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Research Article

The contribution of facet joints, axial compression, and composition to human lumbar disc torsion mechanics†

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Abstract

Stresses applied to the spinal column are distributed between the intervertebral disc and facet joints. Structural and compositional changes alter stress distributions within the disc and between the disc and facet joints. These changes influence the mechanical properties of the disc joint, including its stiffness, range of motion, and energy absorption under quasi-static and dynamic loads. There have been few studies evaluating the role of facet joints in torsion. Furthermore, the relationship between biochemical composition and torsion mechanics is not well understood. Therefore, the first objective of this study was to investigate the role of facet joints in torsion mechanics of healthy and degenerated human lumbar discs under a wide range of compressive preloads. To achieve this, each disc was tested under four different compressive preloads (300-1200 N) with and without facet joints. The second objective was to develop a quantitative structure-function relationship between tissue composition and torsion mechanics. Facet joints have a significant contribution to disc torsional stiffness (~60%) and viscoelasticity, regardless of the magnitude of axial compression. The findings from this study demonstrate that annulus fibrosus GAG content plays an important role in disc torsion mechanics. A decrease in GAG content with degeneration reduced torsion mechanics by more than an order of magnitude, while collagen content did not significantly influence disc torsion mechanics. The biochemical-mechanical and compression-torsion relationships reported in this study allow for better comparison between studies that use discs of varying levels of degeneration or testing protocols and provide important design criteria for biological repair strategies.

Keywords: intervertebral disc; torsion; facet joints; composition; degeneration

Introduction

The intervertebral disc provides the spine with flexibility and stability over a wide range of motions. Stresses applied to the spinal column are distributed between the intervertebral disc and surrounding structures, including facet joints, which are diarthrodial joints posterior to the vertebral column.[1] Facet joints play a crucial role in restricting excessive motions and thereby protecting the disc from mechanical overloading and structural damage.[2] Compression is the primary loading modality placed on the disc, but the disc also experiences complex coupled motions, including bending and axial rotation. Large complex loads have been shown to cause tissue remodeling and increase the risk of micro-damage and failure.[3-5] Moreover, there is a higher prevalence of lower back pain in people that experience large daily loads with rotation, including factory workers, athletes, and military service personnel.[6-9] While extensive research has been performed to understand changes in compressive mechanics with degeneration, there have been few studies investigating disc torsion mechanics with degeneration.

The relative contribution of load sharing between the intervertebral disc and facet joints depends on posture and loading modality.[10,11] It is difficult to evaluate the effects of injury or degeneration on disc mechanics, separate from degenerative changes in the facet joints. Therefore, most biomechanical studies remove the facet joints to study disc mechanics separately from the entire disc-joint. The role of facet joints on disc-joint mechanics has been investigated through serial testing of the disc-joint before and after facet joint removal (*i.e.*, facetectomy).[12-16] These studies showed that facet joints support up to 25% of axial compressive forces[15] and 40-65% of rotational and shear forces in healthy disc-joints.[15,16] Characterizing the contribution of the facet joints for healthy and degenerated human discs under axial rotation is important for understanding observed changes in torsional behavior of the spinal column *in vivo*.[17]

Structural and compositional changes, with injury or degeneration, alter stress distribution within the disc and between the disc and facet joints.[14,18-20] These changes influence the mechanical properties of the disc joint, including its stiffness, range of motion, and energy absorption under quasi-static and dynamic loads,[19,21-23] and increase the likelihood of annular tears.[24] For example, radial tears, which are precursors for herniated or bulging discs, reduce joint stiffness in torsion, lateral bending, flexion, and extension.[21] Mild to moderate degeneration has also been shown to decrease stiffness in torsion, but joint stiffness increases again with severe degeneration.[22,23] Assessment of disc degeneration in these early studies has been largely limited to visual inspection of gross morphology or radiographic images; therefore, the relationship between tissue composition and torsion mechanics is not well understood.

The objectives of this study were to 1) investigate the role of facet joints in torsion mechanics of healthy and degenerated human lumbar disc joints under a wide range of compressive preloads, and 2) develop a quantitative structure-function relationship between tissue composition and torsion mechanics. We hypothesize that torsional mechanical properties of human lumbar discs will depend on the presence of facet joints, axial compressive preload, and biochemical composition of the disc's subcomponents (nucleus pulposus and annulus fibrosus).

Methods

Specimens and preparation

Seven human lumbar spine segments were obtained using an IRB approved protocol (age range: 43 – 80 years; Table 1). T2-weighted magnetic resonance images were obtained from the mid-sagittal plane to grade disc degeneration based on the Pfirrmann scale.[25] After imaging, the surrounding musculature and ligaments were removed with a scalpel. Bone-disc-bone motion segments with intact facet joints were prepared from the L3-L4 and/or L4-L5 levels by cutting through the mid-vertebrae with an industrial bone saw (n = 10; Model no: MS-72ALV, General Slicing Red Goat Disposers, Murfreesboro, TN). Samples were potted in polymethylmethacrylate bone cement (PMMA, Bosworth Co., Skokie, IL) to ensure parallel loading surfaces for mechanical testing. Prior to potting, five screws were inserted into the superior and inferior vertebral bodies to improve attachment between the vertebral body and bone cement. Specimens were wrapped with saline-soaked gauze and stored in plastic bags at -20°C until testing. Before testing, samples were hydrated in 0.15 M phosphate buffer solution (0.15 M PBS) at 4º C for 24 hours and allowed to equilibrate to room temperature.

Mechanical testing

Potted specimens were mounted onto a servohydraulic materials-testing machine (Bionix 858, MTS Corp.) consisting of a custom-built bath that allowed specimens to remain hydrated during testing (0.15 M PBS). Specimens were secured in place with evenly spaced screws that attached the grips to the bone cement (30º spacing). Each test consisted of an axial compression preload followed by axial rotation applied through the disc's geometric center. First, axial compression (300, 600, 900 and 1200 N) was applied at a rate of 20 N/sec and held for 2 hours to allow for creep deformations. The range of compressive preloads was selected to represent low to moderate physiological stresses.[26] Following creep, ten cycles of cyclic rotation (haversine function; ± 2º at 0.05 Hz) were applied based on values reported for moderate *in vivo* rotations.[17] Each sample was tested under four axial compressive preloads, applied in a random order, with full recovery between tests.[27] Following intact motion segment testing, facet joints were removed with the bone saw and samples were retested under the same loading conditions. Force, displacement, rotation angle, and torque were recorded during each test.

Disc Geometry

After mechanical testing, each disc was isolated from the vertebral bodies using a scalpel. The axial plane was imaged with a digital camera to measure disc area, anterior-posterior width, and lateral width using a custom-written algorithm in Matlab (Mathworks, Inc.).[28] Disc height was measured with digital calipers and reported as an average of measurements taken from five locations: posterior, anterior, left and right lateral sides, and the center of the disc. The applied axial stress was calculated as the compressive load divided by cross-sectional area. Digital images were graded using the Thompson scale,[29] and results of macroscopic grading were compared with radiographic-based grades. Pfirrmann and Thompson grading was performed by a trained orthopaedic surgeon.

Data Analysis

Disc height loss following two hours of creep was measured, and disc geometry was used to compute the axial compressive and creep moduli. Axial compressive modulus was defined as the slope of the stress-strain response in the linear-region during loading. Creep modulus was calculated by dividing the applied stress by disc height loss at the end of the 2-hour hold.

The last cycle of axial rotation was used to calculate torsional mechanical properties. Stiffness was calculated as the slope of the torque-rotation curve, where the toe-region stiffness was calculated between 0° - 0.5° and the linear-region stiffness was calculated between 1.5° - 2°. Torque range was calculated as the difference between torques measured at $+2^{\circ}$ and -2°. Strain energy (U) was calculated as the area under the loading curve and represents energy stored in a material during loading. Torsional hysteresis (E_H) was calculated as the area between the loading and unloading torque-rotation curves during a full cycle and represents energy dissipation. Strain and hysteresis energy measurements were used to calculate the specific damping capacity (*i.e.*,

$E_{H,L}/U$, where $E_{H,L} = E_H/2$), which describes the material's ability to absorb energy during dynamic loading. The specific damping capacity also assesses viscoelasticity, where values close to 0 represent a more solid-like behavior and values close to 1.0 indicate a more fluid-like behavior. The percent contribution of facet joints to disc joint mechanics was calculated by finding the percentage change in disc mechanics after facetectomy with respect to disc mechanics of intact disc-joints.

During axial rotation, we observed a sinusoidal response in axial displacement, u_z, with two separate and distinct amplitudes. A mathematical model was developed to describe the response using a superposition of two Fourier series (Equation 1). The unknown parameters, DR_S and DR_B, describe the ranges of the small and big peaks, respectively.

$$u_z = DR_B\left(\frac{1}{4} - \frac{1}{4}cos\left(\frac{2\pi t}{T}\right) + \sum_{n=1}^{\infty} b_{2n-1}\, sin\left(\frac{\pi t(2n-1)}{T}\right)\right) + DR_S\left(\frac{1}{4} - \frac{1}{4}cos\left(\frac{2\pi(t-T)}{T}\right) + \sum_{n=1}^{\infty} b_{2n-1}\, sin\left(\frac{\pi(t-T)(2n-1)}{T}\right)\right), \quad (1)$$

where $b_k = \frac{4}{\pi}\frac{1}{(k-2)\cdot k\cdot(k+2)}$ for $k \geq 1$, and T is the time for half a cycle (10 sec. in this study).

Biochemical Analysis

Cylindrical tissue cores were harvested from the nucleus pulposus (NP) and annulus fibrosus (AF) using a biopsy punch (4.0 mm diameter; 6 locations/disc; Figure 1A). Samples were lyophilized for 48 hours to obtain dry weights. Dried samples were digested with proteinase-K (10 μg/μL at 65 °C for 24 hours; Sigma-Aldrich, St. Louis, MO). Glycosaminoglycan (GAG) content was measured using the dimethyl methylene blue (DMMB) assay. Collagen content was determined using the orthohydroxyproline (OHP) colorimetric assay and measurements were converted to collagen composition, assuming a OHP:collagen ratio of 1:7.5.[30] GAG and collagen contents were normalized by dry weight to account for differences in tissue swelling[31] and expressed as μg/mg of dry tissue weight (dw).

Statistical Analyses

A two-way analysis of variance (ANOVA) was performed on torsional mechanical properties with factors of axial compressive preload and facet joint condition (with or without facet joints). A one-way ANOVA was performed on biochemical composition to evaluate regional differences in the AF (*i.e.*, anterior, posterior, and lateral AF). Biochemical composition of the left and right lateral AF were averaged because no significant differences were observed between the two sides (paired t-test, $p > 0.80$). A Bonferroni post-hoc analysis was performed when significance was found using the ANOVA test. A Student's paired t-test was used to compare the biochemical compositions of the NP and AF. A Pearson's correlation was performed between torsion mechanics and biochemical properties and between facet joint contribution to torsion mechanics and biochemical composition. All statistical analyses were performed using R statistical software, and significance was assumed at $p \leq 0.05$. A moderate correlation was defined as $0.5 \leq r < 0.7$ and a strong correlation was defined for $r \geq 0.7$. All values were reported as mean ± standard deviation.

Prior to testing, power analyses were performed to determine the appropriate sample size (G*Power, power analysis inputs: power ≥ 0.80 and $\alpha = 0.05$). Separate sample size estimations were performed for each factor (preload and facet joint condition) and the interaction term for two-way ANOVA (input: effect size ≥ 0.40). Finally, a power analysis was conducted to determine the sample size required to perform correlations between torsion mechanics and tissue composition. Based on previous work on disc torsion mechanics,[22-23] strong correlations were expected and this was taken into account in sample size estimation (input: $\rho \geq 0.80$). Based on the results of all power analyses, nine discs were required to achieve the desired power and alpha level.

Results

The disc cross-sectional area was 1781 ± 285 mm^2 (range = 1352 – 2337 mm^2) and disc height was 8.6 ± 1.5 mm (range = 5.9 – 10.5 mm). There were no significant differences in degenerative grades between the Thompson and Pffirmann scales (Student's paired t-test, $p = 0.3$; Table 1). AF GAG and collagen contents did not depend on spatial location (one-way ANOVA, $p = 0.23$ for GAG and $p = 0.68$ for collagen; Figure 1B-C). Thus, AF data were pooled and average values were used for further statistical analyses (AF GAG: 125 ± 43 µg/mg dw; AF collagen: 676 ± 74 µg/mg dw). The NP GAG content (186 ± 115 µg/mg dw) was not different from AF GAG content (Student's paired t-test, $p = 0.89$); however, AF collagen content was greater than NP collagen content (252 ± 82 µg/mg dw; Student's t-test, $p < 0.001$; Figure 1C). There was a strong negative correlation between the Pfirrmann grade and NP GAG content ($p = 0.02$, $r = -0.71$), as expected.[32,33] In contrast, there were no significant correlations between disc degeneration and collagen content in the NP ($p > 0.8$) or AF ($p = 0.16$).

Combined effect of axial compressive preload and facet joints

There was a strong correlation between disc height loss during creep and axial compressive preload (F_{ax}, $\Delta h = 0.0021*F_{ax} + 1.03$ mm; $p < 0.0001$, $r = 0.84$), and this relationship was not affected by facet joint removal (paired t-test, $p = 0.1$). A four-fold increase in axial compressive load from 300 N to 1200 N resulted in a two-fold increase in disc height loss. Similarly, axial compressive modulus during loading ($E_{ax} = 0.0024*F_{ax} + 1.63$ MPa; $p < 0.0001$, $r = 0.75$) and creep modulus ($E_c = 9E\text{-}05*F_{ax} + 0.09$ MPa; $p < 0.0001$, $r = 0.61$) increased with compressive load and this response was not affected by facet joint removal ($p > 0.5$).

Torsional mechanical properties were strongly influenced by axial compressive preload and the presence of facet joints (two-way ANOVA, $p < 0.001$ for all properties; Figure 2). The torque-rotation response was nearly linear for all discs before and after facetectomy, and there were no significant differences between the toe- and linear-region stiffness (two-way ANOVA, $p = 0.10$ for intact and $p = 0.26$ with facetectomy); hence, average stiffness measurements were reported. Torsional stiffness and torque range increased linearly with compression; however, the magnitude of these properties and the rate of increase with compression depended on the presence of facet joints (Figure 2A-B). Torsional stiffness and torque range decreased by 50-60% for all compressive loads after facetectomy (*e.g.*, not dependent on the magnitude of compression; one-way ANOVA, $p > 0.2$; Figure 2A-B).

Strain energy and hysteresis energy increased linearly with axial compression and decreased with facetectomy ($p < 0.001$; Figure 2C-D). Energy absorption during rotation decreased by approximately 70% after facet joint removal (Figure 2C – blue *vs.* yellow bars; $p < 0.001$ for all groups). The rate of increase in strain and hysteresis energies with compressive load decreased after facetectomy. That is, there was 74% increase in disc-joint strain energy from 300 N to 1200 N, compared to a 62% increase in disc-only strain energy for the same range of axial compression (Figure 2C). The specific damping capacity of intact and facetectomy disc joints did not change with compression ($p > 0.05$); however, it decreased by ~0.1 after facet joint removal (0.34 ± 0.09 to 0.24 ± 0.07, Student's paired t-test, $p < 0.001$).

The superposition of two Fourier series fit well to the sinusoidal response in axial displacement ($r^2 > 0.98$; Figure 3A – solid lines). The axial displacement range decreased with an increase in axial compression, and the rate of decrease did not depend on facet joint condition (Figure 3B; *i.e.*, intact *versus* facetectomy; one-way ANOVA $p > 0.3$).

Effect of biochemical composition

Torsional mechanical properties were strongly correlated with GAG content, but the linearity of the correlation depended on the disc region (Figure 4 – left *vs*. right column). That is, the correlations between the AF GAG content and torsion mechanics were linear, while the trends between the NP GAG content and torsion mechanics were nonlinear. Therefore, a linear correlation was not performed with NP GAG content. Torsional mechanical properties were linearly correlated with AF GAG content under all compressive preloads ($r \geq 0.42$, Table 2), and there was an approximately six-fold increase in torsional mechanical properties with a three-fold increase in AF GAG content (Figure 4).

There was a strong linear correlation between percent facet contribution to hysteresis energy and AF GAG content ($E_H = -0.26*AF_{GAG}+104.5$, $p < 0.0001$, $r = -0.88$) and a moderate correlation between percent facet contribution to torque range and AF GAG content ($TR = -0.13*AF_{GAG}+78.8$, $p = 0.05$, $r = -0.64$). No other correlations were observed with respect to facet joint contribution to elastic properties (*e.g.*, torsional stiffness and strain energy) and biochemical composition ($p > 0.4$).

Discussion

The cross-ply structure of the annulus fibrosus is well suited to withstand shear stresses developed from axial rotation,[34] and the facet joints play a crucial role in restricting spinal rotation, protecting the disc from exposure to excess shear stresses and damage.[2] In this study, we characterized the role of facet joints under torsion combined with axial compression, and established a structure-function relationship between biochemical composition and torsion mechanics. Torsional mechanical properties decreased significantly with the removal of the facet joints under all compressive preloads and highly depended on GAG content. Defining relationships between GAG content, disc geometry, and loading modality (*i.e.*, compression-torsion behavior) allows for comparison across studies that use discs of varying levels of degeneration or testing protocols (*e.g.*, with or without compression).[35]

Collagen fibers are thought to be primarily responsible for absorbing AF tensile stresses that arise during compression, bending, and axial rotation, while GAGs are important for withstanding compressive stresses.[34, 36-38] Although AF fibers are under tension during axial rotation,[34] we did not observe any correlations between collagen content and torsion mechanics. Interestingly, disc torsion mechanics strongly depended on NP and AF GAG content (linear relationship with AF GAG content). Based on our findings, disc torsion mechanics are likely resistant to compositional changes during early degeneration, which is noted by a decrease in NP GAG content.[32] However, as the AF GAG content decreases with moderate degeneration, there are significant decreases in elastic and viscoelastic torsional properties. That is, we observed an 85% decrease in torsional stiffness from the healthiest disc to the most degenerated disc (based on AF GAG content), which agreed with data reported in the literature.[22,23] The decrease in torsional stiffness increases disc compliance, which has been observed as an increase in rotation range in patients with disc degeneration,[39] resulting in a greater proportion of hysteresis energy being absorbed by the facet joints.

The intervertebral disc and surrounding facet joints act together to absorb loads applied to the spinal column, and the overall contribution of facet joints depends on the loading modality. In this study, there were no significant changes in relative disc height loss, axial compressive modulus, and creep modulus after facetectomy, indicating a negligible effect of facet joints on compressive mechanics, corroborating data reported by Gardner-Morse *et al*.[40] In contrast, disc torsion mechanics were greatly influenced by facet joint removal (>50% decrease in properties), regardless of the magnitude of axial compression preload. However, the rate of increase in torsion mechanics with axial compression was higher for intact disc joints than discs without facet joints, highlighting the interaction between axial compression and facet joints on mechanical properties. The intervertebral disc alone was less viscoelastic than the intact disc joint, as observed by a ~30% decrease in specific damping capacity after facet joint removal. The findings of this study indicate that facet joints have a significant contribution to disc-joint viscoelasticity besides disc torsional stiffness as reported previously.[22-23] Therefore, design criteria for total disc replacement strategies should consider the load distribution between the artificial disc and facet joints during rotation to prevent abnormal loading of the facet joints and the increased rate of osteoarthritis.[41]

The intervertebral disc experiences complex stress distributions throughout the NP and AF. Under axial compression, the pressurized NP transfers stresses radially to the AF, resulting in tensile circumferential and axial strains[19,42] that pre-stress collagen fibers prior to rotation. During rotation, we observed a sinusoidal response in axial displacement, resulting in changes in disc height (Figure 3). This response was confirmed with a finite element model, where differences in amplitudes were due to the alternating fiber architecture in adjacent layers, causing slight differences in moment arms between the outermost lamella and the adjacent layer.[34] Changes in axial displacement during torsion shifted stresses from the AF back to the NP, where an increase in disc height during maximum rotation decreases collagen fiber stretch and increases NP pressure.[34] Thus, the architecture of the AF may act to protect collagen fibers from failure and reduce shear stresses at the AF-endplate interface, which is a common site for disc failure under large rotation angles.[5] It should be noted that while the AF structure may protect the disc from catastrophic failure, high local shear stresses may still initiate tissue remodeling towards the degenerative cascade.[43,44]

This study had some limitations. Although discs from the same spines might share some similarities, we neglected inter-spinal differences. Axial compression was maintained for two hours to allow for creep deformation (creep slope < 0.01 mm/min); however, human discs require more than 12 hours to achieve full equilibrium.[30,45] To limit creep time, we only investigated between-group differences and minimized creep effects during torsion by using a short loading protocol (<4 minutes). During torsion, the axis of rotation of each disc was aligned with its geometric center to make comparisons across discs with different degeneration grades; however, the geometric center may be different from the *in vivo* axis of rotation and degeneration is known to alter the axis of rotation.[46, 47] Biochemical correlations were performed only on disc mechanics without facet joints because we were not able to assess facet joint osteoarthritis. While the contribution of facet joints to disc-joint torsion mechanics may differ with facet joint osteoarthritis, it is difficult to decouple these changes and disc degeneration is thought to precede facet degeneration.[48]

Lastly, wherever possible, we attempted to compare our results with previously reported data. Torsional stiffness for the healthy disc (0.10 MPa/deg) in this study was in agreement with previously reported values for bovine (0.10 MPa/deg)[35], mouse (0.095 ± 0.030 MPa/deg)[49] and human (0.087 ± 0.019 MPa/deg)[49] discs when normalized by disc geometry. The maximum normalized hysteresis energy (*i.e.*, hysteresis energy divided by disc volume) agreed with predicted values obtained for bovine caudal discs from our previous study ($E_{H,measured}$ = 0.17 MPa-deg *vs.* $E_{H,predicted}$ = 0.12 MPa-deg).[35] However, the positive correlation between torsional hysteresis energy and GAG content disagreed with the negative correlation reported by Zirbel *et* *al.*,[22] and we did not observe a parabolic change in torsion mechanics with degeneration, which was likely due to differences in loading protocols (3 degrees of freedom (DOF) under combined compression-torsion *vs.* 6 DOF).[22,23] Although coupled motions involving multiple loading modalities (*i.e.*, axial rotation, lateral bending, flexion-extension) more closely represent physiological motions, disc mechanical properties associated with each loading modality might not be independent, resulting in the differences observed between studies.

In conclusion, disc torsion mechanics are greatly dependent on NP and AF GAG content, the magnitude of compressive preload, and the presence of facet joints. Loss in GAG content with degeneration reduced torsion mechanics by more than an order of magnitude, while changes in collagen content did not significantly influence disc torsion mechanics. Facet joints have a significant role in torsion mechanics through their contribution to joint viscoelasticity and stiffness, allowing the disc-joint to withstand large dynamic loads. However, the relative contribution of the facet joints to torsion mechanics did not change with respect to the magnitude of axial compression, indicating that both the disc and facet joints are capable of resisting high amounts of torque.

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Conflict of Interest

The authors certify that there is no conflict of interest related to the work presented in this manuscript.

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Figure captions

Figure 1: A) The location of tissue plugs harvested from each disc for biochemical analyses. B-C) GAG and collagen content normalized by dry weight within disc regions. * indicates significant difference ($p < 0.05$) between the average NP and AF values.

Figure 2: A) Torsional stiffness (N-m/deg), B) torque range (N-m), C) hysteresis energy (N-m-deg), and D) strain energy (N-m-deg) before and after facetectomy. All axial groups were statistically different than one another ($p < 0.05$).

Figure 3: A) Axial displacement during torsional loading and B) displacement range for both peaks for discs with intact facet joints.

Figure 4: The correlation between GAG content and torsion mechanics (representative graphs - only shown under 600 N).

Spine No	Age	Sex	Level	Pfirrmann	Thompson
1	80	M	L4-L5	III	III
2	43	M	L3-L4	IV	III
2	43	M	L4-L5	II	II
3	71	F	L3-L4	III	III
4	78	M	L3-L4	III	IV
4	78	M	L4-L5	V	V
5	80	M	L3-L4	II	III
6	77	F	L3-L4	III	V
6	77	F	L4-L5	IV	V
7	44	M	L4-L5	I	I

M: male, F: female

Table 1: Age, sex, spinal level and degeneration grades of lumbar discs

Parameter	Preload (N)			
	300	**600**	**900**	**1200**
Torsional Stiffness	0.91	0.92	0.93	0.94
Hysteresis Energy	0.78	0.88	0.74	0.42
Strain Energy	0.90	0.92	0.92	0.92
Torque Range	0.89	0.92	0.92	0.94

Table 2: Pearson's correlation coefficients for correlations between disc torsion mechanics and AF GAG content. Blue color indicates high correlations ($r \geq 0.7$).

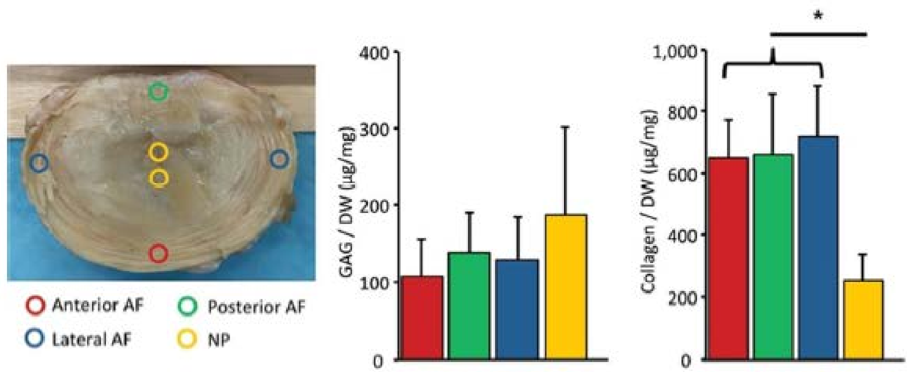
Anterior AF
Posterior AF
Lateral AF
NP
GAG / DW (µg/mg)
400
300
200
100
0
Collagen / DW (µg/mg)
1,000
800
600
400
200
0
*

Figure 1

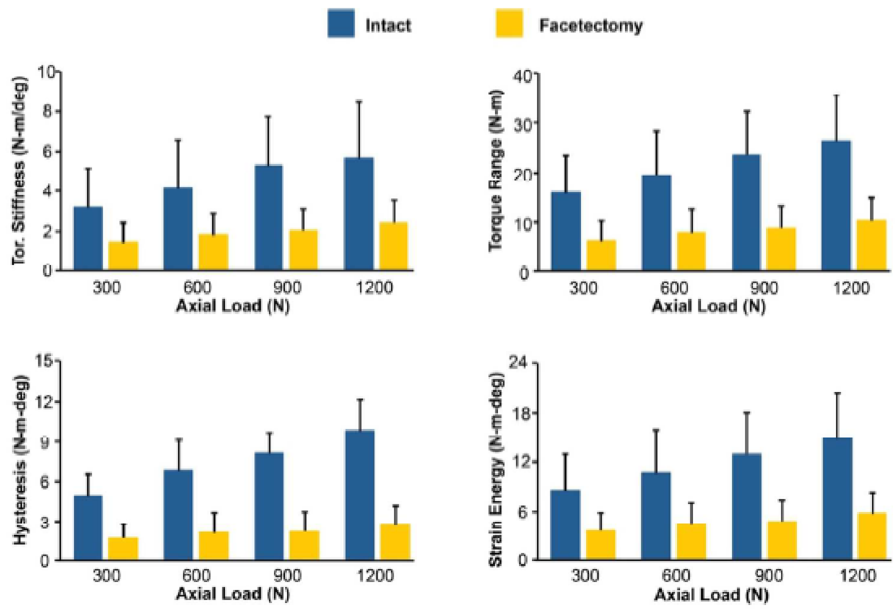
Intact
Facetectomy
Tor. Stiffness (N-m/deg)
Torque Range (N-m)
Hysteresis (N-m-deg)
Strain Energy (N-m-deg)
Axial Load (N)
300
600
900
1200

Figure 2

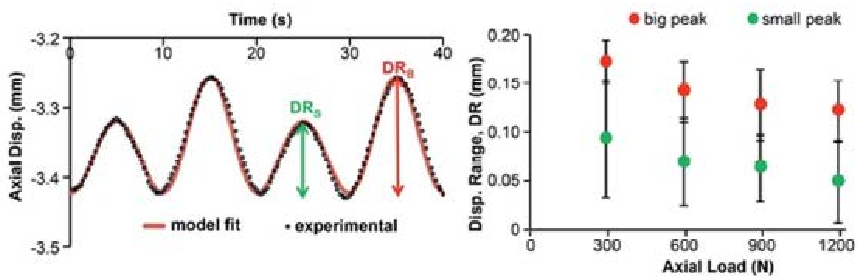
Time (s)
-3.2
-3.3
-3.4
-3.5
0
10
20
30
40
Axial Disp. (mm)
DR$_S$
DR$_B$
model fit
experimental
big peak
small peak
0.20
0.15
0.10
0.05
0
300
600
900
1200
Disp. Range, DR (mm)
Axial Load (N)

Figure 3

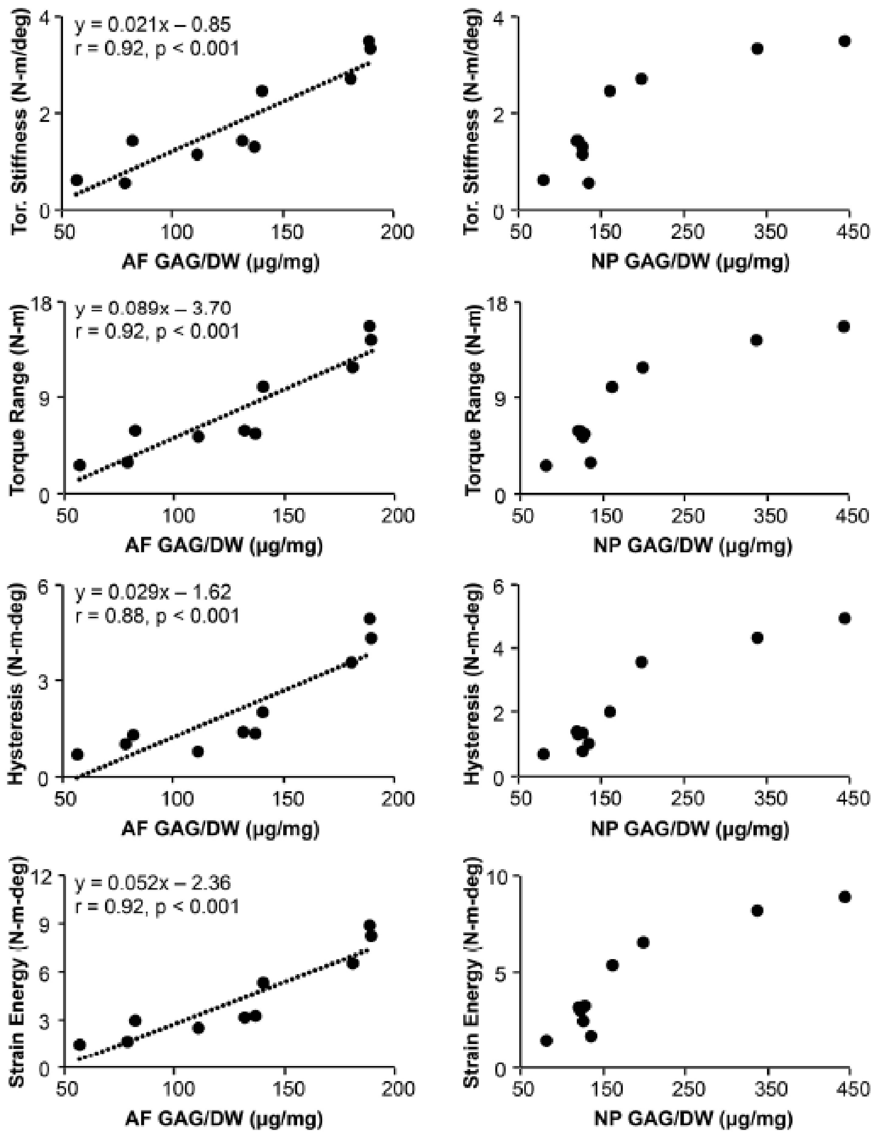
y = 0.021x – 0.85
r = 0.92, p < 0.001
Tor. Stiffness (N-m/deg)
AF GAG/DW (µg/mg)
NP GAG/DW (µg/mg)
y = 0.089x – 3.70
r = 0.92, p < 0.001
Torque Range (N-m)
y = 0.029x – 1.62
r = 0.88, p < 0.001
Hysteresis (N-m-deg)
y = 0.052x – 2.36
r = 0.92, p < 0.001
Strain Energy (N-m-deg)

Figure 4