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The degenerative state of the intervertebral disk independently predicts the failure of human lumbar spine to high rate loading: an experimental study

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Abstract

Background—In the elderly, 30–50% of patients report a fall event to precede the onset of vertebral fractures. The dynamic characteristics of the spine determine the peak forces on the vertebrae in a fall. However, we know little about the effect of intervertebral disk degeneration on the failure of human spines under the high loading rates associated with such falls. We hypothesized that MR estimates of disk hydration and viscoelastic properties will provide better estimates of failure strength than bone density alone.

Methods—Seventeen L1–L3 human spine segments were imaged (Magnetic Resonance imaging, Dual-energy X-ray absorptiometry), their dynamic responses quantified using pendulum based impact, and the spines tested to failure under high rate loading simulating a fall event. The spines' stiffness and damping constants were computed (Kelvin–Voigt model) with disk hydration and geometry assessed from T2 and proton density images.

Findings—Under impact, the spines exhibited a second-order underdamped response with stiffness and damping ranging (17.9–754.5)kN/m and (133.6–905.3)Ns/m respectively. Damping, but not stiffness, was negatively correlated with higher ultimate strength ($p < 0.05$). Higher bone mineral density and MR-estimated disk hydration correlated with higher ultimate strength ($p < 0.01$ for both). No such correlations were observed for the T2 values. Adding disk hydration yielded a 20% increase in the model's association with failure load compared to bone density alone (MANOVA, $p < 0.001$).

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Interpretation—The strong correlation between disk viscoelastic properties and MR-estimated hydration with the spine segments ultimate strength clearly demonstrate the need to include disk degeneration as part of fracture risk assessment in the elderly spine.

Keywords

Intervertebral disk; Experimental study; Dynamic characteristics; Magnetic resonance imaging

Introduction

In the USA, an estimated 1.64 million elderly adults were treated for accidental falls in hospital emergency departments during 2001 [1]. Fractures were the most reported injury, representing 37.3% for both genders [2]. This proportion was reported to further increase by 10.5% between the years 2001 and 2008, once it adjusted for the aging population during this period. The vertebrae represent the second most prevalent site for fracture [3–5] with a high proportion (30–50%) of patients reporting a fall event to precede the onset of the spinal fracture [4, 6]. The associated pain and the considerable physical and psychological discomfort result in functional impairments in up to 30% of these patients [4, 7, 8]. Reductions in mobility, and extended bed rest increase the risk for severe complications, including pneumonia, deep vein thrombosis and decubitus ulcers. These can lead to substantial morbidity and mortality.

At present, prediction of vertebral fracture risk in the elderly relies on X-ray methods such as two dimensional (dual-energy X-ray absorptiometry (DXA) and Quantitative computed tomography (QCT). DXA provides a noninvasive estimate of bone mineral density (BMD) as a surrogate measure for the mechanical competence of the vertebrae [9–12]. QCT, provides volumetric assessment of the cancellous and cortical bone compartments of the vertebrae [13, 14] and offers improved prediction of vertebral fracture risk [15]. However, the biomechanical support for both methods pertains only to quasi-static loading conditions [16–18]. Critically, the biomechanical models do not account for the effect of the nonlinear stiffness and damping properties of the spine, largely determined by the intervertebral disk [23–28], on the magnitude of forces likely to act on the vertebral bone in the event of a fall [29, 30].

Within the functional spinal unit (FSU), the intervertebral disk joint confers mobility and provides static and dynamic load attenuation to the spinal column [31–38]. In response to time varying compressive loading [39–41], the disk's visco-elastic response exhibits time constants on the order of hours [27], with peak transmission and damping ratios for human thoraco-lumbar FSU's reported to range from 23.5 to 33 Hz and 0.05 to 0.13 respectively for a body mass of 40 kg [26]. This corresponds to a single FSU stiffness of 872 to 1720 kN/m and damping constant of 590 to 2156 Ns/m. Rostedt *et al.* [42] found stiffness values to range between 1400 and 2800 kN/m with the stiffness increasing with preload. These values are much higher than the instantaneous stiffness found in creep studies [27], suggesting the rate of loading to be an important contributor to FSU stiffness. Employing an optimization based geometric model of osteopenic thoraco-lumbar spine, Wilson *et al.*, [43], predicted the stiffness and damping characteristics of the spinal segments to influence the peak forces

sustained by the vertebrae due to a backwards fall from standing height. Assuming the subject would land in a seated position, such a fall was modeled to produce a high rate axial load and thus would produce a compression fracture. Although little is known about the magnitude and type of forces or the moments to which the thoraco-lumbar spine is exposed due to a backward fall [44, 45], the model predicted a 10% increase in the stiffness of the segment to yield an increase of 2–5% in the peak forces acting on the vertebrae [43]. This increase in peak forces, may increase the risk of vertebral fracture (VFX) in osteoporotic individuals. Thus, assessment of the disk dynamic properties would be important for quantifying the risk of fall-associated VFX in the elderly.

T2 weighted magnetic resonance (MR) images provide assessments of anatomical features and an inferred hydration state of the disk [46–48] while T1 and T2 relaxation rates demonstrate inverse relationships with disk water [49, 50] and glycosaminoglycan content [51]. Degenerative changes in the annulus and nucleus, which strongly affect their static and viscoelastic material properties [52–54] and the associated loss of hydration [55–57], result in the degradation of the FSU's long-term creep behavior and energy dissipation [24, 28, 57]. Although hydration and the degenerative state of the disk are important determinants of the spine's stiffness and damping characteristics, little is known about their effect on the failure load of elderly spines under high rate loading [29, 44, 58]. In view of the intimate relationship between the disk's hydration and its mechanical competence, we hypothesize that MR classification of degenerative changes will be associated with loss of the dynamic stiffness and damping properties of the spinal segments. We further hypothesize that the MR measures, T2 and estimated water content, combined with measures of bone density, will provide better estimates of failure strength than bone density alone.

Methods

Specimens

Five male and 14 female, L1 to L3 cadaveric spine segments were obtained fresh-frozen from donors aged 62 to 85 years of age (mean (Standard deviation) 75.5 (6.9) years) through the Harvard Anatomic Gifts program. Each segment was radiographed (Faxitron, HP, McMinnville, OR) to exclude existing pathology or fractures, the spine submerged in a saline bath to simulate soft tissues and bone mineral density (BMD) measured for each vertebra using a DXA scanner (QDR 2000+, Hologic Inc., Waltham, MA). The segments were dissected clean of all musculature, with care taken to leave the posterior structures and ligaments intact, wrapped in saline soaked gauze in plastic bags and then stored at -20°C .

Magnetic resonance imaging

The spines were allowed to thaw wrapped in saline gauze at 4°C for approximately 8 hours and then vacuum degassed for a period of eight hours in polyethylene tubes filled with saline at 20°C . A GE 1.5 Tesla MR scanner (General Electric Medical Systems, Waukesha, WI) was used to obtain sagittal T2-weighted (TR/TE:2000/80)ms images and three sets of axial images (TR/TE: 2000/80, 2000/60, 2000/20)ms with a field of view of 150mm and a matrix size of 128×128 . The first and last axial image represented T2 weighted and proton weighted images respectively. A series of six phantoms, with known volume fractions of heavy water

(D₂O) and saline, placed under the tubes during imaging, provided estimation of the hydration state of the disks. At the end of the imaging session, the segments were removed and returned to the freezer until the day of testing. This was necessary because of the time constraints of both the imaging and testing.

A custom program written in AVS (V5.0, Advanced Visualization Systems, Waltham, MA) was used first to segment the disk from the axial T2 images and then to compute cross sectional area and average intensity. Based on the geometry and intensity values, the nucleus and annulus were manually segmented and their cross sectional areas computed. The sagittal, mid-disk images were used to measure the disk's anterior central and posterior heights, computed as the shortest distance between the corresponding vertebral bone endplates, and disk height computed as the mean value of the three measurements [59]. From the proton density weighted images, the relative intensity of disk to phantom intensity was computed and the hydration level (percent H₂O) quantified using a linear regression model.

Characterization of dynamic properties

A custom impact tester (Figure 1), employing a pendulum to impart impact based loading, was used to characterize the dynamic response of the FSU. A 6-degree of freedom load cell (MC5-5000, resonance frequency of 1250 Hz, AMTI, Watertown, MA), secured to the stationary base of the pendulum, was used to measure the transmitted force resulting from the impact (Figure 1). Prior to testing, each FSU was thawed for six hours at 4 C followed by a two hour immersion in saline at 37 C under constant compression (178)N to re-hydrate the disks. This load, within the range of compressive loads (150–250) N suggested to occur at the lumbar spine for a recumbent position [60], represented the upper body mass of the lightest specimen and was chosen to avoid the potential of fracturing the more osteoporotic spines by application of the constant load. Using an alignment jig, the L1 and L3 vertebrae were embedded in polymethylmethacrylate cement (Fastray, Bosworth, Skokie, IL) with the FSU wrapped in plastic to prevent dehydration. The embedded spine was secured to the impact apparatus (Figure 1) and a pre-load (112±14N) applied using seven-inch rubber bands. This method was used to allow pre-loading while providing for minimum restraint of spinal kinematics during the test. The pendulum arm, having an end mass of 20.9 kg was positioned to deliver an impact with a kinetic energy of 2.50J. Based on our preliminary studies, this value allowed for quantification of the dynamic response of the segment under repeated tests without causing fracture of the vertebral structures [58]. The FSU was tested under impact with the force channel sampled at a rate of 3000Hz (V.8.0 FDS, Labview, National Instruments, TX). To prevent the occurrence of a secondary impact, the pendulum was caught immediately after the impact using a brake triggered by the load cell and controlled by LabView. registered as a smaller secondary peak prior to the underdamped response (Fig 3).

Mechanical test to failure

Figure 2 presents a diagrammatic illustration of the high rate test set up. The test procedure consisted of the following sequential steps.

- a. *Identification of the FSU's kinematic center of rotation (KCR, Fig 2):* A compressive load of 100N was imposed on the FSU and the lower X-Y table adjusted until the 6DOF output for the lateral and forward bending moments was at a minimum. This location, representing the FSU KCR, allowed normalizing of the test boundary conditions between the different FSU's.
- b. *High rate test:* Under force control, the FSU was conditioned by the application of 50 sinusoidal cycles of compressive loading (50 to 200 ($R=125$))N, at a rate of 1Hz. Under displacement control, the FSU was subjected to a monotonically increasing compressive load at a rate of $0.4\text{m}\cdot\text{sec}^{-1}$ until either of the following criteria was met: i) a 50% target reduction in the mean height of the vertebral body or ii) the load cell safe capacity (10000N) was reached. Throughout the test, the applied axial displacement and the 6DOF force and moment output, were recorded at a rate of 10,000Hz (V.8.0 FDS, Labview, National Instruments, TX). Custom code [61], written in Labview, was used to correct for the overestimation of the measured moments due to the difference between shear center of the load cell ($D_{LCS-VCS}$, Fig. 2) and the coordinate system of interest located at the geometrical center of the vertebral body (V_{CS} , Fig. 2).

Data analysis

Dynamic disk response—In MATLAB (V.12. Mathworks, Natick, MA), fast Fourier transform (FFT) was applied to the force signal from the time where the pendulum left the end mass to obtain the peak frequency (Figure 3). A finite impulse response (FIR) low pass filter, with Hamming windowing and a cutoff frequency of twice the peak frequency, was applied to filter the signal. The damping factor, ζ , was computed by fitting a linear model to the logarithms of several successive peaks of the response (Appendix A). The dynamic stiffness (K) and damping (B) parameters could then be computed for the Voight model using an end mass of 20.9 kg (Appendix A)..

High rate mechanical tests—For each FSU, the force-displacement response to failure was plotted and the specimen's ultimate failure strength measured (Figure 4).

To test the hypothesis that disk degeneration, as assessed with MR, would alter the viscoelastic properties of the spine, linear regression was used to relate the T2 values, computed from the three T2 images, to the dynamic stiffness and damping coefficients. To test the hypothesis that disk degeneration would alter the ultimate strength of the vertebrae, linear regression was used to determine the correlation of each of the following parameters with the FSU's failure strength: disk geometry: (height (mm), area (mm^2)), MRI: (T2: (ms); normalized PD); dynamic properties: (stiffness (kN/m), damping (Ns/m)) and vertebral BMD: (A-P and lateral scans (g/cm^2)). Multivariate regression was used to establish a predictive model from the variables found to be significant in the univariate analysis. Analysis of covariance was used to establish the independent contribution of each of the variables chosen. Statistical significance was set at 5%.

Results

For the L1 to L3 vertebrae, BMD were found to have a (mean (standard deviation)) of 0.72 (0.19) g/cm² for the AP scan and 0.32(0.13) g/cm² for the lateral scan, representing a range consistent with that of osteopenic vertebrae [62]. Higher T2 relaxation values were significantly correlated with the increase in the disk's cross-sectional area ($r^2=0.24$, $p<0.05$) and height ($r^2=0.35$, $p<0.01$). T2 weighted values correlated with proton density (PD) measures ($r^2=0.30$, $p<0.05$). No statistically significant correlations were found between age, gender or BMD values and either the MR (T2, T1) or disk geometry (height, cross-sectional area) parameters, $p>0.05$.

Dynamic characterization

Under applied impact, the FSU's showed a second-order, under-damped, oscillations (Figure 3). From the viscoelastic model, the disks were found to exhibit a mean (SD) of 293.4 (157.6) kN/m for dynamic stiffness and 376.7 (159.9) Ns/m for the damping coefficients. Table 1 presents the results of the regression analysis for the association between the disk's dynamic stiffness and damping coefficient and its geometry and MR measures. Higher damping coefficient was significantly associated with the decrease in PD values ($r^2=0.44$, $p<0.01$). No statistically significant correlations were found between age, gender or BMD values and dynamic properties of the disk, $p>0.05$.

Mechanical test to failure

Figure 4 illustrates a displacement-force response of L1–L3 FSU under the applied high rate loading. The FSU exhibited a nearly linear force response under increased deformation, culminating with frank failure at 18% strain, as computed for the whole FSU. Mirroring the impact test, the failure was followed by a series of second-order under-damped oscillations with a rapid decrease in the force response post-failure. Physically, the vertebrae failed via either a crush or a burst type fracture that occurred at the middle vertebra (L2). Figure 5, showing a fluoroscopic image of a motion segment before and after the test, demonstrate the resulting extensive damage to the vertebra and disks. Ranging from 2168.1N to 9222.0N with a mean (SD) of 4795.1 (1818.5) N, the FSU's ultimate strength was significantly correlated with the decrease in the damping coefficient of the disks ($r^2 = 0.28$, $p<0.05$, Figure 6.B) and with the increase in A-P measurement of vertebral BMD ($r^2 = 0.46$, $p<0.01$, Figure 7.B). No statistically significant correlation was found between the ultimate strength and either the dynamic stiffness of the disk (Figure 6.A), or the disk's geometrical properties (Table 1), $p>0.05$.

MR parameters vs. Mechanical response

The increase in the FSU's ultimate strength was significantly correlated with higher PD values (whole disk: $r^2=0.43$, $p<0.01$ (Figure 7.A)). T2 relaxation values were not correlated with the FSU ultimate strength, $p>0.05$. Addition of the PD term as a main effect to the predictive model, FSU(Ultimate Strength) = $-65043 + 6194.3 \times (\text{BMD(A-P)} + 9822.1 \times \text{PD}$, resulted in the R^2 value increasing from 46% for BMD only to 69%, the increase being highly significant (MANOVA, $p<0.001$). Analysis of covariance (ANCOVA) demonstrated no statistically significant interaction between the two predictor variables ($p>0.05$).

Discussion

Vertebral fractures, representing the second most prevalent site for fragility based fractures [3–5], cause considerable physical and psychological discomfort resulting in functional impairments in up to 30% of this patient population [4, 7, 8]. As the population ages, the burden of fractures and resulting complications will increase. Accurately predicting fracture risk in individuals will improve patient management and strengthen prevention strategies. This study is the first to investigate the role of intervertebral disk mechanics in the failure load of human spines under high rate loading, simulating a fall. With MR being the gold standard for imaging degenerative disk disease, the study further ascertained whether MR findings could be independent predictors of the failure load.

The stiffness and damping characteristics of the intervertebral disk form a critical component in determining the spine's dynamic response [27, 39, 44, 63, 64]. In this study, the dynamic stiffness computed for the lumbar spine (having two disks in series) can be compared to previous experiments using a single functional spinal unit by doubling the stiffness and damping values measured in this study. Ranging from 17.9 to 755 kN/m, our results are in agreement with that reported for lumbar FSU under cyclic loading [26] and with the observed effect of preload on the stiffness [42]. However, these stiffness values are higher than the “instantaneous” stiffness values reported for creep response of spine [27] whilst being generally lower than those reported for axial stiffness of spine segments [42]. Apart from the normal variation expected due to differences inherent to biological cadaveric samples, these differences might reflect both the dissimilarity in test conditions, impact vs. cyclic or creep test, as well as the age of our sample population.

Disk degeneration, including compositional [65] and physiological [53] modification of the tissue, results in the degradation of static, dynamic and viscoelastic properties [27, 37, 54, 57, 65–69]. The associated increase in the ratio of instantaneous (E1) to long term (E2) stiffness of the disk [27] was predicted to amplify the compressive loads on the vertebrae yielding an increased risk of vertebral fracture in the event of a fall from standing height [58]. The computational model further predicted a one standard deviation increase in stiffness (163 kN/m) to result in up to an 18% increase in axial compressive loading. With elderly cadaveric vertebrae reported to have an average axial failure force of approximately 2000N [70], such an increase can significantly increase the risk of vertebral fracture in a fall event. The results of this study, however, found no statistically significant correlation between the measured dynamic stiffness of the FSU's and their ultimate strength (Figure 7). Computational [71, 72] and experimental [73] studies suggested that injury to the annulus modifies the dynamic response of the spine. However, the large viscoelastic time constants of the annulus [27] and observed small deformation of the segment prior to failure (Figure 6), suggest that tissue deformation based mechanisms have little effect on the high rate response of the FSU.

The increase in the MR-estimated hydration of the disk (Figure 7.B) and, the decrease in the disk's viscoelastic damping coefficient (Figure 6.B), were significantly correlated with the increase in the FSU's ultimate strength. We found the expected importance of BMD to the ability of the FSU to sustain high rate loading (Figure 6A). However, the MANOVA analysis

demonstrated disk hydration to be a significant, independent, contribution to the prediction of the failure load. These findings, in agreement with the reported effects of strain [74] and loading frequency [68, 75] on the permeability of the disk, suggest that the interaction of water with the macromolecular environment of the disk has a significant role in the response of the spine to high rate loading. In response to an impact, a finite element study by Lee et al., [21], predicted the constant pore pressure within the nucleus, rather than the solid matrix of the disk, to support the compressive load on the FSU during its initial response (up to 200msc). With the annulus restricting the motion of fluid in the axial plane of the disk, the model predicted the high fluid flow through the endplates. This was expected to result in development of high pressure within the vertebral cancellous bone and to result in endplate and vertebral cortex fractures. The findings of our study provide, for the first time, experimental support to the computational model predictions. The intervertebral disk has important contributions to the static [31, 32, 36, 44] and dynamic response [69, 76] of the human spine. Our study suggests that future models for the prediction of fracture risk of the elderly, should include characterization of the mechanical and hydration status of the disk.

Magnetic resonance imaging (MRI) is the current diagnostic tool of choice for clinical evaluation of spinal pathology [47, 51, 77, 78]. However, the result of this study did not support the hypothesis that T2 based MR classification of degeneration is associated with the loss of dynamic stiffness and damping properties of the spinal segments (Table 1). It is possible that this reflected in part a restriction of range, with our specimens all derived from elderly spines, in which there would be a high prevalence of disk degeneration. An inclusion of spines from younger donors, with a wider range of T2 values and degeneration, might have revealed relationships between T2 and viscoelastic properties. However, we believe our focus on aged spines is appropriate to the questions at hand. The observed lack of correlation could have been the result of the MRI measurements being done with low preload level, known to effect the dynamic response of the FSU [72, 73]. However, this scanning simulates a clinical MRI assessment typically done in a supine posture during which there is a low compressive load on the spine. In agreement with the study by Chiu *et al.*, [79], the results of this study suggest that current clinical MR based classification of degenerative changes may not reflect the upright, loaded characteristics of the spine and thus cannot be used to assess the loss of dynamic stiffness and damping properties of the spinal segments.

This study suffers from the obvious limitations faced when attempting to translate an *in vitro* study to the clinical setting. The temperature may have influenced the results. Although Panjabi [80] found that biomechanical properties of the spine did not change with freezing, the specimens in this study were forced to go through several cycles, including one between imaging and testing which may have had some effect on the properties. This study was also performed at room temperature rather than at body temperature. Little is known about the effects of temperature on the viscoelastic behavior of the intervertebral joint, but it is possible that temperature could play a role. Though we have attempted to standardize the testing methodology, little knowledge exists about the loading conditions that occur on the spine due to a fall event or the contribution of the muscular system to the stability and load attention of the spine. During a fall, it is likely that both the geometrical and kinematical configuration of the spinal column will vary considerably from the pure axial alignment we employed, resulting in a complex set of loading modes [58]. Hence, the axial compression

test, represents the simplest mode of loading that may act on the spine due to a fall event. For the analysis of the dynamic response of the disk, we have used a Voight based linear model to compute the stiffness and damping parameters for the disk. Although the spine demonstrates non-linear response under dynamic and time dependent load conditions [23, 25–27], the impact test performed in this study employed only small oscillations at each preload allowing us to consider a linearized model of the frequency-load (F-L) relationships. This methodology, previously used by several authors to characterize the mechanics of impact on the hip [81], the thoraco-lumbar [43, 58] and lumbosacral spine, in effect provides the slope of the F-L curve at each preload. When extrapolating to larger motions, the non-linearity for the spring and dashpot elements as a function of the applied load must be included. This could be done by putting together the slopes at each preload into a nonlinear F-L function.

We based our attempt to estimate the effect of water content on routine clinical MR acquisitions. These frequently include so-called proton density weighted images, but not more elaborate and accurate attempts to determine water content. Given the TR we employed and the reported relaxation rates of the disk, we expect that our PD measure also contained T1 information. Removing this would have required acquisitions with far longer TR, prolonging data collection and moving the images farther from those obtained under clinical conditions. Nonetheless, our estimates of PD were highly significant in uni- and multivariate associations with failure strength

In conclusion, degenerative changes in the intervertebral disks, indicated on MR imaging as a decreased T2, were weakly correlated with stiffness. MR estimates of disk hydration were significantly correlated with the viscoelastic and ultimate strength of the FSU, suggesting a fluid flow mechanism within the disk and vertebrae to be an important determinant of the ability of the spine to sustain high rate loading. The strong correlation between MR based assessment of disk hydration and the FSU's ultimate strength (Figure 7), suggests that MR estimates of hydration may be incorporated along with BMD into models that predict vertebral fracture risk. Although BMD alone provides useful estimates of this risk, in some circumstances, a more precise assessment may be required, and our results suggest that the addition of MR may be helpful.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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References

1. Stevens JA, Sogolow ED. Gender differences for non-fatal unintentional fall related injuries among older adults. *Inj Prev*. 2005; 11(2):115–9. [PubMed: 15805442]

2. Orces CH. Emergency department visits for fall-related fractures among older adults in the USA: a retrospective cross sectional analysis of the National Electronic Injury Surveillance System All Injury Program, 2001–2008. *BMJ Open*. 2013; 3(1):e001722.
3. Taylor AJ, et al. Clinical and demographic factors associated with fractures among older Americans. *Osteoporos Int*. 2011; 22(4):1263–74. [PubMed: 20559818]
4. Cooper C, et al. Incidence of clinically diagnosed vertebral fractures: a population-based study in Rochester, Minnesota, 1985–1989. *J Bone Miner Res*. 1992; 7(2):221–227.
5. Melton LJ, et al. Epidemiology of vertebral fractures in women. *Am J Epidemiology*. 1989; 129(5): 1000–1011.
6. Myers ER, Wilson SE. Biomechanics of osteoporosis and vertebral fracture. *Spine*. 1997; 22(24): 25S–31S. [PubMed: 9431641]
7. Gold DT. The clinical impact of vertebral fractures: quality of life in women with osteoporosis. *Bone*. 1996; 18(3 Suppl):185S–189S. [PubMed: 8777086]
8. Lyles KW, et al. Association of osteoporotic vertebral compression fractures with impaired functional status. *Am J Med*. 1993; 94(6):595–601. [PubMed: 8506884]
9. Lochmüller EM, et al. Mechanical strength of the thoracolumbar spine in the elderly: prediction from in situ dual-energy X-ray absorptiometry, quantitative computed tomography (QCT), upper and lower limb peripheral QCT, and quantitative ultrasound. *Bone*. 2002; 31(1):77–84. [PubMed: 12110416]
10. Bürklein D, et al. Correlation of thoracic and lumbar vertebral failure loads with in situ vs. ex situ dual energy X-ray absorptiometry. *J Biomech*. 2001; 34(5):579–87. [PubMed: 11311698]
11. Duboeuf F, et al. Assessment of vertebral fracture using densitometric morphometry. *J Clin Densitom*. 2005; 8(3):362–8. [PubMed: 16055969]
12. Lewiecki EM. Bone Densitometry and Vertebral Fracture Assessment. *Curr Osteoporos Rep*. 2010; 8:123–130. [PubMed: 20556673]
13. Engelke K, et al. Advanced CT based In Vivo Methods for the Assessment of Bone Density, Structure, and Strength. *Curr Osteoporos Rep*. 2013 May 28. (Epub).
14. Adams JE. Quantitative computed tomography. *Eur J Radiol*. 2009; 71(3):415–24. [PubMed: 19682815]
15. Wang XA, et al. Prediction of new clinical vertebral fractures in elderly men using finite element analysis of CT scans. *J Bone Miner Res*. 2012; 27(4):808–16. [PubMed: 22190331]
16. Kopperdahl DL, Morgan EF, Keaveny TM. Quantitative computed tomography estimates of the mechanical properties of human vertebral trabecular bone. *J Orthop Res*. 2002; 20(4):801–5. [PubMed: 12168670]
17. Helgason B, et al. Mathematical relationships between bone density and mechanical properties: a literature review. *Clin Biomech*. 2008; 23(2):135–46.
18. Crawford RP, Cann CE, Keaveny TM. Finite element models predict in vitro vertebral body compressive strength better than quantitative computed tomography. *Bone*. 2003; 33(4):744–50. [PubMed: 14555280]
19. Ochia RS, Tencer AF, Ching RP. Effect of loading rate on endplate and vertebral body strength in human lumbar vertebrae. *J Biomech*. 2003; 36(12):1875–81. [PubMed: 14614941]
20. Ochia RS, Ching RP. Rate dependence of hydraulic resistance in human lumbar vertebral bodies. *Spine*. 2006; 31(22):2569–74. [PubMed: 17047546]
21. Lee CK, et al. Impact response of the intervertebral disc in a finite-element model. *Spine*. 2000; 25(19)
22. Ochoa JA, et al. In vivo observations of hydraulic stiffening in the canine femoral head. *J Biomech Eng*. 1997; 119(1):103–8. [PubMed: 9083856]
23. Kasra M, Grynblas MD. Static and dynamic finite element analyses of an idealized structural model of vertebral trabecular bone. *J Biomech Eng*. 1998; 120(2):267–72. [PubMed: 10412389]
24. Laible JP, et al. A poroelastic-Swelling finite element model with application to the intervertebral disc. *Spine*. 1993; 18(5):659–670. [PubMed: 8484158]
25. Kalpes I, Kazarian LE, Burns ML. Analysis of compressive creep behavior of the intervertebral unit subjected to uniform axial loading using exact parametric solution equations of kelvin-solid

- models-Part I. Rhesus monkey intervertebral joints. *Journal of Biomechanics*. 1984; 17:131–136. [PubMed: 6725292]
26. Kasra M, Shirazi-Adl A, Drouin G. Dynamics of human lumbar intervertebral joints. Experimental and finite-element investigations. *Spine*. 1992; 17(1):93–102. [PubMed: 1536019]
 27. Keller TS, Spengler DM. Mechanical behavior of the human lumbar spine. I. Creep analysis during static compressive loading. *J Orthop Res*. 1987; 5(4):467–478. [PubMed: 3681521]
 28. Lai WM, How J, Mow VC. A triphasic theory for the swelling and deformation behavior of cartilage tissue. *Journal of Biomechanical Engineering*. 1991; 113:145–158.
 29. Sran MM, Robinovitch SN. Preventing Fall-Related Vertebral Fractures Effect of Floor Stiffness on Peak Impact Forces During Backward Falls. *Spine*. 2008; 33(17):1856–1862. [PubMed: 18670338]
 30. Kroonenberg, AJvd, Hayes, WC., McMahon, TA. Hip impact velocities and body configurations for voluntary falls from standing height. *J Biomech*. 1996; 29:807–11. [PubMed: 9147979]
 31. Adams MA, Hutton WC. The effect of posture on the fluid content of lumbar intervertebral discs. *Spine*. 1983; 8:665–671. [PubMed: 6685921]
 32. Adams MA, et al. Sustained loading generates stress concentrations in lumbar intervertebral discs. *Spine*. 1996; 21(4):434–438. [PubMed: 8658246]
 33. Bodine AJ, et al. Viscoelastic shear modulus of the human intervertebral disc. *Trans Orthop Res Soc*. 1982
 34. Botsford DJ, Esses SI, Ogilvie-Harris DJ. In vivo diurnal variation in intervertebral disc volume and morphology. *Spine*. 1994; 19:935–940. [PubMed: 8009352]
 35. Brinckmann P, et al. Deformation of the end-plate under axial loading of the spine. *Spine*. 1983; 8:851–856. [PubMed: 6670020]
 36. Brown T, Hansen RJ, Yorra AJ. Some mechanical test on the lumbosacral spine with particular reference to the intervertebral discs. *Journal of Bone and Joint Surgery*. 1957; 39A(7):1135–1164.
 37. Burns ML, Kalpes I, Kazarian LE. Analysis of compressive creep behaviour of the intervertebral unit subjected to uniform axial loading using exact parametric solution equations of kelvin-solid models-Part I. Human intervertebral joints. *Journal of Biomechanics*. 1984; 17:113–130. [PubMed: 6725291]
 38. Chinn, H., et al. The effect of posture on stress distribution inside lumbar intervertebral discs. 38th Annual Meeting Orthopaedic Research Society; 1992.
 39. Kong WZ, Goel VK. Ability of the finite element models to predict response of the human spine to sinusoidal vertical vibration. *Spine*. 2003; 28(17):1961–7. [PubMed: 12973142]
 40. Guo LX, Teo EC. Prediction of the modal characteristics of the human spine at resonant frequency using finite element models. *Proc Inst Mech Eng H*. 2005; 219(4):277–84. [PubMed: 16050218]
 41. Guo LX, et al. Vibration modes of injured spine at resonant frequencies under vertical vibration. *Spine*. 2009; 34(19):E682–8. [PubMed: 19730200]
 42. Rostedt M, et al. Axial stiffness of human lumbar motion segments, force dependence. *J Biomech*. 1998; 31(6):503–509. [PubMed: 9755034]
 43. Wilson, SE., Myers, ER. Estimation of forces on the spine during a fall using lumped parameter modeling. 6th Annual Symposium on Computational Methods in Orthopaedic Biomechanics; 1998; Atlanta, Georgia.
 44. Toen CV, et al. Transmission of force in the lumbosacral spine during backward falls. *Spine*. 2012; 37(9):E519–27. [PubMed: 22076645]
 45. Leucht P, et al. Epidemiology of traumatic spine fractures. *Injury*. 2009; 40(2):166–72. [PubMed: 19233356]
 46. Boss, N., et al. Tissue characterization of normal and herniated intervertebral discs by quantitative MRI. 39th Annual Meeting; Orthopaedic Research Society; 1993. p. 417
 47. Boos N, et al. Quantitative MR imaging of lumbar intervertebral disks and vertebral bodies: influence of diurnal water content variations. *Radiology*. 1993; 188(2):351–4. [PubMed: 8327677]
 48. Terti M, et al. Disc degeneration in magnetic resonance imaging. A comparative biochemical, histologic, and radiologic study in cadaver spines. *Spine*. 1991; 16(6):629–634. [PubMed: 1862401]

49. Fullerton, GD., Camron, IL. Relaxation of biological tissues. In: Wehrli, FH. Shaw, D., Kneeland, JB., editors. Biomedical magnetic Resonance imaging: principles, methodology, and Applications. VCH; New York: 1988. p. 115-155.
50. Crooks LE, et al. The value of relaxation times and density measurements in clinical MRI. Invest Radiol. 1987; 22(2):158–69. [PubMed: 3557889]
51. Boos N, Boesch C. Quantitative magnetic resonance imaging of the lumbar spine. Potential for investigations of water content and biochemical composition. Spine. 1995; 20 (21):2358–2365. discussion 2366. [PubMed: 8553128]
52. Ebara S, et al. Tensile properties of non degenerated human lumbar annulus fibrosus. Spine. 1996; 21(4):452–461. [PubMed: 8658249]
53. Acaroglu ER, et al. Degeneration and aging affect the tensile behavior of human lumbar annulus fibrosus. Spine. 1995; 20(24):2690–2701. [PubMed: 8747247]
54. Iatridis JC, et al. Degeneration affects the anisotropic and nonlinear behaviors of human anulus fibrosus in compression. Journal of Biomechanics. 1998; 31:535–544. [PubMed: 9755038]
55. Galante JO. Tensile properties of the human lumbar annulus fibrosus. Acta Orthopedica Scandinavia. 1967; 100(suppl):1–91.
56. Panagiotacopulos ND, et al. Mechanical model for the human intervertebral disc. Journal of Biomechanics. 1987; 20(9):839–850. [PubMed: 3680310]
57. Best BA, et al. Compressive mechanical properties of the human annulus fibrosus and their relationship to biochemical composition. Spine. 1994; 19(2):212–221. [PubMed: 8153833]
58. Wilson SE, Myers ER. Forces on the spine during a fall from standing height. Journal of Biomechanical Engineering. 2000
59. Pfirrmann CW, et al. Effect of aging and degeneration on disc volume and shape: A quantitative study in asymptomatic volunteers. J Orthop Res. 2006; 24(5):1086–94. [PubMed: 16609964]
60. Adams, MA., et al. The Biomechanics of Back Pain. Churchill Livingstone; Edinburgh, London, New York: 2002.
61. Alkalay RN, et al. The effect of cement augmentation on the geometry and structural response of recovered osteopenic vertebrae: an anterior-wedge fracture model. Spine. 2008; 33(15):1627–36. [PubMed: 18594454]
62. Kanis JA, McCloskey EV. Epidemiology of vertebral osteoporosis. Bone. 1992; 13 (2):S1–S10. [PubMed: 1385718]
63. Shirazi-Adl A. Finite element simulation of changes in the fluid content of human lumbar discs. mechanical and clinical implications. Spine. 1992; 17(2):206–212. [PubMed: 1553592]
64. Verver MM, et al. Estimation of spinal loading in vertical vibrations by numerical simulation. Clin Biomech. 2003; 18(9):800–11.
65. Bibby SR, et al. The pathophysiology of the intervertebral disc: Review. Joint Bone Spine. 2001; 68(6):537–542. [PubMed: 11808995]
66. O'Connell GD, et al. Axial creep loading and unloaded recovery of the human intervertebral disc and the effect of degeneration. J Mech Behav Biomed Mater. 2011; 4 (7):933–42. [PubMed: 21783103]
67. Pollintine P, Tunen MSv. Time-dependent compressive deformation of the ageing spine: relevance to spinal stenosis. Spine. 2010; 35(4):386–94. [PubMed: 20110846]
68. Iatridis JC, et al. The viscoelastic behavior of the non-degenerate human lumbar nucleus pulposus in shear. J Biomech. 1997; 30(10):1005–1013. [PubMed: 9391867]
69. Kasra M, Shirazi-Adl A, Drouin G. Dynamics of human lumbar intervertebral joints: Experimental and finite element investigations. Spine. 1982; 17(1):93–101.
70. Moro M, et al. Failure load of thoracic vertebrae correlates with lumbar bone mineral density measured by DXA. Calcif Tissue Int. 1995; 56(3):206–209. [PubMed: 7750025]
71. Chon TE, Li-Xin G. Material property sensitivity analysis of dynamic characteristics of human spine using finite element method. Journal of Biomechanics. 2007; 40(2):S273.
72. Wang JL, et al. Rate effect on sharing of passive lumbar motion segment under load-controlled sagittal flexion: viscoelastic finite element analysis. Theoretical and Applied Fracture Mechanics. 1999; 32:119–128.

73. Engelen, SJv, et al. The feasibility of modal testing for measurement of dynamics characteristics of goat vertebral motion segments. *J Biomech*. 2011; 28 Epub ahead of print.
74. Lee KK, Teo EC. Poroelastic Analysis of Lumbar Spinal Stability in Combined Compression and Anterior Shear. *Journal of Spinal Disorder and Technique*. 2004; 17 (15):429–438.
75. Kuo YW, Wang JL. Rheology of intervertebral disc: an ex vivo study on the effect of loading history, loading magnitude, fatigue loading, and disc degeneration. *Spine*. 2010; 35(16):E743–52. [PubMed: 20581749]
76. Kemper AR, McNally C, Duma SM. The influence of strain rate on the compressive stiffness properties of human lumbar intervertebral discs. *Biomed Sci Instrum*. 2007; 43:176–81. [PubMed: 17487077]
77. Cassar-Pullicino VN. MRI of the ageing and herniating intervertebral disc: Review. *Eur J Radiol*. 1998; 27(3):214–28. [PubMed: 9717637]
78. Haughton V. Medical imaging of intervertebral disc degeneration: current status of imaging. *Spine*. 2004; 29(23):2751–2756. [PubMed: 15564924]
79. Chiu EJ, et al. Magnetic resonance imaging measurement of relaxation and water diffusion in the human lumbar intervertebral disc under compression in vitro. *Spine*. 2001; 26(19):437–444.
80. Panjabi MM, et al. Biomechanical time-tolerance of fresh cadaveric human spine specimens. *J Orthop Res*. 1985; 3(3):292–300. [PubMed: 4032102]
81. Robinovitch SN, Hayes WC, McMahon TA. Predicting the impact response of a nonlinear single-degree-of-freedom shock-absorbing system from the measured step response. *J Biomech Eng*. 1997; 119(3):221–7. [PubMed: 9285332]

Highlights

- We tested the effect of a simulated fall event on the failure of lumbar human spine.
- We evaluated the role of disk dynamic response, bone density and MR hydration.
- Disk viscoelastic properties and hydration strongly correlated with spine failure.
- These correlations were independent of bone density measures in the vertebrae.
- Disk degeneration may form a part of fracture risk assessment in the elderly spine.

Figure 1.

The pendulum apparatus consisted of a pendulum arm that struck and end mass (20.9 kg). The end mass attached to the posterior potted end of the spine segment. The anterior, potted end was attached to a force transducer that was mounted to the wall.

Figure 2.

A diagrammatic illustration of the mechanical test setup used for obtaining the specimen kinematic center of rotation (KCR) and for the performance of high rate testing to failure. Adjustment for the location of the spine KCR with respect to the loading axis of the hydraulic test system was performed using the X-Y tables. Use of the needle bearing allowed application of compressive force to the spine while allowing the spine to freely deform, thus keeping the imposition of coupled loading to a minimum. Compensation for the difference in the measurement of forces and moment acting on the spine vs. that measured at the vertebrae ($D_{Lcs-Vcs}$) is achieved in software.

Figure 3. Force transmission after impact exhibits underdamped oscillations. Fit of the oscillations with Kelvin model is shown as a dark, thick line.

Figure 4.

A displacement-force response of a L1-L3 FSU under applied high rate loading simulating a fall event. The initial failure observed is followed by an immediate renounce and a series of a second-order, under-damped, oscillations with a rapid decrease in the force response post-failure. This sequence of responses clearly indicates the contribution of the intervertebral disks to the immediate post failure response of the spine to high rate loading.

Figure 5.

A fluoroscopic image of a spinal L2–L3 motion segment prior-to and after the high-rate fracture test. Clear damage can be observed at the vertebral body and the fracture of the endplates (a) and the pedicle (c). Extensive damage and fissures can be observed for both intervertebral disks, (b). The anterior aspects of the vertebral body show evidence of buckling and failure of the vertebral cortex (d).

Figure 6.

Regression model demonstrates a significant ($P=0.034$), negative, relationship between the damping parameter [B] and the ultimate strength of the spine. No such correlation was observed for the dynamic stiffness parameter [A].

Figure 7.

The increase in bone mineral density (BMD), [A], and higher MR estimated hydration of the disk [B], independently correlate with the increase in the spine ultimate strength.

Table 1

Correlation coefficients (r) relating stiffness constants with MRI based measurements and with age, height and gender.

Measured property	Mean (S.D)	Correlations (r^2 / $p>F$)		
		Intervertebral disc		FSU
		Stiffness (kN/m)	Damping (Ns/m)	Ultimate strength (N)
Donor age (years)	75.5 (6.9)	-0.11	0.27	-0.1
Donor gender (0 – women)	-----	0.05	0.10	0.00
Donor height (cm)	163 (33.0)	0.12	-0.09	0.0
Disk crosssectional area (cm ²)	16.8 (3.2)	0.04	0.15	-0.1
Mean Disc Height (cm)	0.9 (0.2)	0.07	0.17	-0.1
T2 (ms)	65.6 (16.0)	0.21	0.10	-0.1
PD (normalized)	6.8 (0.1)	0.04	-0.44 /**	0.43 /**
Dynamic stiffness (KN/m)	420.4 (203.8)	-	0.0	0.0
Damping (Ns/m)	376.7 (160.0)	-0.0	-	0.28 *
BMD: A-P (g/cm ²)	0.7 (0.2)	-0.0	0.0	0.46 /**
BMD: Lateral (g/cm ²)	0.3(0.1)	0.1	0.1	0.17

Gender was assigned as 0 for women and 1 for men. Intensities are relative to the heavy water phantoms to compute % of disc hydration. Values marked with a * are significant at the 5% level.