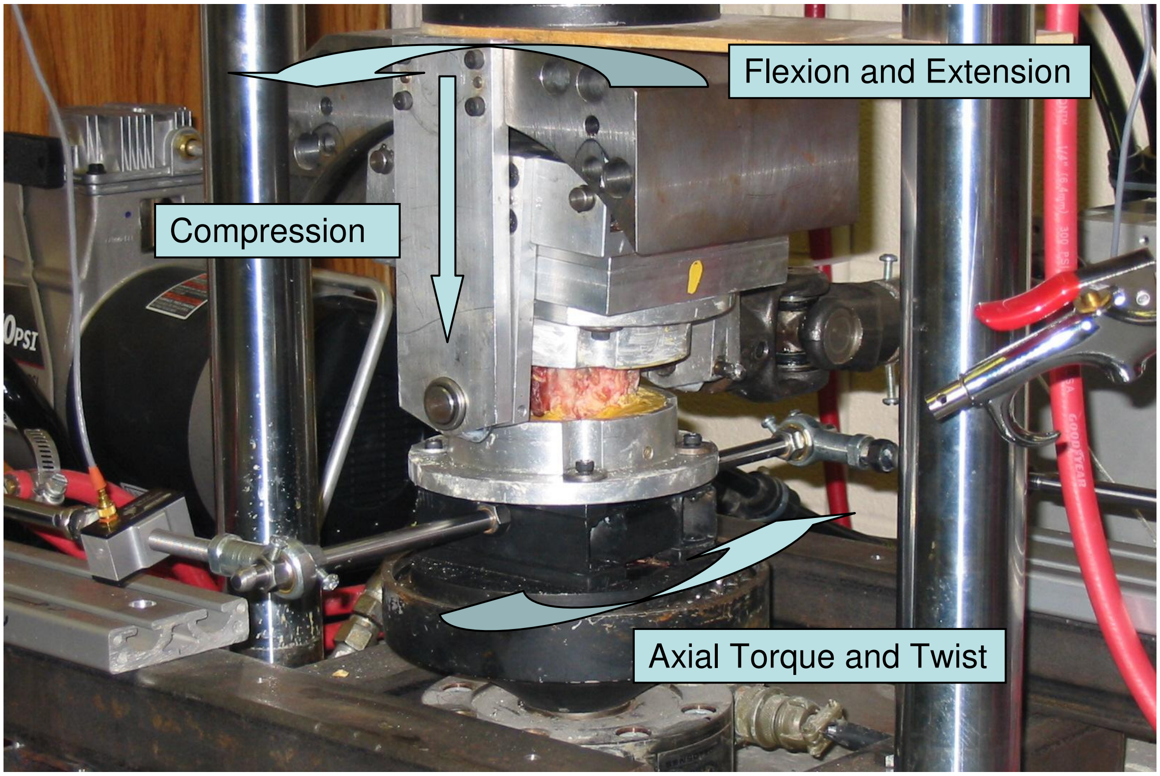


Figure 3.3 This was a photograph of the equipment set-up which allowed loading of the functional spinal units in flexion/extension, compression and axial torque.

3.2 – Testing Group Methodology

Following injection of the radio-opaque dye mixture a computed tomography scan of the functional spinal unit was taken in the transverse plane through the intervertebral disc. The computed tomography voxel size was set at 0.5 mm and the width of the scan was 1.0 mm. The functional spinal units were preloaded for 15 minutes with 300 N of compression to counter any swelling that occurred postmortem (Callaghan and McGill, 2001). During the preload the flexion/extension torque was set to zero and this angular position was considered the zero position for the remainder of the testing protocol (Callaghan and McGill, 2001). The preload was followed by a passive test under 1500 N of compression to determine the flexion and extension testing values. During the passive test the functional spinal units were brought through flexion and extension at a rate of 0.5 °/s and the angular position and torque data was sampled at a rate of 15 Hz. The points on the torque versus angular position curve where the curve deviated from linear was used as the flexion and extension testing values; this was similar to the neutral and elastic zone described by Panjabi et al., (1989). An x-ray image of the functional spinal units was taken in the sagittal plane following the passive test.

A compressive load of 1500 N was used during the remainder of the testing protocol. The functional spinal units were loaded according to one of the following six protocols:

1. 6000 cycles of repetitive flexion extension motions followed by 2000 cycles of axial torque cyclically loaded with 17.5 Nm resulting in rotation to the left and 12.5 Nm resulting in rotation to the right (table 3.1);
2. 6000 cycles of repetitive flexion extension motions followed by 4000 cycles of axial torque cyclically loaded with 17.5 Nm resulting in rotation to the right and 12.5 Nm resulting in rotation to the left (table 3.1);
3. 2000 cycles of axial torque cyclically loaded with 17.5 Nm resulting in rotation to the left and 12.5 Nm resulting in rotation to the right followed by 6000 cycles of repetitive flexion extension motions (table 3.1);
4. 4000 cycles of axial torque cyclically loaded with 17.5 Nm resulting in rotation to the left and 12.5 Nm resulting in rotation to the right followed by 6000 cycles of repetitive flexion extension motions (table 3.1);
5. 6000 cycles of repetitive flexion extension motions only (table 3.1); or

6. 4000 cycles of axial torque only cyclically loaded with 17.5. Nm resulting in rotation to the left and 12.5 Nm resulting in rotation to the right (table 3.1).

Table 3.1 This is a table of the distribution of the functional spinal units in each of the six test groups.

	Flexion/Extension First	Flexion/Extension Second	Twist Only	Flexion/Extension Only
0 Cycles Axial Torque	X	X	X	10
2000 Cycles Axial Torque	10	10	X	X
4000 Cycles of Axial Torque	10	10	5	X

During protocols 1 and 2 the functional spinal units were brought through repetitive flexion extension motions at a cycle rate of 1 Hz for 6000 cycles while under 1500 N of compression. The functional spinal units were repetitively loaded in flexion extension while in position control. The functional spinal units were then x-ray imaged in the sagittal plane. The functional spinal units were then placed into the angular position of zero, where flexion extension torque was found to be zero during the preload test, for the axial torque portion of the study. The pressure gauge of the axial torque jig was set to a pressure in which the functional spinal units were loaded to 17.5 Nm of axial torque (standard deviation 0.5 Nm) resulting in rotation to the left, and 12.5 Nm of axial torque (standard deviation 0.5 Nm) resulting in rotation to the right. The pressure gauge was kept at this pressure level for the remainder of the axial torque portion of the study. 17.5 Nm of axial torque was used as pilot work indicated this load was large enough to cause damage to the intervertebral disc but did not cause fractures to the facet joints. The functional spinal units were exposed to repeated cycles of axial torque for 2000 or 4000 cycles while under 1500 N of compression. The functional spinal units were exposed to repeated cycles of axial torque at a rate of 0.5 Hz; torque and position data were sampled at 50 Hz. Following testing the functional spinal units were x-ray imaged in the sagittal plane. The functional spinal units were injected with 0.15 mL of Omnipaque and a computed tomography scan was taken of the functional spinal unit in the transverse plane through the intervertebral disc. The functional spinal units were sectioned for further visual inspection of the injury pattern.

During protocols 3 and 4 the functional spinal units were placed into the angular position of zero, where the flexion extension torque was found to be zero during the preload, for the axial torque portion of the study. The pressure gauge of the axial torque jig was set to a pressure in which the functional spinal units were loaded to 17.5 Nm of axial torque (standard deviation 0.5 Nm) resulting in rotation to the left and 12.5 Nm of axial torque (standard deviation 0.05 Nm) resulting in rotation to the right. The pressure gauge was kept at this pressure level for the remainder of the test. The functional spinal units were exposed to cyclic loading of axial torque for 2000 or 4000 cycles while under 1500 N compression. The functional spinal units were exposed to cyclic axial torque at a cycle rate of 0.5 Hz and position and torque data were sampled at a rate of 50 Hz. The functional spinal units were then x-ray imaged in the sagittal plane. Next, the functional spinal units were brought through repetitive flexion extension motions at a cycle rate of 1 Hz for 6000 cycles under 1500 N of compression. The functional spinal units were once again x-ray imaged in the sagittal plane. The functional spinal units were injected with 0.15 mL of Omnipaque and a computed tomography scan was taken of the functional spinal unit in the transverse plane through the intervertebral disc. The functional spinal units were sectioned for further visual inspection of the injury patterns.

During protocol 5 the functional spinal units were brought through repetitive flexion extension motions at a cycle rate of 1 Hz for 6000 cycles while under 1500 N of compression. The functional spinal units were then x-ray imaged in the sagittal plane and sectioned for visual inspection of the injury patterns.

During protocol 6 the functional spinal units were placed into the angular position of zero, where flexion extension torque was zero as previously determined during the preload, for axial torque testing. The pressure gauge of the axial torque jig was set to a pressure in which the functional spinal units were loaded with 17.5 Nm of axial torque (standard deviation 0.5 Nm) resulting in rotation to the left and 12.5 Nm of axial torque (standard deviation 0.5 Nm) resulting in rotation to the right. The pressure gauge was kept at this pressure level for the remainder of the test. The functional spinal units were exposed to cyclic axial torque for 4000 cycles while under 1500 N compression. The functional spinal units were exposed to axial torque at a cycle rate of 0.5 Hz and position and torque data were sampled at a rate of 50 Hz. The functional spinal units were then x-ray imaged in the sagittal plane and sectioned for visual inspection of the injury patterns.

The following variables were recorded: endplate area, flexion angle, extension angle, and injury pattern. The endplate area was calculated using the formula for the surface area of an ellipse, $\pi/4 * A * B$, where A was the medial-lateral length of the exposed endplate and B was the anterior-posterior length of the exposed endplate (Tampier, 2006). The average surface area of the exposed endplates was used as the surface area of the intervening disc's endplate. The injury patterns were classified into three groups: no herniation; posterior or posterolateral herniation; or posterior herniation and radial delamination.

The following variables were calculated from the axial torque-rotation curves during axial torque loading (figure 3.4): stiffness (17.5 Nm), stiffness (12.5 Nm), rotation to the left, and rotation to the right. These variables were calculated at cycles 100, 1000 and 2000 for functional spinal units taken to 2000 cycles of axial torque, and at cycles 100, 1000, 2000, 3000 and 4000 for functional spinal units taken to 4000 cycles of axial torque. The stiffness values were measured between 15 – 17Nm of axial torque when loaded to 17.5 Nm and measured between 10 – 12Nm of axial torque when loaded to 12.5 Nm. The rotation values to the left and right were measured with respect to the neutral position as determined during the flexion extension preload.

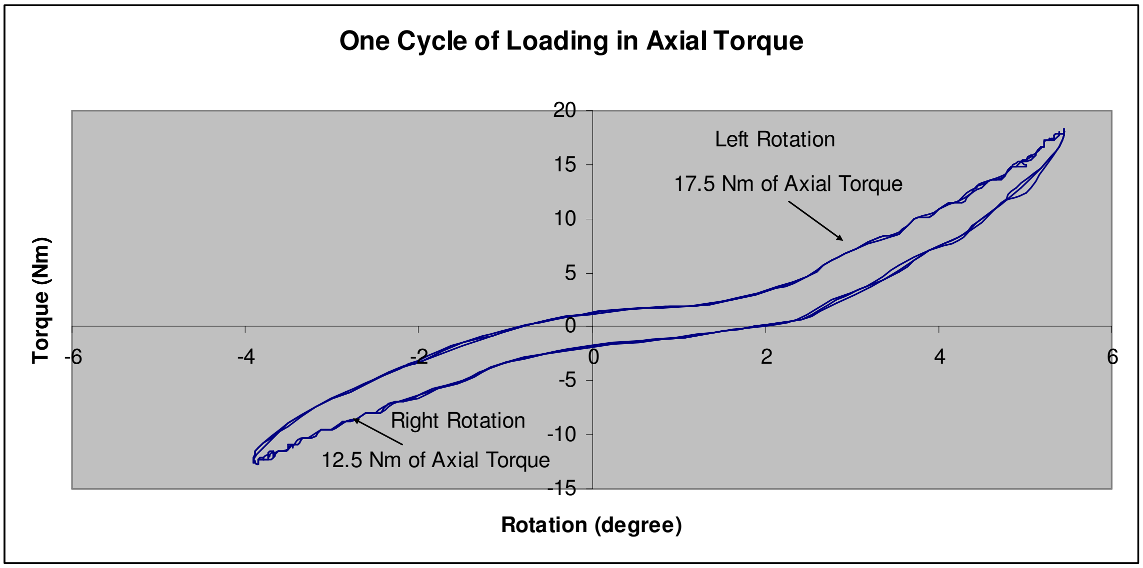


Figure 3.4 A graph of one cycle of loading in axial torque. Cyclic axial loading was created through application of 17.5 Nm of axial torque resulting in rotation to the left and 12.5 Nm of axial torque resulting in rotation to the right.

3.3 - Statistical Analysis

A rejection level of 95% was used for all the ANOVA and post-hoc tests. A one-way ANOVA was used to compare endplate areas of all the groups: 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; 4000 axial torque second; axial torque only; and flexion/extension only. A one-way ANOVA was also used to compare flexion and extension angle differences between the following groups: 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; 4000 axial torque second; and flexion/extension only.

A three factor ANOVA was used to compare stiffness (17.5 Nm), stiffness (12.5 Nm), rotation to the left, and rotation to the right at 100, 1000 and 2000 cycles of axial torque for the following groups: 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; and 4000 axial torque second. The three factors were total cycles of axial torque (2000 or 4000), order of axial torque (first or second) and cycles (100, 1000 or 2000). The least significant differences post-hoc test was used on any variables found to be significantly different.

A two factor ANOVA was used to compare stiffness (17.5 Nm), stiffness (12.5 Nm), rotation to the left, and rotation to the right at 100, 1000, 2000, 3000 or 4000 cycles of axial torque for the following groups: 4000 axial torque first; 4000 axial torque second; and 4000 axial torque only. The two factors were group and cycles (100, 1000, 2000, 3000 or 4000). The least significant differences post-hoc test was used on any variable found to be significantly different.

The injury patterns of the aforementioned groups (2000 axial torque first, 2000 axial torque second, 4000 axial torque first, 4000 axial torque second) were compared using a chi-square. A chi-square was also used to compare injury patterns between these groups (collapsed into one group) and the flexion/extension only group.

The concordance between the x-ray image and visual inspection of the dissected disc and the concordance between the computed tomography scan and visual inspection of the dissected disc were compared using a chi-square.

4 – RESULTS

4.1 – Similarity of Samples

A one-way ANOVA was used to compare the flexion and extension angles of the following groups: 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; 4000 axial torque second; and flexion/extension only. No significant differences were found in either flexion angles or extension angles determined from the passive range of motion tests between the groups. The average flexion angle was 12° (standard deviation = 3°) and the average extension angle was 6° (standard deviation = 2°).

A one-way ANOVA was also used to compare the endplate areas of the following groups: 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; 4000 axial torque second; flexion/extension only; and axial torque only. No significant differences were found in the endplate area of the groups. The average endplate area, collapsed across all the groups, was 5.1 cm^2 (standard deviation = 0.5 cm^2).

4.2 –Injury Patterns

A chi square test was used to compare the injury patterns of the following four groups: 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; and 4000 axial torque second. The injury patterns were classified into one of the following groups: no herniation (0); posterior or posterolateral herniation (1); or posterior herniation and radial delamination (2). The chi square found no significant differences in injury patterns between these four groups (p=0.5296) (Table 4.1). Therefore, these four groups were collapsed into one group (combined group) and compared with the injury patterns of the flexion/extension only group.

Table 4.1 A chart of the chi square distribution comparing the injury patterns of the following groups: 2000 axial torque first, 2000 axial torque second, 4000 axial torque first and 4000 axial torque second. The first number in each cell represents the total number of functional spinal units in that cell and the second number represents the percentage of functional spinal units from that group.

	No Herniation	Posterior or Posterolateral Herniation	Posterior Herniation with Radial Delamination
2000 Axial Torque First	2 20%	2 20%	6 60%
2000 Axial Torque Second	4 40%	1 10%	5 50%
4000 Axial Torque First	1 10%	1 10%	8 80%
4000 Axial Torque Second	2 20%	0 0%	8 80%
Total	9	4	27
Predicted Percent	22.5%	10%	67.5%

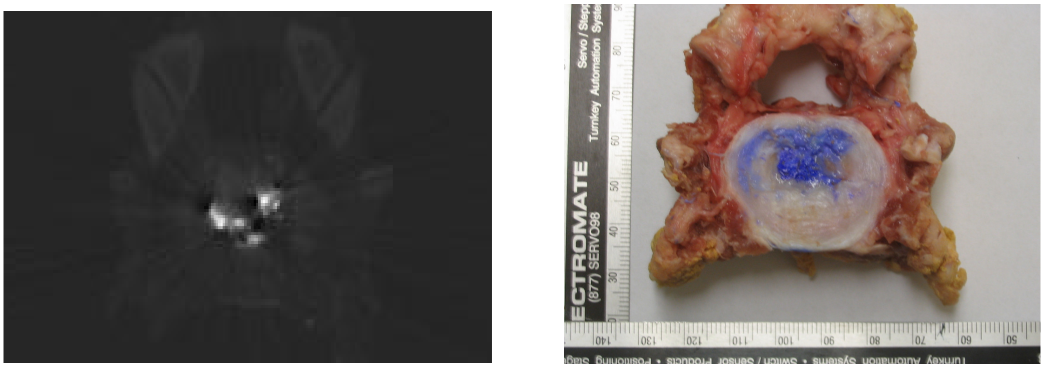
A chi square test was used to compare the injury patterns of the combined group to the flexion/extension only group. The axial torque only group was not included in this analysis as 0% of the functional spinal units tested showed evidence of a disc herniation. The chi square test found significant differences between the injury patterns of the combined group compared to the flexion/extension only group (p=0.0139). The chi square indicated the combined group had more functional spinal units than predicted with a posterior herniation and radial delamination and the flexion/extension only group had more functional spinal units than predicted with posterior or posterolateral herniations (Table 4.2).

Table 4.2 A chart of the chi square distribution comparing the injury patterns of the combined group with the flexion/extension only group. The first number in each cell represents the number of functional spinal units in that cell and the second number represents the percentage of functional spinal units from that group.

	No Herniation	Posterior or Posterolateral Herniation	Posterior Herniation and Radial Delamination
Combined Treatment (Flexion/Extension and Axial Torque)	9 22.5%	4 10%	27 67.5%
Flexion/Extension Only	4 40%	4 40%	2 20%
Total	13	8	29
Predicted Percent	26%	16%	58%

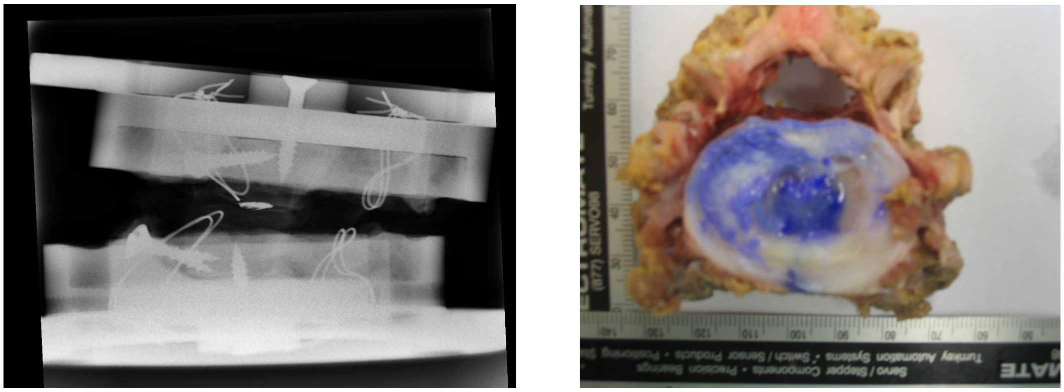
4.3 – Computed Tomography, X-Ray and Dissection Concordance

A chi square was used to compare the concordance between the computed tomography scan and dissection and the x-ray image and dissection. The concordance was classified as no match with the dissected specimen (0) (figure 4.1), partial match with the dissected specimen (1) (figure 4.2), and full match with dissected specimen (2) (figure 4.3).



Computed Tomography Scan = No Herniation

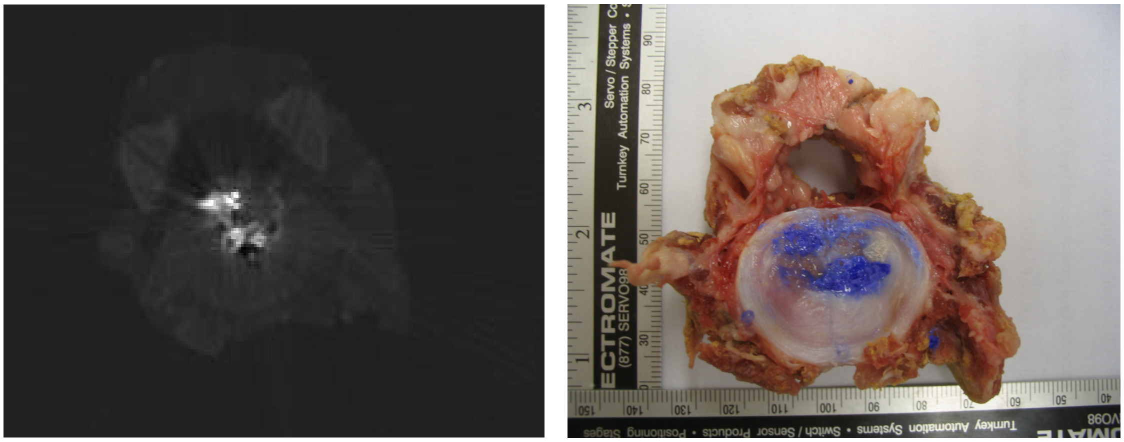
Dissection = Posterior Herniation and Radial Delamination



X-Ray Image = No Herniation

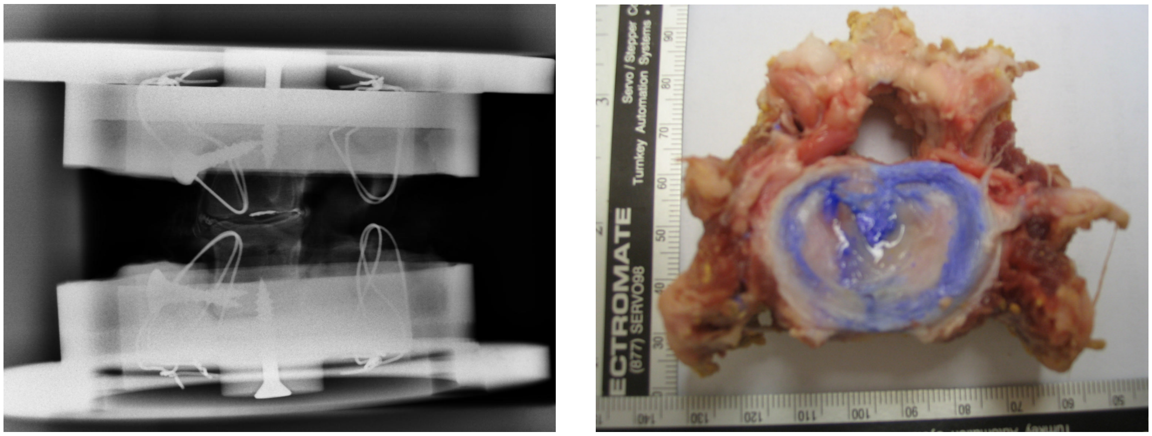
Dissection = Posterior Herniation and Radial Delamination

Figure 4.1 These images were examples of a computed tomography scan and x-ray image which did not match to the dissection.



Computed Tomography Scan =
Posterolateral Herniation

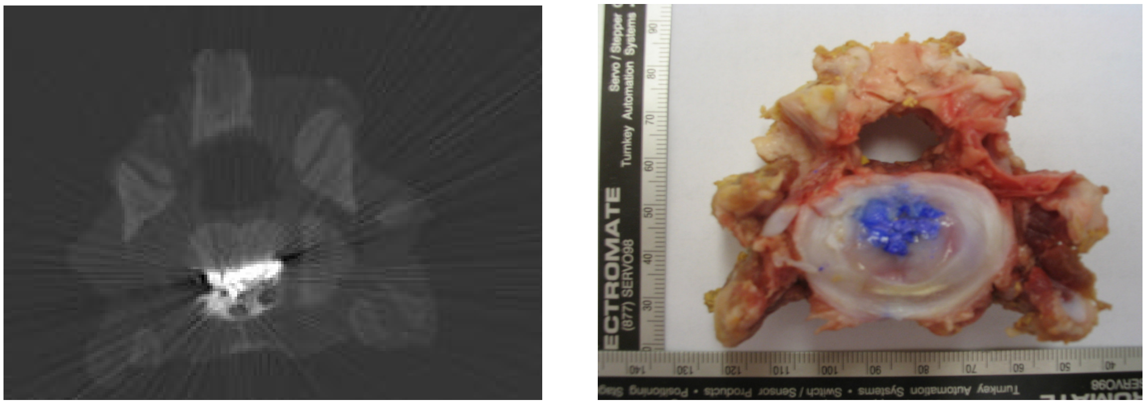
Dissection = Posterior Herniation
and Radial Delamination



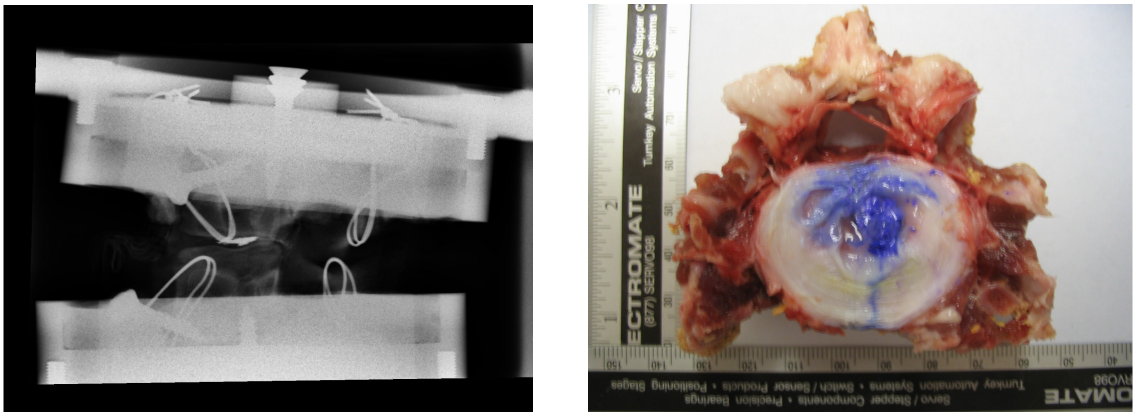
X-Ray Image = Posterior Herniation

Dissection = Posterior Herniation and
Radial Delamination

Figure 4.2 These images were examples of a computed tomography scan and x-ray image which partially matched the dissection.



Computed Tomography Scan = No Herniation Dissection = No Herniation



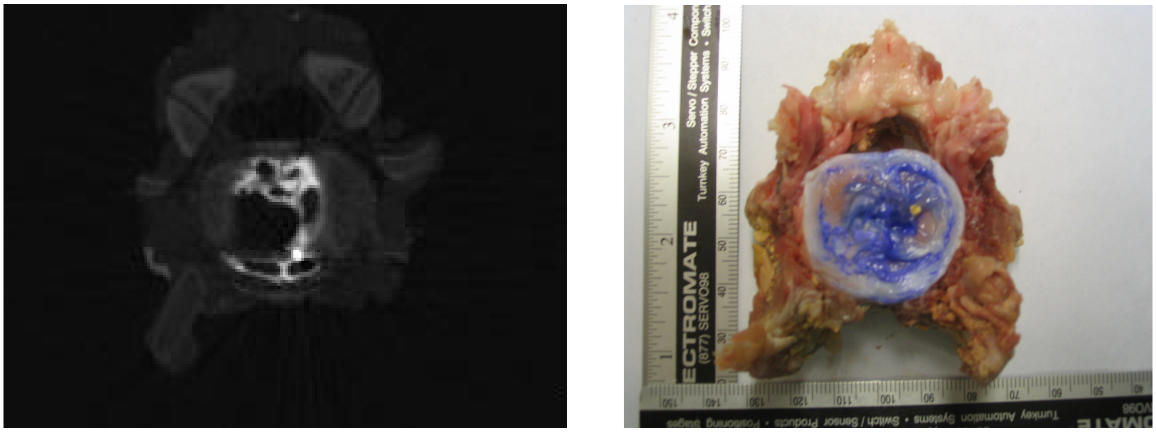
X-Ray Image = Posterior Herniation Dissection = Posterior Herniation

Figure 4.3 These images were examples of computed tomography scans and x-ray images which matched the dissection.

The chi square found no significant differences in the concordance between the x-ray image and dissection and the computed tomography scan and dissection (p=0.9661) (table 4.3). However of note is five of the computed tomography scans were able to detect some radial delamination while zero of the x-ray images were able to detect radial delamination (figure 4.4).

Table 4.3 A table of the chi-square distribution for concordance between the injury patterns observed in the computed tomography scans and x-ray image with dissection. The first number in each cell represents the number of functional spinal units in that cell and the second number represents percentage of functional spinal units in that group.

	No Match	Partial Match	Full Match
CT Scan	6 15%	23 57.5%	11 27.5
X-Ray Image	6 15%	24 60%	10 25%
Total	12	47	21
Predicted Percent	15%	58.75%	26.25%



Computed Tomography Scan = Posterior Herniation and Uni-Radial Delamination

Dissection = Posterior Herniation and Bi-Radial Delamination

Figure 4.4 This was an example of a computed tomography scan in which some delamination was detected in the computed tomography image.

4.4 – Stiffness

Stiffness values were taken at the end ranges of motion during both rotation to the left and right during axial torque. The stiffness values were measured between 15 – 17 Nm of axial torque during rotation to the left and between 10 – 12 Nm of axial torque during rotation to the right. The linear regression lines had an average r^2 value of 0.8909 (standard deviation = 0.0542) for the stiffness values taken during rotation to the left and an average r^2 value of 0.8993 (standard deviation = 0.0478) for the stiffness values taken during rotation to the right.

A three-way ANOVA was used to compare the stiffness values taken during application of both 17.5 Nm of axial torque resulting in rotation to the left and 12.5 Nm of axial torque resulting in rotation to the right between the following four groups: 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; and 4000 axial torque second. The three factors were total twist cycles (2000 or 4000), order (axial torque first or axial torque second), and cycles of axial torque (100, 1000 and 2000). There were no total axial torque cycles or order effects found for the stiffness values.

A cycle effect was found for the stiffness values taken during application of 17.5 Nm of axial torque ($p<0.0001$). The least significant difference post-hoc test indicated the stiffness values taken at 100 cycles of axial torque was significantly different than both 1000 and 2000 cycles of axial torque (figure 4.5 and table 4.4). There was an increase in stiffness with increasing cycles of axial torque. A cycle effect was also found for the stiffness taken during application of 12.5 Nm of axial torque ($p<0.0001$). The least significant difference post-hoc test indicated the stiffness at 100 cycles of axial torque was significantly different than 2000 cycles of axial torque (figure 4.6 and table 4.4.). There was an increase in stiffness with increasing cycles of axial torque.

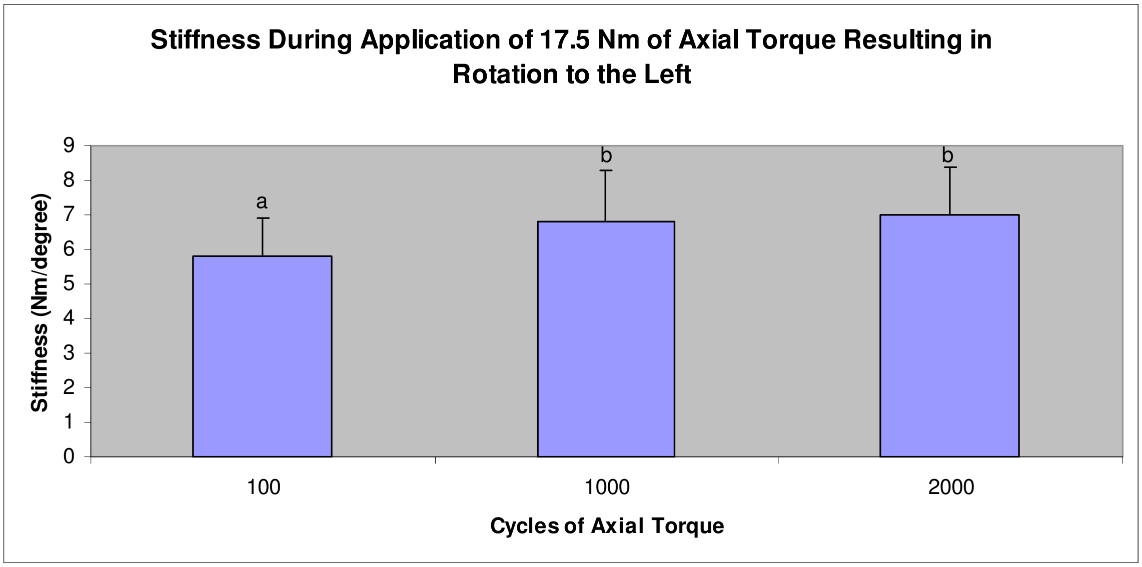


Figure 4.5 A graph of the stiffness at 100, 1000, and 2000 cycles of axial torque during application of 17.5 Nm of axial torque which resulted in rotation to the left. The following groups were compared in this analysis: 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; and 4000 axial torque second. There was a statistically significant cycle effect found and statistical differences found during the least significant differences post-hoc test are marked with different letters.

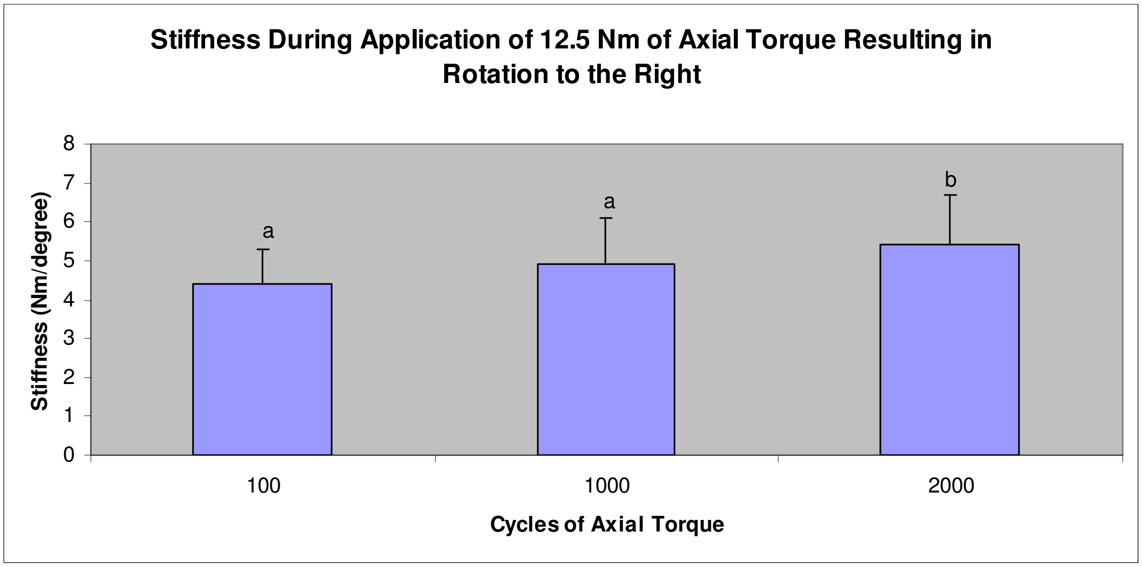


Figure 4.6 A graph of the stiffness at 100, 1000, and 2000 cycles of axial torque taken during application of 12.5 Nm of axial torque resulting in rotation to the right. The following groups were compared in this analysis: 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; and 4000 axial torque second. There was a statistically significant cycle effect found and statistical differences found in the least significant differences post-hoc are marked with different letters.

Table 4.4 A table of the average and standard deviations of the stiffness during application of 17.5 Nm of axial torque resulting in rotation to the left and 12.5 Nm of axial torque resulting in rotation to the right for the following groups: 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; and 4000 axial torque second. These values were calculated from forty functional spinal units.

	100 Cycles of Axial Torque	1000 Cycles of Axial Torque	2000 Cycles of Axial Torque
Stiffness (17.5 Nm)	5.8 (1.1)	6.8 (1.5)	7 (1.4)
Stiffness (12.5 Nm)	4.4 (0.9)	4.9 (1.2)	5.4 (1.3)

A two-way ANOVA was used to compare the stiffness during both application of 17.5 Nm of axial torque resulting in rotation to the left and 12.5 Nm of axial torque resulting in rotation to the right between the following four groups: 4000 axial torque first; 4000 twist second; and 4000 twist only. The factors were group and cycles of axial torque (100, 1000, 2000, 3000 and 4000). There were no group effects found for either stiffness.

A cycle effect was found for the stiffness during application of 17.5 Nm of axial torque ($p<0.0001$). The least significant differences post-hoc test indicated 100 cycles of axial torque was significantly different than 1000, 2000, 3000, and 4000 cycles of axial torque; 1000 cycles of axial torque was significantly different than 3000, and 4000; and 2000 cycles of axial torque was significantly different than 3000, and 4000 cycles of axial torque (figure 4.7 and table 4.5). There was an increase in stiffness with increasing cycles of axial torque. A cycle effect was also found for the stiffness during application of 12.5 Nm of axial torque ($p<0.0001$). The least significant differences post-hoc test indicated 100 cycles of axial torque was significantly different than 1000, 2000, 3000, and 4000 cycles of axial torque; 1000 cycles of axial torque was significantly different than 2000, 3000, and 4000; and 2000 cycles of axial torque was significantly different than 4000 cycles of axial torque (figure 4.8 and table 4.5). There was an increase in stiffness with increasing cycles of axial torque.

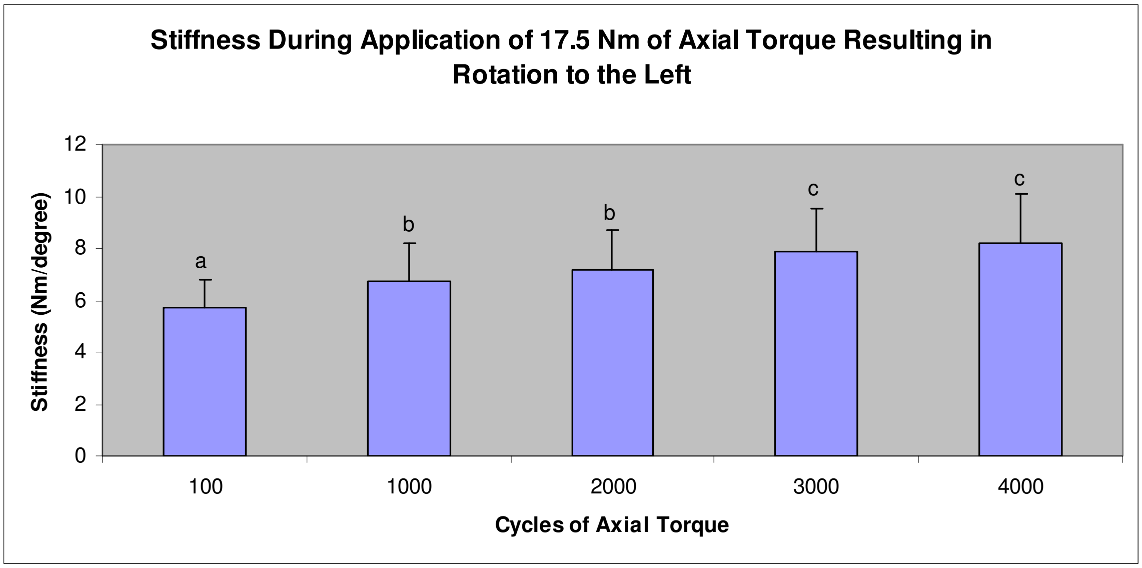


Figure 4.7 A graph of the stiffness taken at 100, 1000, 2000, 3000 and 4000 cycles of axial torque during application of 17.5 Nm of axial torque resulting in rotation to the left. The following groups were compared in this analysis: 4000 axial torque first; 4000 axial torque second; and 4000 axial torque only. There was a statistically significant cycle effect found and statistical differences found during the least significant differences post-hoc test are marked with different letters.

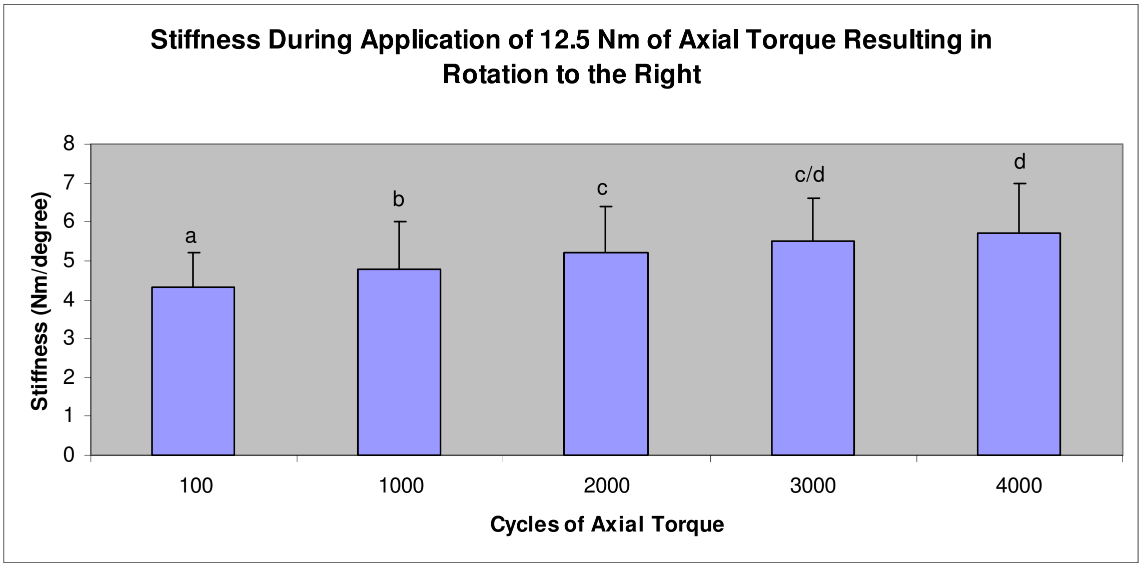


Figure 4.8 A graph of the stiffness taken at 100, 1000, 2000, 3000 and 4000 cycles of axial torque taken during application of 12.5 Nm of axial torque resulting in rotation to the right. The following groups were compared in this analysis: 4000 axial torque first; 4000 axial torque second; and 4000 axial torque only. There was a statistically significant cycle effect found and statistical differences found in the least significant differences post-hoc are marked with different letters.

Table 4.5 A table of the average and standard deviations of the stiffness during application of 17.5 Nm of axial torque resulting in rotation to the left and 12.5 Nm of axial torque resulting in rotation to the right for the following groups: 4000 axial torque first; 4000 axial torque second; and 4000 axial torque only. These values were calculated from twenty-five functional spinal units.

	100 Cycles of Axial Torque	1000 Cycles of Axial Torque	2000 Cycles of Axial Torque	3000 Cycles of Axial Torque	4000 Cycles of Axial Torque
Stiffness (17.5 Nm)	5.7 (1.1)	6.7 (1.5)	7.2 (1.5)	7.9 (1.6)	8.2 (1.9)
Stiffness (12.5 Nm)	4.3 (0.9)	4.8 (1.2)	5.2 (1.2)	5.5 (1.1)	5.7 (1.3)

4.5 – Rotation

A three-way ANOVA was used to compare the rotation of the functional spinal unit during application of 17.5 Nm of axial torque resulting in rotation to the right and 12.5 Nm of axial torque resulting in rotation to the left between the following groups: 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; and 4000 axial torque second. The three factors were order (axial torque first or axial torque second), total number of cycles (2000 or 4000), and cycles of axial torque (100, 1000 and 2000). There were no significant differences in rotation for either order or total number of cycles.

A cycle effect was found for rotation to the left during application of 17.5 Nm of axial torque ($p<0.0001$). The least significant differences post-hoc test indicated there were no significant differences found between the cycles of axial torque (figure 4.9 and table 4.6). There was an increase in rotation with increasing cycles of axial torque. A cycle effect was also found for rotation to the right during application of 12.5 Nm of axial torque (12.5 Nm). The least significant differences post-hoc test indicated 100 cycles of axial torque was significantly different than 2000 cycles of axial torque (figure 4.10 and table 4.6). There was an increase in rotation with increasing cycles of axial torque.

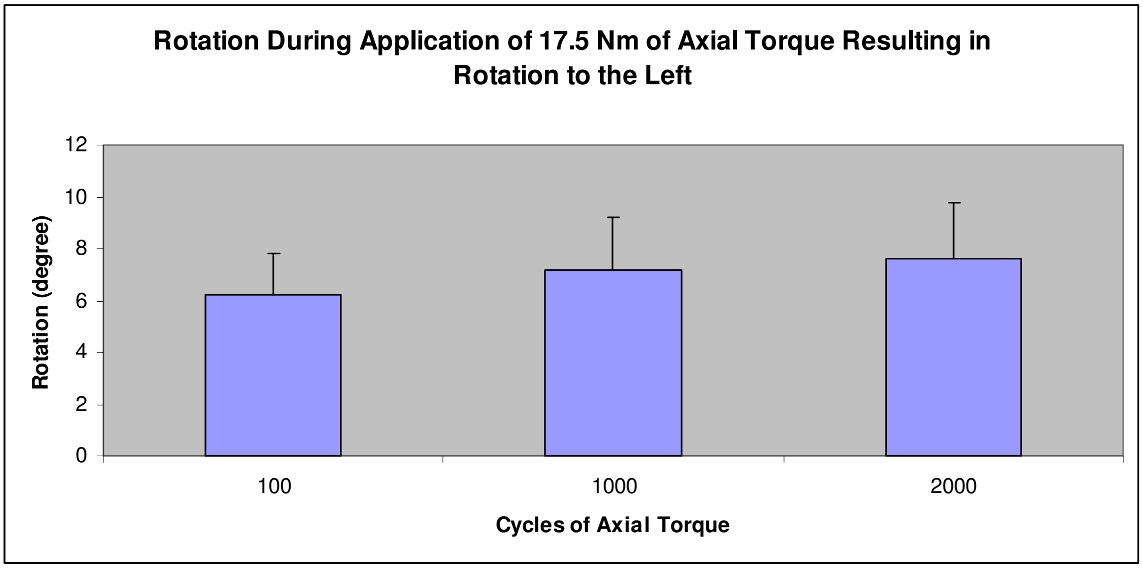


Figure 4.9 A graph of the rotation taken at 100, 1000, and 2000 cycles of axial torque during application of 17.5 Nm of axial torque resulting in rotation to the left. The following groups were compared in this analysis: 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; and 4000 axial torque second. There was a statistically significant cycle effect; however, the least significant differences post-hoc test found no statistical differences between the individual cycles of axial torque.

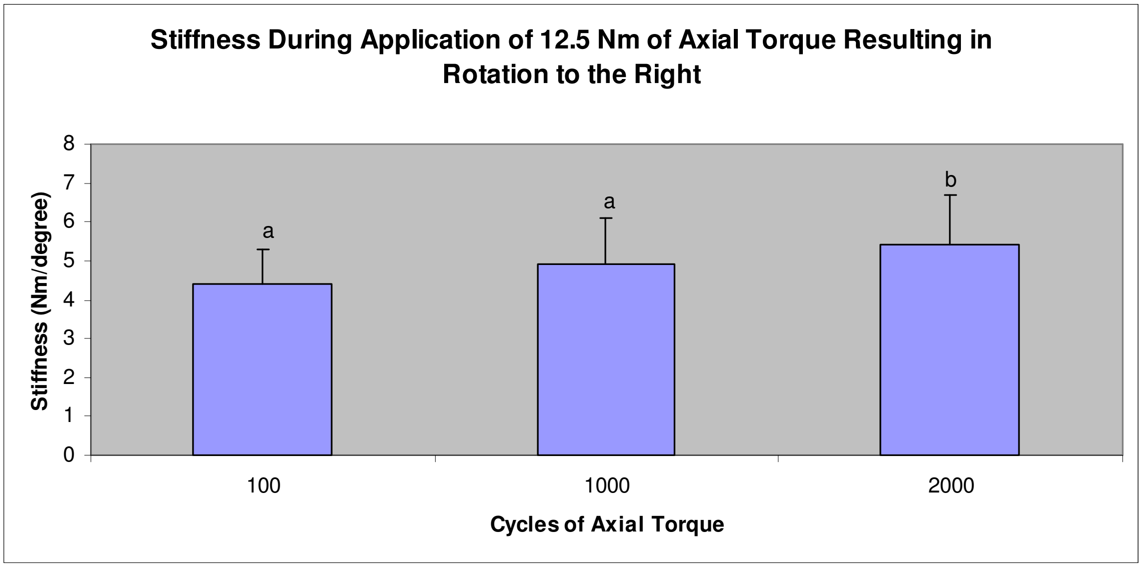


Figure 4.10 A graph of the rotation taken at 100, 1000, and 2000 cycles of axial torque during application of 12.5 Nm of axial torque resulting in rotation to the right. The following groups were included in the analysis: 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; and 4000 axial torque second. A statistically significant cycle effect was found and statistical differences found in the least significant differences post-hoc analysis are marked by different letters.

Table 4.6 A table of the average and standard deviations of the rotations during application of 17.5 Nm of axial torque resulting in rotation to the left and 12.5 Nm of axial torque resulting in rotation to the right for the following groups: 2000 axial torque first, 2000 axial torque second, 4000 axial torque first, and 4000 axial torque second. These values were calculated from forty functional spinal units.

	100 Cycles of Axial Torque	1000 Cycles of Axial Torque	2000 Cycles of Axial Torque
Rotation (17.5 Nm)	6.2 (1.6)	7.2 (2.0)	7.6 (2.2)
Rotation (12.5 Nm)	4.5 (1.0)	5.3 (1.3)	5.6 (1.4)

A two-way ANOVA was used to compare rotation during application of 17.5 Nm of axial torque resulting in rotation to the left and 12.5 Nm of axial torque resulting in rotation to the right between the following groups: 4000 axial torque first; 4000 axial torque second; and 4000 axial torque only. The two factors were the group and cycles of axial torque (100, 1000, 2000, 3000, and 4000). There were no group effects found.

A cycle effect was found in rotation during application of 17.5 Nm of axial torque resulting in rotation to the left (p<0.0001). The least significant differences post-hoc test indicated 100 cycles of axial torque was significantly different than 1000, 2000, 3000, and 4000 cycles of axial torque; 1000 cycles of axial torque was significantly different than 3000, and 4000 cycles of axial torque; and 2000 cycles of axial torque was significantly different than 4000 cycles of axial torque (figure 4.11 and table 4.7). There was an increase in rotation with increasing cycles of axial torque. A cycle effect was also found in rotation during application of 17.5 Nm of axial torque resulting in rotation to the right (p<0.0001). The least significant differences post-hoc test indicated 100 cycles of axial torque was significantly different than 1000, 2000, 3000, and 4000 cycles of axial torque; 1000 cycles of axial torque was significantly different than 2000, 3000, and 4000 cycles of axial torque; and 2000 cycles of axial torque was significantly different than 4000 cycles of axial torque (figure 4.12 and table 4.7). There was an increase in rotation with increasing cycles of axial torque.

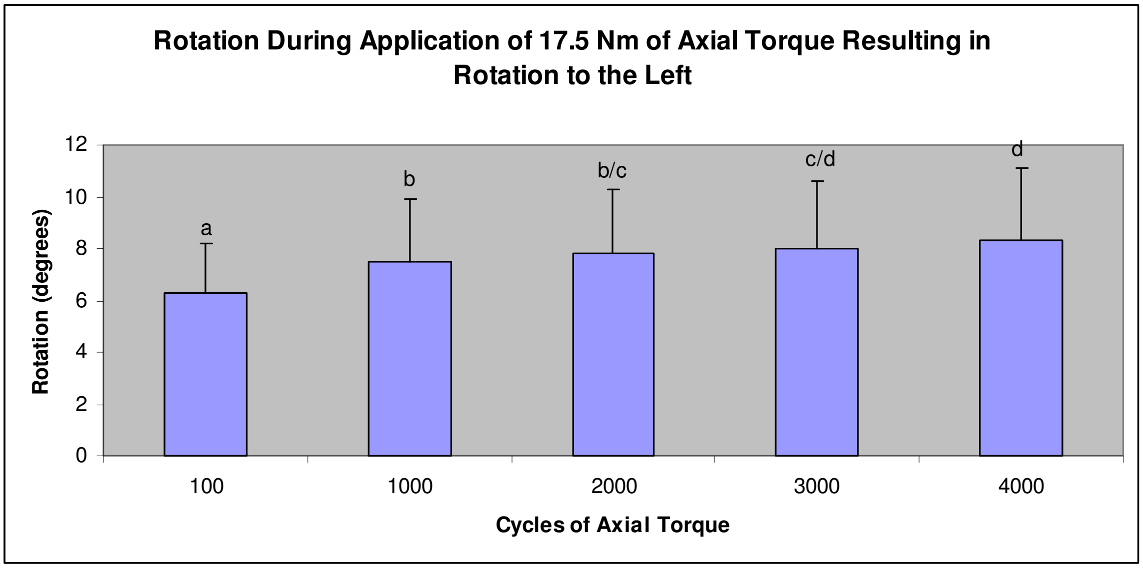


Figure 4.11 A graph of the rotation taken at 100, 1000, 2000, 3000 and 4000 cycles of axial torque during application of 17.5 Nm of axial torque resulting in rotation to the left. The following groups were compared in this analysis: 4000 axial torque first; 4000 axial torque second; and 4000 axial torque only. There was a statistically significant cycle effect and statistical differences found in the least significant differences post-hoc test are marked with different letters.

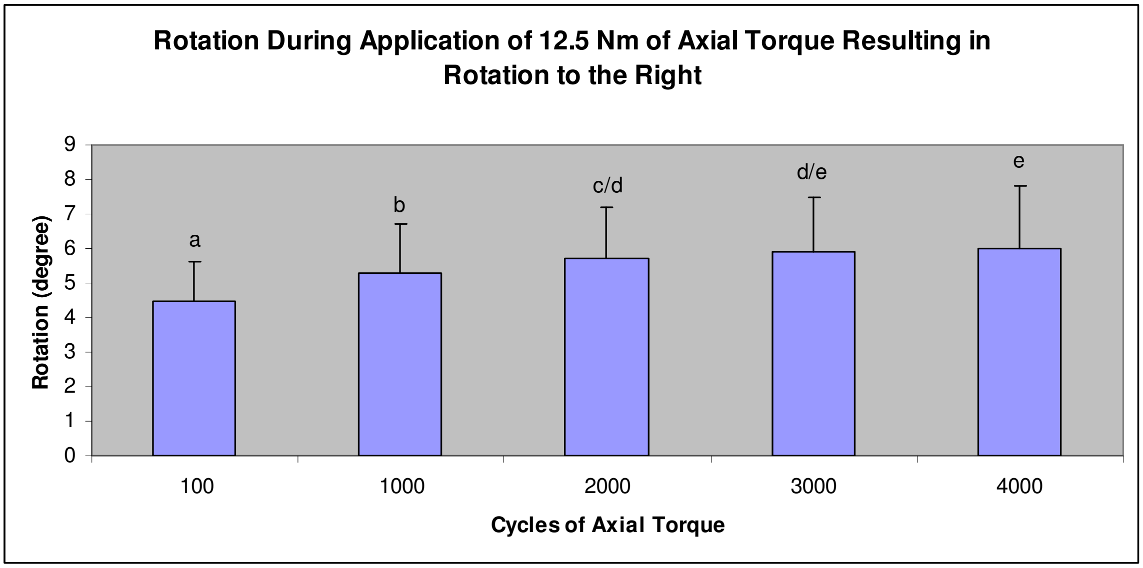


Figure 4.12 A graph of the rotation taken at 100, 1000, 2000, 3000 and 4000 cycles of axial torque during application of 12.5 Nm of axial torque resulting in rotation to the right. The following groups were compared in this analysis: 4000 axial torque first; 4000 axial torque second; and 4000 axial torque only. There was a statistically significant cycle effect and statistical differences found in the least significant differences post-hoc test are marked with different letters.

Table 4.7 A table of the average and standard deviations of the rotations during application of 17.5 Nm of axial torque resulting in rotation to the left and 12.5 Nm of axial torque resulting in rotation to the right for the following groups: 4000 axial torque first; 4000 axial torque second; and 4000 axial torque only. These values were calculated from twenty-five functional spinal units.

	100 Cycles of Axial Torque	1000 Cycles of Axial Torque	2000 Cycles of Axial Torque	3000 Cycles of Axial Torque	4000 Cycles of Axial Torque
Rotation (17.5 Nm)	6.3 (1.9)	7.5 (2.4)	7.8 (2.5)	8.0 (2.6)	8.3 (2.8)
Rotation (12.5 Nm)	4.5 (1.1)	5.3 (1.4)	5.7 (1.5)	5.9 (1.6)	6.0 (1.8)

5 - DISCUSSION

5.1 – Hypotheses

It was hypothesized that the functional spinal units loaded in axial torque only would have circumferential tears in the periphery of the annulus and the nucleus pulposus would be unaffected. Dissection methods produced what appeared to be separation of the annular fibers; therefore, comments could not be made with regards to damage of the annular fibers. Zero of the functional spinal units had herniated after 4000 cycles of axial torque alone; therefore, the nucleus pulposus was not affected by axial torque only. The results from the present study compared well to previous research which also found axial torque alone was not able to initiate a disc herniation (Adams and Hutton, 1981). Previous researchers found that rotating spinal segments to failure produced fractures at the base of the articular processes and circumferential rents to the peripheral of the annulus; however, no disruption of the inner annular fibers were evident (Liu et al., 1985, Adams and Hutton, 1981, Farfan et al., 1970).

It was also hypothesized that the functional spinal units loaded in repetitive flexion extension motions followed by axial torque would result in a posterior herniation and radial delamination with damage to the entire disc; and the functional spinal units loaded in axial torque followed by repetitive flexion extension motions would result in earlier onset of disc herniation and more serious disc destruction. Repetitive flexion extension motions followed by axial torque encouraged a posterior herniation and radial delamination; therefore, the first part of the hypothesis was supported. The injury patterns observed post dissection were not affected by either the order of axial torque (first or second) or the total twist cycles (2000 or 4000). There were no differences in injury patterns of the groups; therefore, axial torque preceding repetitive flexion extension motions did not appear to result in earlier onset of disc herniation and more serious tissue destruction at 6000 cycles of repetitive flexion extension motions; therefore, the second part of the hypothesis was not supported.

5.2 –Injury Patterns

The injury patterns observed in the following four groups were combined to form one group (combined group): 2000 axial torque first; 2000 axial torque second; 4000 axial torque first; and 4000 axial torque second. The combined group's injury patterns were compared to the injury patterns observed in the flexion/extension only group to determine whether axial torque either preceded or followed by repetitive flexion extension motions affected the disc herniation mechanism. The combination of repetitive flexion extension motions and axial torque, regardless of order, encouraged posterior herniations and radial delamination. Gordon et al. (1991) also found lumbar motion segments failed by nuclear extrusion and annular protrusion when loaded repetitively in flexion, rotation, and compression; however, the authors did not discuss if radial delamination was present. Repetitive flexion extension motions only encouraged posterior or posterolateral herniations when taken to 6000 cycles. Therefore, axial torque in combination with repetitive flexion extension motion appeared to play a role in the progression of the nucleus pulposus through the annular layers by encouraging radial delamination.

Contrary to the hypothesis made by Adams et al., (1981) the intervertebral disc was compromised by repetitive axial torque without damaging the articular facet joints. Farfan et al.'s, (1970) hypothesis also failed to hold as repetitive axial torque alone did not lead to intervertebral disc herniations when taken to 4000 cycles of axial torque. Only through the combination of both axial torque and repetitive flexion extension motions was axial torque able to play a role in the disc herniation mechanism.

It was hypothesized there were two potential processes which accounted for the radial delamination depending on the order of axial torque. The functional spinal units loaded in axial torque first most likely weakened the lamellar walls in the outer portion of the annulus fibrosus during repetitive loading by axial torque. The outer layers of the annulus fibrosus were the tissues most likely to sustain the greatest stress during axial torque due to its radial distance from the twisting axis. The repetitive flexion extension motions initiated the disc herniation and the nucleus pulposus migrated to the posterior of the intervertebral disc through the weakest portions of the lamellar walls (Tampier, 2006). Once the nucleus pulposus reached the lamellar walls that were weakened by the axial torque, the nucleus pulposus continued to migrate around the circumference of the annular fibers through the lamellar walls weakened by axial torque.

The functional spinal units loaded in axial torque second only herniated as a result of the initial repetitive flexion extension motions. Axial torque alone would not have initiated the disc herniation process. The repetitive flexion extension motions caused the nucleus pulposus to migrate to the posterior of the intervertebral disc through the weakest portions of the lamellar walls (Tampier, 2006). When the functional spinal units were loaded in axial torque the outer layers of the annulus fibrosus were weakened. The motion created through cyclic axial torque continued to pump the nucleus pulposus through the weakened lamellar walls and radial delamination continued along the periphery of the annulus fibrosus.

5.3 – Concordance of X-Ray and Computed Tomography with Dissection

Both x-ray images and computed tomography scans are used to diagnose intervertebral disc herniations (Mundt et al., 1993, Gordon et al, 1991, Butler et al., 1990). In the present study no differences were found in the ability to detect intervertebral disc injuries using the x-ray images versus the computed tomography scan. X-Ray images and computed tomography scans were equally good at identifying functional spinal units which had not herniated or herniated in a posterior or posterolateral direction; however, neither were good at identifying functional spinal units with radial delamination. Radial delamination occurred in 27 of the 40 functional spinal units tested but was only evident in five of the computed tomography scans and zero of the x-ray images. This is cause for concern as computed tomography was unable to identify the path taken by the nucleus pulposus in 81.5% of the functional spinal units with radial delamination and x-ray imaging was unable to identify the path taken by the nucleus pulposus in 100% of the functional spinal units with radial delamination. Therefore, rehabilitation which focuses treatment on the diagnosis of these images would not be affective at treating intervertebral discs with radial delamination.

5.4 –Rotation

The functional spinal units became more compliant in rotation with increasing cycles of axial torque. The degree of rotation of the functional spinal units was not affected by the order in with they were loaded in axial torque versus repetitive flexion extension. This indicated that repetitive flexion extension motions had no effect on the intervertebral discs ability to resist axial torque. The functional spinal units continued to have increased rotation up to 4000 cycles of axial torque.

The degree of rotation found in the present study was larger than rotations found by other researchers (Ochia et al., 2006, Krismer et al., 2000, Scultz et al., 1979, Pearch and Tibrewal, 1984). Krismer et al., (2000) tested human lumbar spine segments in axial torque by applying moments in five steps to 8.5 Nm. The resultant rotations measured were 0.9 – 2.3° for normal discs. These values compared well with rotations found in other *in vivo* studies (Ochia et al., 2006, Pearch and Tibrewal, 1984). Ochia et al., (2006) measured rotation of lumbar segments using computed tomography and found 0.6 – 2.2° of axial rotation to the left and right when subjects were passively rotated 50° to the left and right from neutral. Similarly, Pearcy and Tibrewal (1984) found 2 - 3° of rotation during maximal voluntary twisting as measured from x-ray images.

The higher magnitude of axial torque used in the present study may partially explain the larger rotations observed. Schultz et al., (1979) applied loads in a quasi-static manner to cadaver motion segments to test the mechanical behavior in response to these loads. The motion segment's rotation on average was 0.69° with 4.6 Nm of applied axial torque and 1.5° with 10.6 Nm of applied axial torque. The rotations of these motion segments were dependent on the load magnitude. Another explanation for the larger rotations observed was there was movement between the cups and the functional spinal units during testing.

The increased rotation observed with increased number of axial torque cycles may be an indication that the intervertebral disc was becoming damaged. Haughton et al., (2000) compared the response of cadaver lumbar motion segments with normal discs to those with transverse tears and radial tears to axial torque. The authors found the axial rotation of the normal intervertebral discs (1 - 1.2°) was significantly smaller compared to the intervertebral discs with transverse tears (2.5 – 2.7°) or radial tears (2.5 – 2.6°). These injuries match well with those observed in the present study and help to explain the increased rotation with increased cycles of axial torque. As the intervertebral disc became more damaged it became more compliant in rotation and greater rotations were observed with the same applied axial torque.

5.5 - Stiffness

The functional spinal units also became increasing more stiff in response to axial torque with increasing cycles. The stiffness of the functional spinal units were not affected by the order in which they were exposed to axial torque versus repetitive flexion extension motions; therefore, repetitive flexion extension motions did not affect the stiffness of the intervertebral disc in response to axial torque. This appeared to be the first study to look at the effect of cyclic axial torque on the stiffness of the spinal segments to axial torque.

The stiffness values found in the present study compared well with the stiffness values found in previous research (Haughton et al., 1999, Schmidt et al., 1998). Haughton et al., (1999) measured stiffness of lumbar spinal segments in response to an application of 6.6 Nm of axial torque. The resulting stiffness of the spinal segments was 7 Nm/degree. Schmidt et al., (1998) found spinal motion segments had a stiffness value of 8.4 Nm/degree when loaded with 6.6 Nm of axial torque in a step-wise manner. The authors also found the spinal motion segments were most stiff in response to axial torque compared to flexion torque, extension torque and lateral bend torque. Thompson et al., (2000) measured the rotation of spinal segments in response to dynamic axial torque and found stiffness values slightly lower than both Haughton et al., (1999) and Schmidt et al., (1998). The segments were rotated at a rate of 20°/minute and the stiffness was measured between 3 – 4 Nm of axial torque. The stiffness values found in Thompson et al., 2000 ranged from 2.24 – 5.45 Nm/degree.

There was an increase in stiffness in response to repetitive cyclic axial torque. The increase in stiffness could be due to increase in damage to the intervertebral disc with increasing cycles of axial torque. Callaghan and McGill (2001) found increase stiffness in flexion extension of porcine functional spinal units in response to repetitive flexion extension motions with increasing exposure time. Drake (2008) also found increase stiffness in flexion extension when porcine functional spinal units were loaded to failure by axial torque; however, the stiffness response to axial torque was not tested.

Thompson et al. (2000) tested the stiffness response in torsion and found a decrease in stiffness with increasing severity of rim lesions and concentric tears to the intervertebral disc. This study differed from the present study in that they did not control the motion patterns that caused the observed tears. The authors compared the stiffness of cadaver spinal segments with pre-existing injuries and it may be the motion patterns that caused the injuries which resulted in the decreased stiffness in response to axial torque observed with rim lesion and concentric tears.

5.6 – Limitations

The main limitation in this study was the inability to prevent movement between the steel cups and the vertebrae. This movement affected the rotations of the functional spinal units recorded during testing. Consequently, the rotations measured during axial torque were not solely due to rotation of the intervertebral disc and was also rotation of the functional spinal units and the steel cups.

Another limitation of the current study was using an animal model to determine the effect of axial torque on disc herniation mechanisms. However, previous work has shown the porcine cervical spines were a good approximation to the human lumbar spine functionally, geometrically and anatomically and most closely resemble the spine of a human adolescent (Tampier, 2006, Yingling et al., 1999, Oxland et al., 1991).

6 – CONCLUSION

The purpose of this study was to determine the injury patterns of functional spinal units exposed to axial torque either preceded or followed by repetitive flexion extension motions. Axial torque in combination with repetitive flexion extension motions, regardless of order, encouraged radial delamination. This was contrasted by functional spinal units exposed to repetitive flexion extension only which encouraged posterior or posterolateral herniations. Axial torque alone, unaccompanied by repetitive flexion extension motions either prior to or following axial torque, was not able to initiate a disc herniation.

The functional spinal units became more compliant in rotation as well as increasingly stiff with increasing cycles of axial torque. Rotation and stiffness were not affected by the order in which flexion extension and axial torque occurred. Ability to identify injury patterns using x-ray images and computed tomography scans were not statistically different; however, computed tomography was able to detect some radial delamination in a small proportion of the cases.

Future work should examine how combined loading in flexion, extension, and axial torque as well as axial torque in a flexed posture influence the injury patterns observed in the present study.

REFERENCES

Adams, M.A., Dolan, P., 1995. Recent Advances in Lumbar Spinal Mechanics and Their Clinical Significance. Clinical Biomechanics. 101, 3-19.

Adams, M.A., Hutton, W.C., 1985. Gradual Disc Prolapse. Spine. 106, 524-531.

Adams, M.A., Hutton, W.C., 1982. Prolapsed Intervertebral Disc: A Hyperflexion Injury 1981 Volvo Award in Basic Science. Spine. 73, 184-191.

Adams, M.A., Hutton, W.C., 1981. The Relevance of Torsion to the Mechanical Derangement of the Lumbar Spine. Spine. 63, 241-248.

Andersson, G.B., 1981. Epidemiologic Aspects on Low-Back Pain in Industry. Spine. 61, 53-60.

Aultman, C.D., Scannel, J., McGill, S.M., 2005. The Direction of Progressive Herniation in Porcine Spine Motion Segments is Influenced by the Orientation of the Bending Axis. Clinical Biomechanics. 20, 126-129.

Beach. T.A.C., Parkinson, R.J., Stothart, J.P., Callaghan, J.P., 2005. Effects of Prolonged Sitting on the Passive Flexion Stiffness on *In Vivo* Lumbar Spine. The Spine Journal. 5, 145-154.

Butler, D., Trafimow, J.H., Andersson, G.B.J., McNeill, T.W., Huckman, M.S., 1990. Discs Degenerate Before Facets. Spine. 15, 111-113.

Callaghan, J.P., McGill, S.M., 2001. Intervertebral Disc Herniation: Studies on a Porcine Model Exposed to Highly Repetitive Flexion/Extension Motion with Compressive Force. Clinical Biomechanics. 16, 28-37.

Deyo, R.A., Loeser, J.D., Bigos, S.J., 1990. Herniated Lumbar Intervertebral Disk. Annals of Internal Medicine. 112, 598-603.

Drake, J.D., 2008. Axial Twist Loading of the Spine: Modulators of Injury Mechanisms and the Potential for Pain Generation. Doctor of Philosophy thesis, University of Waterloo, Waterloo.

Drake, J.D., Aultman, C.D., McGill, S.M., Callaghan, J.P., 2005. The Influence of Static Axial Torque in Combine Loading on Intervertebral Joint Failure Mechanics Using a Porcine Model. Clinical Biomechanics. 20, 1038-1045.

Duncan, N.A., Ahmed, A.M., 1991. The Role of Axial Rotation in the Etiology of Unilateral Disc Prolapse. An Experimental and Finite-Element Analysis. Spine. 16, 1089-1098.

Fardon, D.F., Milette, P.C., 2001. Nomenclature and Classification of Lumbar Disc Pathology: Recommendations of the Combined Task Forces of the North American Spine Society, American Society of Spine Radiology, and American Society of Neuroradiology. 26, E93-E113.

Farfan, H.F., 1984. The Torsional Injury of the Lumbar Spine. Spine. 91, 53.

Farfan, H.F., 1973. Mechanical Disorders of the Low Back, 1 ed. , Henry Kimpton Publishers, Great Britain.

Farfan, H.F., Cossette, J.W., Robertson, G.H., Wells, R.V., Kraus, H., 1970. The Effects of Torsion on the Lumbar Intervertebral Joints: The Role of Torsion in the Production of Disc Degeneration. Journal of Bone and Joint Surgery American Volume. 523, 468-497.

Farfan, H.F., 1969. Effects of Torsion on the Intervertebral Joints. Canadian Journal of Surgery. 123, 336-341.

Frymoyer, J.D., 1988. Back Pain and Sciatica. New England Journal of Medicine. 318, 291-300.

Galante, J.O., 1967. Tensile properties of the human lumbar annulus fibrosus. Acta Orthopaedica Scandinavica. Suppl. 1005-91.

Gordon, S.J., Yang, K.H., Mayer, P.J., Mace, A.H., Kish, V.L., Radin, E.L., 1991. Mechanism of Disc Rupture: A Preliminary Report. 6, 450-456.

Greenough, C.G., Fraser, R.D., 1994. Aetiology, Diagnosis and Treatment of Low Back Pain. European Spine Journal. 31, 22-27.

Haughton, V.M., Rogers, B., Meyerand, M.E., Resnick, D.K., 2002. Measuring the Axial Rotation of Lumbar Vertebrae in Vivo with MR Imaging. American Journal of Neruoradiology. 23, 1110-1116.

Haughton, V.M., Schmidt, T.A., Keele, K., An. H.S., Lim, T.H., 2000. Flexibility of Lumbar Spinal Motion Segments Correlated to Type of Tears in the Annulus Fibrosus. Journal of Neurosurgery. 91, 81-86.

Haughton, V.M., Lim, T.H., An, H., 1999. Intervertebral Disk Appearance Correlated with Stiffness of Lumbar Spinal Motion Segments. American Journal of Neuroradiology. 20, 1161-1165.

Haughton, V.M., 1988. MR Imaging of the Spine. Radiology. 166, 297-301.

Hickey, D.S., Hukins, D.W., 1979. Effect of Methods of Preservation on the Arrangement of Collagen Fibrils in Connective Tissue Matrices: An X-Ray Diffraction Study of Annulus Fibrosus. Connective Tissue Research. 6, 223-228.

Hicks, G.S., Duddlestone, D.N., Russell, L.D., Holman, H.E., Shepherd, J.M., Brown, A., 2002. Low Back Pain. The American Journal of the Medical Sciences. 324, 207-211.

Hukins, D.W., Meakin, J.R., 2000. Relationship Between Structure and Mechanical Function of the Tissues of the Intervertebral Joint. American Zoologist. 40, 42-52.

Kelsey, J.L., Githens, P.B., Walter, S.D., Southwick, W.O., Weil, U., Holford, T.R., Ostfeld, A.M., Calogero, J.A., O'Connor, T., White, A.A., 1984. An Epidemiological Study of Lifting and Twisting on the Job and Risk for Acute Prolapsed Lumbar Intervertebral Disc. Journal of Bone and Joint Surgery American Volume. 2, 61-66.

Ko, H.Y., Park, B.K., 1997. Facet Tropism in Lumbar Motion Segments and Its Significance in Disc Herniation. Archives of Physical Medicine and Rehabilitation. 78, 1211-1214.

Krismer, M., Haid, C., Behensky, H., Kapfinger, P., Landauer, F., Rachbauer, F., 2000. Motion in Lumbar Functional Spine Units During Side Bending and Axial Rotation Moments Depending on the Degree of Degeneration. Spine. 25, 2020-2027.

Krismer, M., Haid, C., Rabl, W., 1996. The Contribution of Annulus Fibers to Torque Resistance. Spine. 21, 2551-2557.

Liu, Y.K., Goel, V.K., DeJong, A., Njus, G.O., Nishiyama, K., Buckwalter, J., 1985. Torsional Fatigue of the Lumbar Intervertebral Joints. Spine. 10, 894-900.

Marras, W.S., 2000. Occupational Low Back Disorder Causation and Control. Ergonomics. 43, 880-902.

Marras, W.S., Lavender, S.A., Leurgans, S.E., Rajulu, S.L., Allread, W.G., Fathallah, F.A., Ferguson, S.A., 1993. The Role of Dynamic Three-Dimensional Trunk Motion In Occupationally-Related Low Back Disorders. Spine. 18, 617-628.

McGill, S.M., 2002. Low Back Disorders: Evidence-Based Prevention and Rehabilitation, 1 ed. , Human Kinetics Canada Limited.

Mundt, D.L., Kelsey, J.L., Golden, A.L., Pastides, H., Berg, A.T., Sklar, J., Hosea, T., Panjabi, M.M., 1993. An Epidemiological Study of Non-Occupational Lifting as a Risk Factor for Herniated Lumbar Intervertebral Disc. Spine. 18, 595-602.

Ochia, R.S., Inoue, M., Renner, S.M., Lorenz, E.P., Lim, T., Andersson, G.B.J., An, H.S., 2006. Three-Dimensional In Vivo Measurement of Lumbar Spine Segmental Motion. Spine. 31, 2073-2078.

Osti, O.L., Vernon-Roberts, B., Fraser, R.D., 1990. Annulus Tears and Intervertebral Disc Degeneration: An Experimental Study Using an Animal Model. Spine. 15, 762-767.

Osti, O.L., Vernon-Roberts, B., Moore, R., Fraser, R.D., 1992. Annular Tears and Disc Degeneration in the Lumbar Spine. The Journal of Bone and Joint Surgery. 74-B, 678-682.

Oxland, T.R., Panjabi, M.M., Southern, E.P., Duranceau, J.S., 1991. An Anatomical Basis for Spinal Instability: A Porcine Trauma Model. Journal of Orthopaedic Research. 9, 452-462.

Panjabi, M.M., Duranceau, J.S., Oxland, T.R., Bowen, C.E., 1989. Multidirectional Instabilities of Traumatic Cervical Spine Injuries in a Porcine Model. Spine. 14, 1111-1115.

Pearcy, M.J., Hindle, R.J., 1991. Axial rotation of lumbar intervertebral joints in forward flexion. Proceedings of the Institution of Mechanical Engineers. 205, 205-209.

Pearcy, M.J., Tibrewal, S.B., 1984. Axial Rotation and Lateral Bending in the Normal Lumbar Spine Measured by Three-Dimensional Radiography. Spine. 9, 582-587.

Plante, D.A., Rothwell, M.G, Tufo, H.M., 1997. The Adult Spine: Principles and Practice, 2nd ed., Lippincott-Raven.

Schmidt, T.A., Howard, S.A., Lim, T., Nowicki, B.H., Haughton, V.M., 1998. The Stiffness of Lumbar Spinal Motion Segments With a High-Intensity Zone in the Annulus Fibrosus. Spine. 23, 2167-2173.

Schultz, A.B., Warwick, D.N., Berkson, M.H., Nachemson, A.L., 1979. Mechanical Properties of Human Lumbar Spine Motion Segments – Part I: Responses in Flexion, Extension, Lateral Bending and Torsion. Transactions of American Society of Mechanical Engineering. 101, 46-52.

Tampier, Claudio, 2006. Progressive Disc Herniation: An Investigation of the Mechanism using Histochemical and Microscopic Techniques. Masters thesis, University of Waterloo, Waterloo.

Thompson, R.E., Pearcy, M.J., Downing, K.J.W., Manthey, B.A., Parkinson, I.H., Fazzalari, N.L., 2000. Disc Lesions and the Mechanics of the Intervertebral Joint Complex. Spine. 25, 3026-3035.

Tsuji, H., Hirano, N., Ohshima, H., Ishihara, H., Terahata, N. Motoe, T., 1993. Structural Variation of the Anterior and Posterior Annulus Fibrosus in the Development of Human Lumbar Intervertebral Disc: A Risk Factor for Intervertebral Disc Rupture. 18, 204-210.

Urban, J.P.G., McMullin, J.F., 1988. Swelling Pressure of the Lumbar Intervertebral Discs: Influence of Age, Spinal Level, Composition, and Degeneration. Spine. 13, 179-187.

Wilder, D.G., Pope, M.H., Frymoyer, J.W., 1988. The Biomechanics of Lumbar Disc Herniation and the Effect of Overload and Instability. Journal of Spinal Disorders. 11, 16-32.

Yingling, V.R., Callaghan, J.P., McGill, S.M., 1999. The Porcine Cervical Spine as a Model of the Human Lumbar Spine: An Anatomical, Geometric, and Functional Comparison. Journal of Spinal Disorders. 12, 415-423.