Title: Comparison of specimen-specific vertebral body finite element models with experimental digital image correlation measurements

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Abstract

The purpose of this study was to load cadaveric vertebral bodies ($n = 6$) in compression and compare the response, measured with digital image correlation (DIC) on the cortex, with the predicted response from specimen-specific vertebral finite element (FE) models. Five modulus-density relationships were evaluated, and for the strongest modulus-density relationship, the correlation between the DIC and FE displacements had R^2 values from 0.75 to 0.93. The stiffnesses derived from the DIC measurements were strongly predicted by the FE stiffnesses ($R^2 = 0.90$). DIC provides full-field measurements of surface displacement, eliminating the influence of system compliance, for validation of specimen-specific models.

Keywords: digital image correlation; vertebra; compression; specimen-specific; finite element modeling

Introduction

Vertebral fractures are associated with serious consequences, such as pain, disability, or neurological deficits, and deleterious changes in quality of life [1]–[4]. Vertebral fractures can occur from a single traumatic loading or, in the context of osteoporosis, from damage accumulated during repeated activities of daily living. Specimen-specific finite element (FE) models are an important tool to help researchers understand the relationship between vertebral loading and fracture. These models account for vertebral geometry, bone density and density distribution and better predict vertebral strength and stiffness as compared with dual-energy x-ray absorptiometry (DEXA) [5], [6]. Numerous studies have compared specimen-specific FE models of vertebral bodies loaded in compression to experimental measurements. In general, the models have predicted the experimental strength well ($R^2 = 0.79$ to 0.96) [6]–[10] while whole bone stiffnesses are predicted less well ($R^2 = 0.49$ to 0.86) [6], [7], [9]–[11]. Studies that predict strength do not use consistent yield or failure metrics. Furthermore, the stiffnesses have been derived based on displacements measured for the entire tested specimen including the ~~potting~~ potting. The platen-specimen interfaces may influence the stiffness, and inaccurate bone properties may have been derived to compensate for compliance at these interfaces. More precise and specific validation metrics may improve the assessment of the quality of specimen-specific FE vertebral body models.

One validation metric that has not yet been used is surface displacement of the vertebral bodies. This can be measured using digital image correlation (DIC), a non-contact optical method for measuring the displacement or strain on the surface of a material or structure during loading. Images of the surface deformation are analyzed based on pattern matching criterion to determine the motion of pixel subsets. The use of DIC for validation of FE models has been performed for composite bones [12]–[15], but to our knowledge only one study has compared DIC results to specimen-specific FE models of cadaveric bones [16]. In that study, three FE models were created of proximal femora and compared to the experimental DIC strains from cadaveric testing. Strong correlation ($R^2 = 0.92$ to 0.95) was found on the femora for the FE to DIC principal strains, and the failure load and location were predicted well by the FE models. While these results are promising, no investigations have been performed using DIC measurements from vertebrae loaded *ex vivo* for FE model validation of predicted displacements and derived stiffness values. This is an important step to identify an appropriate modulus-density relationship, necessary for future work establishing yield or failure criteria in bone generally and specifically in the context of vertebral fracture prediction.

The purpose of this study was to compare the surface displacements predicted by specimen-specific vertebral body FE models with the displacements measured by DIC during quasi-static compressive loading of ex vivo specimens. We hypothesize that agreement will be seen between the predicted and observed bone surface displacements and also for the derived stiffness values.

Materials and Methods

Experimental procedure

Human fresh-frozen cadaveric thoracolumbar vertebrae ($n = 6$) were obtained from a body donation program (Table 1). The specimens were scanned in air at an isotropic resolution of 246 microns in a high-resolution peripheral quantitative computed tomography (HR-pQCT) scanner calibrated with a hydroxyapatite phantom (XtremeCT and Q1 Phantom, Scanco Inc., Switzerland). During preparation, the specimen was kept hydrated using saline. The soft tissue including the periosteum was removed using a scalpel. Care was taken to not cut into the cortical shell while removing the soft tissue. The posterior elements were removed from the vertebrae by cutting through the pedicles. Each vertebral body was potted using polymethylmethacrylate (PMMA) to create parallel surfaces. For the DIC analysis, the anterior cortex was blotted dry with paper towel and painted with a thin layer of white paint. Then an airbrush was used to create a random speckle pattern with black paint. Following this, four registration points were painted on the bone; these dots were used to register the DIC surface to the surface of the vertebra from the CT scans (Figure 1).

The inferior PMMA was rigidly attached via screws to a metal plate (Figure 2). The specimen was placed in a material testing system (MTS) (Electropuls E10000, Instron, Norwood, MA, USA) and compressed to a preload of 100 N. A 3D point digitizer (Optotrak Certus, Northern Digital Inc., Waterloo, ON, Canada) was used to measure the coordinates of the four registration points, the loading vector, and points on the vertebral body to use for registration with the geometry segmented from CT (113 to 131 points per specimen). Two video cameras (V12, Vision Research, Wayne, NJ, USA) were positioned an average of $22.2^\circ (\pm 0.5^\circ)$ apart to image the anterior surface of the vertebra and the specimen was illuminated with halogen lights. Camera images were recorded at 100 frames per second with an image resolution of 1280 by 800 pixels. A monotonic loading ramp at a rate of 0.1 mm/s was applied up to 3 mm, a displacement that resulted in failure for all specimens. The load was recorded at 12.5 kHz.

Computational procedure

Specimen-specific models of each vertebral body were developed. To create the geometry, open-source software [17] was used to interactively segment the vertebral bodies from the CT scans. Each slice of the CT scan was visually checked to ensure consistent vertebral body segmentation. A convergence investigation was performed for nominal element edge lengths of 1, 2, 3, and 4 mm; an edge length of 2 to 2.5 mm was selected based on the convergence of the average displacement of the superior endplate. The volumes were meshed with 10-node quadratic tetrahedrons using commercial software (V15, ANSYS Inc., Canonsburg, PA). The PMMA was modeled by extruding the endplates a nominal height of 10 mm. The PMMA was assigned linear elastic material properties ($E = 2.9 \text{ GPa}$, $\nu = 0.3$)[18](Figure 2).

Bone material properties were assigned to each element based on five different Young's modulus-density relationships previously experimentally determined for vertebrae (Table 2 and Figure 3). These relationships were previously reviewed by Helgason et al. [19]. The ash density (ρ_{ash}) in each CT voxel was related to the calibrated bone mineral content (mgHA) using $\rho_{\text{ash}} = (\text{mgHA}/1000 + 0.09)/1.14 \text{ g/cm}^3$ and to apparent density (ρ_{app}) with $\rho_{\text{app}} = \rho_{\text{ash}}/0.6$ [20]. The distribution of the apparent densities for each specimen is provided (see supplemental materials).

The material properties from the empirical modulus-density relationships were mapped to the FE elements using Method B from a recent study [26]. This method accounted for the partial volume effect by removing the voxels at the air-bone boundary and expanding the unaffected layer of voxels outward. The Young's moduli were then calculated for each voxel and the moduli of the voxels were used to calculate the moduli of the elements. The Poisson's ratio for the bone was set to 0.3. To determine the volumetric bone mineral density (vBMD), the ash density for each specimen was determined by finding the volumetric average of the FE elements (Table 1).

The models were solved in ANSYS (V17, ANSYS Inc., Canonsburg, PA). The simulations were performed in two steps, in order to perform a relevant and consistent comparison of experimental and simulation data. First, the model with the PMMA was solved, applying a uniform vertical displacement of 1 mm to the superior surface of the PMMA while constraining the transverse motion of the superior PMMA. The inferior PMMA was fixed to be stationary and the superior endplate nodes and the adjacent PMMA nodes were tied, providing the force distribution at the superior PMMA-vertebra interface. The nodes were tied since this best simulated the experimental conditions in which the PMMA was well attached to the endplate. For comparison, the simulations were also performed using sliding contact between the PMMA and the endplate; sliding contact resulted in an average R^2 value of 0.87 for the correlation

compared to 0.88 with the tied contact but slightly improved stiffness agreement (See Supplemental Materials). Second, the model was simulated without the PMMA, using the determined force distribution to apply a total force of 1000 N to the superior endplate, again constraining transverse motion of the superior endplate, while the inferior endplate was constrained to be stationary. In this way we ensure that the FE to DIC comparison is carried out at the same load, with a reasonable estimate of the force distribution at the endplates applied in the FE model. This method isolates the response of the bone, given that compliance between the bone and the PMMA is unknown.

Data processing

Images were processed using commercially available DIC software (StrainMaster 8.2, LaVision, Göttingen, Germany). For processing, a subset size of 31 by 31 pixels was used with a spacing of 15 pixels between the subset centers. The displacements were low-pass filtered at 10 Hz with a fourth-order Butterworth filter. The average spatial resolution of the reconstructions was 32.9 pixels/mm. To account for the vertical displacement of the inferior PMMA-bone interface during loading, all DIC displacements were reported relative to a plane perpendicular to the loading vector at the inferior edge of the DIC measurement area.

To compare DIC and FE measurements, results were transformed to the experimental location of the bone (points digitized during the experiment while the bone was preloaded) (Figure 4). For the DIC surface, the x-,y- and z- coordinates of the registration points were exported from the DIC software. These coordinates were aligned to the digitized control point coordinates, minimizing the sum of squared errors between the corresponding control points. To align the segmentation from the CT to the experimental bone location, an iterative closest point registration algorithm without scaling was used to transform the CT surface coordinates to the digitized points resulting in the two surfaces (FE and DIC) in the same experimental coordinate system. For the center of each DIC subset, the ten closest exterior FE nodes were found and averaged, giving the FE displacement at a force of 1000 N that corresponded to the DIC displacement. The goodness of the alignment was assessed by the average resultant distance between the points on the DIC surface and the corresponding point on the FE surface.

Stiffnesses were calculated for each specimen based on the DIC displacements. For the DIC-measured displacements, the time histories of the displacements of the superior DIC subsets were found relative to the inferior subset displacements, aligned by the loading vector (Figure 5). Since displacements calculated over short distances would have higher errors, lengths were eliminated from analysis if the distance was less than 80% of the maximum distance for each

specimen, and because the original lengths varied across specimens and between specimens, the displacements were normalized to the original lengths. The axial force applied was plotted against the normalized DIC displacements. The stiffnesses for each subset were calculated by fitting a line to the force-displacement curve between 25 and 75% of the maximum force reached during the testing. The specimen's DIC stiffness was defined as the average of all the subset-derived stiffnesses. For comparison with the DIC-derived stiffness, the displacements predicted by FE were found at the same locations as the experimental measurements; the stiffness was calculated by dividing the applied force by the average of the displacements, normalized to the original lengths.

Statistics

Linear regressions were used to compare the DIC-measured displacements with the FE displacement predictions location-by-location for the bone at a compressive load of 1000 N. The coefficient of determination (R^2) was used to quantify how well a linear model described the FE to DIC relationship, and the concordance correlation coefficient (CCC) [27] was used to measure the similarity of the measurements to the unity line ($x = y$). Values of the CCC range from -1 to 1 with 1 representing perfect agreement. For the modulus-density relationships, the differences in the R^2 and CCC values were compared using a Kruskal-Wallis test with a Tukey post-hoc test. The agreement between the FE and experimental stiffnesses were assessed with R^2 , and relationships were considered significant for $p < 0.05$. A linear regression between experimental and FE-predicted stiffnesses was performed, and the slope and the intercept of the regression line was determined as well as the coefficient of determination (R^2). Hypothesis testing was used to evaluate whether the slope was different from unity and intercept different from zero, with $p < 0.05$ (Matlab R2015b, The Mathworks, Inc., Natwick, MA, USA).

Results

The average resultant distance between the DIC surface points and the FE nodes after registration was 0.88 mm, when the bone was preloaded to 100 N. The maximum distance between the DIC surface point and FE node for all points was 2.24 mm.

The DIC displacements for the specimens at a load of 1000 N were plotted, demonstrating differences in the displacement distributions between specimens (Figure 6).

The correlation between the FE-predicted displacements and the DIC-measured displacements was quantified using the coefficient of determination and the CCC for the five modulus-density relationships (Table 3). There were no significant differences in the R^2 values for any specimens

and no differences in the CCC for MDR-1, MDR-3, and MDR-5. The relationship with the highest R^2 was MDR-5 which had an average value of 0.88 and CCC of 0.80. The results for each specimen using MDR-5 are presented including the root mean square error (RMSE) of the correlation (Figure 7).

The FE stiffness was compared to the experimental stiffnesses for MDR-5 on the surface of the bone (Figure 8). The R^2 value was 0.90 and the relationship was significant ($p = 0.004$). The slope was not significantly different from one ($p = 0.64$), and the intercept was not different from zero ($p = 0.80$).

Discussion

The goal of this paper was to compare the surface displacements and stiffnesses predicted by specimen-specific FE models of lumbar vertebral bodies with DIC measurements of the vertebral surface displacement and derived stiffnesses during compression. Five modulus-density relationships characterizing vertebral trabecular bone were evaluated. The modulus-density relationship that resulted in the strongest agreement between the model and experimental displacements (MDR-5, [25]) had an average R^2 of 0.88 and an average CCC of 0.80. Considering all specimens, the specimen-specific FE model stiffnesses derived from vertebral surface displacements using MDR-5 were significantly correlated with the stiffnesses derived from experimental DIC-measurements ($R^2 = 0.90$, $p = 0.004$).

For the point-by-point displacement comparisons, the FE and experimental displacements were generally highly correlated. The R^2 values ranged from 0.75 to 0.93. However, one specimen, 4-L1, had a concave shape to the displacement correlation (Figure 7). The displacements for this specimen were generally similar for low values but for higher displacements, the DIC displacements were higher than the FE displacements. It is not clear why the superior cortex experienced more displacement in the experiment than predicted in the model but it may be a result of subtle damage to the bone prior to reaching 1000 N (or pre-existing damage). To evaluate the similarity of the displacement magnitudes, the CCC was used. The modulus-density relationship with the highest average CCC (MDR-5) had greater than 0.8 CCC for five out of six specimens while one specimen (1-L3) had a CCC value of less than 0.5. Specimen 1-L3 was stiffer in the FE model than in the experiment, which again may be the result of some damage to the bone that was not accounted for in the FE model. Although there were some outliers for the shape and magnitudes of the displacement correlation, the correlation between predicted stiffness and derived stiffness on the surface of the bone was significant ($R^2 = 0.90$, $p = 0.004$). The strong agreement between the models and experiments indicates that the

selected modulus-density relationship and boundary conditions resulted in specimen-specific FE models that were highly predictive of the anterior cortex experimental mechanics.

To compare the FE and DIC results, we developed a method to align the relatively feature-less anterior surface of the vertebral body obtained from DIC to the segmented FE geometry using points digitized during the experiment. We obtained good alignment of the DIC and FE surfaces. To compare the FE and DIC measurements, previous studies used registration of the DIC surface to the FE geometry for the femur [16] or landmark alignment for the pelvis [15]. However, the anterior surface of the vertebral body has less distinct geometry than other bones, and therefore, the additional step of aligning both the DIC surface and FE geometry to the experimental point cloud was required.

In this investigation, five modulus-density relationships specific to thoracolumbar vertebrae were used in specimen-specific models. Each MDR had slightly different values for the curve relating the modulus to the bone density (Figure 3). These differences in the modulus-density relationships may be a result of different specimen geometries, whether the studies measured the platen displacement or used an extensometer attached to the bone, differences in bone preparation, or the range of densities tested [19]. MDR-1, MDR-3, and MDR-5 all resulted in good agreement with the FE models based on the CCC. However, there were no obvious experimental commonalities between these studies that would make them more predictive of the experimental bone response than MDR-2 and MDR-4.

Comparing our results for local displacement correlation directly to previous work is difficult, due to methodological differences. However, one study compared FE predictions and digital volume correlation (DVC) measurements for vertebral displacements of three-vertebra segments loaded in compression [28]. DVC is a method similar to DIC but DVC uses volumetric subsets instead of 2D subsets and utilizes the natural trabecular bone structure for pattern matching instead of an applied speckle pattern. The study found some qualitative agreement in the displacement patterns but quantitatively, the displacement errors were high (12 to 279%). While both DIC and DVC provide additional measurements that can be used to validate FE models, the increased number of experimental measurements elucidates differences between the model and experiment that could not previously be quantified.

Previous studies have used DIC for validation of FE models of composite bones [12]–[15] and cadaveric femora [16]. These studies have all used strain to quantify the agreement between the experiments and the models. These previous investigations were performed on the femur [12]–[14], [16] or pelvis [15] while the current investigation loaded vertebral bodies in

compression. Loading in the previous studies resulted in bending stresses and strains on the bone while the current study only applied direct compression to the vertebral body. The loading applied in these previous studies may create more distinct displacement and strain patterns on the femurs and pelvises than were created on the vertebral bodies.

Previously, the FE-experimental agreement for stiffness was measured using a LVDT and was somewhat lower than our values ($R^2 = 0.49$ to 0.86) [6], [7], [9]–[11]. Other studies have not quantified the stiffness based on surface displacement measurements on the bone, as was performed using DIC in this study. We believe that this improvement over previous work can most likely be attributed to our approach for isolating the bone response in the FE models using first a simulation to determine the load distribution on the superior endplate by applying displacement on the PMMA cap and then applying that load distribution in a second simulation where only the bone response was modelled. Using this method, we isolated the response of the bone, eliminating the necessity to know the compliances between the platen and the PMMA and the PMMA and the bone. Previously, the issue of compliance between the bone-PMMA and PMMA-platen has been addressed by removing the endplates to create parallel surfaces, eliminating the need for using PMMA experimentally [6], [10], [11], however at the expense of biofidelity. These studies, along with the current study, show that compliance may negatively affect the comparison between experiments and FE models.

All studies which incorporate both experimental and computational models of biological tissue have limitations. In our case, this study used a small sample size; testing more specimens may result in improved confidence in the FE models for predicting displacements and stiffnesses. Nevertheless, strong and significant correlations were demonstrated. The DIC displacements were only measured on the anterior surface of the bone to allow for further testing of these specimens for a separate study. Future work measuring the posterior cortex displacement would provide additional validation data. The current study compressed the bone through PMMA. Although this loading is not as physiologic as loading the vertebral body through the intervertebral disc (IVD), we took this approach because the PMMA is likely more consistent than the IVDs of elderly donors. The geometries of the bones were obtained in HR-pQCT scanner which allowed for higher resolution scans than typical clinical CTs but cannot be used *in vivo* to image the spine. Another limitation is use of isotropic linear elastic bone properties. Given the range of loading, it is not expected that the bone would be outside the linear range but this material model may not fully capture the complexity of bone compression. The PMMA contact with the circumferential element faces at the superior and inferior edges of the bone

were not modeled; the PMMA contact was only modeled for the vertebral endplates. However, the effect of the added constraint on these elements is expected to be small since the endplate nodes are tied to the PMMA nodes in the first simulation step. Finally, the cortex was not explicitly modeled. However, the elements containing cortical bone had higher stiffnesses than the interior elements meaning the stiffening effect of the cortex was captured. A downside to this approach was that the experimental modulus-density relationships were extrapolated beyond the experimental density regimes which may create additional errors. Previous work has shown that the predictions of global stiffness and strength are improved with explicit cortical shell modeling [9] but the stiffness agreement in the current study was strong, even without modelling an explicit cortical shell, implying that its influence is captured in the chosen material mapping strategy.

In summary, we created specimen-specific FE models and identified a modulus-density relationship that accurately captured the local cortical response quantified experimentally using DIC. For this study, the surface displacements were compared on a point-by-point basis using a novel method for alignment of the DIC surface and FE segmented geometry and good agreement was found for the displacements. The stiffnesses predicted by FE strongly agreed with those derived from experimental measurements ($R^2 = 0.90$). Use of DIC allows for a more comprehensive characterization of surface deformation than has previously been performed and this information enables improved validation of specimen-specific FE models.

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Table 1: Summary of the vertebral body specimens tested. To determine the average volumetric bone mineral density (vBMD) for each specimen, the volumetric average of the ash density was found for FE elements in the model.

	Sex	vBMD (g/cm ³)	Donor Age
1-L2	U/k	0.28	U/k
1-L3		0.29	
2-L1	M	0.15	51
3-T11	M	0.25	36
4-L1	F	0.18	89
5-L1	M	0.24	66

U/K: unknown

Table 2: Summary of the modulus-density relationships used for specimen-specific models of vertebral bodies. All relationships were based on experiments using vertebral trabecular bone. Each modulus-density relationship was given a label for identification (MDR-#).

Label	MDR-1	MDR-2	MDR-3	MDR-4	MDR-5
Author [Reference]	Keaveny [21]	Keller [22]	Kopperdahl [23]	Morgan [24]	Ouyang [25]*
Relationship	$E = 1540\rho_{app}^{-58}$	$E = 1890\rho_{ash}^{1.92}$	$E = 2100\rho_{app}^{-80}$	$E = 4730\rho_{app}^{1.56}$	$E = 2383\rho_{app}^{1.88}$

E in MPa, ρ_{app} and ρ_{ash} in g/cm^3

* Rate dependent component was not included since testing was performed at a quasi-static rate

Table 3: Summary of the CCC and R^2 values for local displacement comparisons between DIC and FE results. The comparison was carried out for each specimen and each modulus-density relationship.

	MDR-1		MDR-2		MDR-3		MDR-4		MDR-5	
	R^2	CCC	R^2	CCC	R^2	CCC	R^2	CCC	R^2	CCC
1-L2	0.93	0.92	0.93	0.30	0.93	0.71	0.93	0.31	0.93	0.92
1-L3	0.92	0.41	0.93	0.61	0.89	0.26	0.93	0.12	0.93	0.44
2-L1	0.87	0.85	0.91	0.18	0.87	0.93	0.88	0.43	0.91	0.88
3-T11	0.83	0.81	0.80	0.30	0.83	0.58	0.82	0.27	0.80	0.84
4-L1	0.89	0.92	0.93	0.30	0.89	0.70	0.91	0.23	0.93	0.87
5-L1	0.75	0.85	0.75	0.21	0.75	0.69	0.75	0.29	0.75	0.86
Average	0.86	0.79	0.88	0.32	0.86	0.65	0.87	0.27	0.88	0.80
Standard Deviation	0.07	0.19	0.08	0.15	0.06	0.22	0.07	0.10	0.08	0.18

Figure 1: Image of a prepared specimen potted in PMMA and painted with a speckle pattern and points for registering the DIC surface to the segmented geometry. The outline indicates the area for which the DIC displacements were measured. The four registration points used to align the experimental and FE results are indicated by the arrows.

Figure 2: Example of a specimen in the experimental setup and a specimen-specific FE model. The specimens were loaded in compression to failure. The FE modeling was performed in two steps. First, the bone was loaded with a constant displacement on the superior PMMA to determine the force distribution on the endplate. Second, that force distribution was used to apply 1000 N on the superior endplate, excluding the PMMA from the model.

Figure 3: Graphical comparison of the relationships used for assignment of material properties to the FE elements. The dashed line and markers represent the different relationship extrapolated for a wide range of apparent bone densities while the thick line represents the range of the bone densities tested experimentally to determine the specific relationship.

Figure 4: Graphic showing the alignment of the FE and DIC measurements by transforming both surfaces to the surface of the bone that was digitized during the experiment. By aligning the surfaces, it was possible to compare the FE and DIC measurements on a point-by-point basis.

Figure 5: Illustration of the lengths used for calculating the stiffness. Each dot represents the center of a DIC subset measured on the anterior surface of the bone. The displacements were measurement between superior and inferior subsets aligned in the direction of the loading vector and normalized to the original distance between the subsets. Lengths were excluded if the distance between the subsets was less than 80% of the longest distance between a superior and inferior subset.

Figure 6: For each specimen, the DIC-measured displacement magnitudes (mm) on the anterior surface of the vertebral body specimens are provided for an applied compressive load of 1000N; displacements were measured relative to a plane perpendicular to the loading vector at the inferior edge of the specimen.

Figure 7: Point-by-point comparison of the anterior surface displacements at 1000 N predicted by the specimen-specific vertebral body FE models using MDR-5 relationship compared to the displacements measured using DIC during experimental loading. The load distribution on the superior endplate was determined based on tied contact between the PMMA and endplate nodes.

Figure 8: Comparison of the FE stiffnesses to the experimental stiffnesses of the bone for MDR-5. The load distribution on the superior endplate was determined assuming tied contact between the PMMA and endplate nodes. The stiffness was calculated based on the displacement measured with DIC and predicted by FE on the anterior cortex. The experimental stiffnesses were found by fitting a line to the force-displacement curve between 25 to 75% of the maximum force. The FE stiffnesses were calculated by dividing the applied force of 1000 N by the average of the normalized displacements corresponding to the locations of the DIC measurements. The slope and intercept were not significantly different from 1 and 0, respectively.

Figure 9: Histograms showing the distribution of the apparent densities of the FE elements in each specimen.

Figure 10: Point-by-point comparison of the anterior surface displacements at 1000 N predicted by the specimen-specific vertebral body FE models using MDR-5 relationship compared to the displacements measured using DIC during experimental loading. The load distribution on the superior endplate was determined based on sliding contact between the PMMA and endplate nodes.

Figure 11: Comparison of the FE stiffnesses to the experimental stiffnesses of the bone for MDR-5. The load distribution on the superior endplate was determined assuming sliding contact between the PMMA and endplate nodes. The stiffness was calculated based on the displacement measured and predicted by FE on the anterior surface. The experimental stiffnesses were found by fitting a line to the force-displacement curve between 25 to 75% of the maximum force. The FE stiffnesses were calculated by dividing the applied force of 1000 N by the average of the normalized displacements corresponding to the locations of the DIC measurements.

Figure 12: Correlation between the volumetric bone mineral density (vBMD) and the DIC stiffness. The vBMD measurement was found based on the HR-pQCT scan. The experimental stiffnesses were found by fitting a line to the force-displacement curve between 25 to 75% of the maximum force for the specimen.