

Full length article

Can proton density-weighted MRI-based finite element models predict bone strength?

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ABSTRACT

Hip fractures are a major clinical concern, and patient-specific finite element (FE) modeling of the proximal femur is increasingly used to assess bone strength and fracture risk at the hip. Although most patient-specific FE models rely on computed tomography (CT) imaging, CT requires relatively high doses of radiation, limiting its application. Magnetic resonance imaging (MRI)-based FE models present a radiation-free alternative that could expand clinical applicability. This study evaluated the predictive capability of proton density (PD)-weighted MRI-based FE models of the proximal femur under sideways fall. Cadaveric femora ($n = 14$) were imaged using MRI ($n = 10$) and CT ($n = 14$) to construct patient-specific FE models, and then mechanically tested to obtain experimental measures of stiffness and failure load. Existing relationships between image voxel intensity and tissue modulus were used; in the case of the MRI models, the relationship was based on a BV/TV-modulus equation previously applied in a T2-weighted MRI-FE study. Digital image correlation (DIC) was used to measure full-field strain distributions during mechanical testing to compare with FE predictions. Values of the whole-bone stiffness and failure load predicted by the MRI-based models showed poor agreement with experimentally measured values and paired CT-based FE predictions. However, when homogeneous material properties were assigned to the FE models, predicted stiffness and failure load no longer differed between MRI and CT. Notably, both MRI- and CT-based FE models qualitatively approximated the spatial distribution of principal strain patterns measured experimentally using DIC. These results indicate that the femoral geometry is captured well by PD-weighted MRI, but the appropriate material mapping relationship to use with this imaging sequence is unknown. Thus, future work should explore the relationship between PD-weighted MRI signal values and measures of modulus or density to advance an MRI-based FE framework for fracture risk assessment.

1. Introduction

Hip fractures are a major worldwide socioeconomic problem. These fractures represent 20% of all fragility fractures [1], and falls to the side of the hip cause over 90% of hip fractures [2]. Following a hip fracture, fewer than 50% of patients recover full mobility [3], and up to 30% die within a year [4]. Hip fractures account for up to 70% of the costs of fracture care and up to \$12 billion annually [5–7]. The substantial mortality, morbidity, and healthcare costs of hip fracture make widely accessible, efficient, and accurate prediction of bone strength and fracture risk a critical priority.

Patient-specific finite element (FE) models have been shown to be a good predictor of the biomechanics of the proximal femur under both habitual (i.e., stance phase of gait) and traumatic (i.e., sideways fall) loading scenarios [8]. While models built from computed

tomography (CT) imaging are the current state of the art for patient-specific FE modeling, they are not routinely used in clinical screening in part because of the radiation dosage of CT [9,10]. Consequently, magnetic resonance imaging (MRI)-based FE modeling has emerged as an attractive alternative [11–15]. While prior studies on MRI-based FE modeling have used T2-weighted MRI and reported good agreement with mechanical testing, with correlations (R^2) as high as 0.88 for stiffness and 0.92 for failure load, [12,16], other MRI sequences are also used for imaging bone [17,18]. For example, Proton Density (PD)-weighted spin echo MRI sequences provide an image wherein fat and fluids appear bright, muscles appear mid-gray, and bone and air appear dark, making this sequence clinically useful for musculoskeletal imaging. The PD sequence provides high structural contrast and enhanced anatomic detail when imaging extremities (e.g., shoulder, elbow, hip,

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Table 1
Donor characteristics. Age and BMI reported as mean ± stdev (range).

n (sex)	Age (years)	BMI (Kg/m ²)	Race (n)
14 (6F/8M)	54.5 ± 17.1 (25 – 84)	25.8 ± 5.3 (20.1 – 38.7)	African American (1) White (12) N/R (1)

knee, and ankle) for the assessment of bone [19–22] and evaluation of ligamentous injury [23]. The use of PD-weighted MRI-based FE modeling has not yet been evaluated and could open up an additional radiation-free avenue for non-invasive, patient-specific assessment of bone strength and fracture risk at the hip.

Thus, the goal of this study was to assess the predictive capacity of PD-weighted MRI-based FE models of the human proximal femur under sideways fall loading. Specifically, the objectives were to: (i) develop patient-specific PD-weighted MRI-based and CT-based FE models of *ex vivo* “healthy” cadaveric femora; (ii) compare the MRI- and CT-based FE predictions of stiffness, strength, and deformation to experimentally measured values from mechanical testing of the same femora; and (iii) compare the predictions from MRI-based models to those from corresponding CT-based FE models.

2. Methods

2.1. Subjects

2.1.1. Cadaver specimens

Fresh-frozen human femora (n = 14, 6F/8M, age range = 25–84 years) from donors with no reported history of musculoskeletal or metabolic disease were acquired from tissue banks (National Disease Research Interchange, Philadelphia, PA; ScienceCare, Phoenix, AZ) (Table 1). Use of de-identified cadaveric specimens is determined by the Institutional Review Board from our institution to be exempt from review.

Specimens were stored frozen between imaging and mechanical testing sessions. Each specimen underwent a total of three freeze-thaw cycles: one prior to MRI scanning, one prior to CT scanning, and one prior to biomechanical testing.

2.2. Imaging and image processing

2.2.1. MRI scanning

MRI scans were performed on a 3-T GE Discovery MR750 W scanner (GE HealthCare Technologies, Chicago, IL) with a flip angle of 111° and an acquisition time of 127.7 ms per slice (Fig. 1A) (Cognitive Neuroimaging Center, Boston University, Boston, MA). These scans consisted of orthogonal, small-field coronal (field of view = 169.98 × 169.98 × 84 mm; voxel size = 0.332 × 0.332 × 3.5 mm; acquisition time ≈ 30 mins) and sagittal (field of view = 169.98 × 169.98 × 126 mm; voxel size = 0.332 × 0.332 × 3.0 mm; acquisition time ≈ 30 min) scans with a proton density (PD) weighted sequence (TR/TE = 4712/28.176 ms). Specimens were thawed at room temperature for 12–24 h and wrapped in gauze soaked in phosphate-buffered saline (PBS) prior to scanning. The femora were positioned in a supine orientation for scanning and supported by malleable aqueous phantoms to boost the signal-to-noise ratio. Some (n = 4) MRI scans were excluded due to insufficient information to render the full proximal femur for evaluation.

Super-resolution reconstruction (SRR) was used to collate image data from the coronal and sagittal MRI scans to generate a single high-resolution 3-D volume of image data of a given proximal femur [24,25]. SRR is the result of image-restoration processing on one or more low-resolution images to improve image spatial resolution. Briefly, the

coronal (voxel size = 0.332 × 0.332 × 3.5 mm) and sagittal (voxel size = 0.332 × 0.332 × 3.0 mm) MRI data were co-registered together in 3-D space according to the patient coordinate system stored in the metadata from the scanner. Each voxel in each scan populates the respective spatial index of an isotropic voxel (0.332 × 0.332 × 0.332 mm) in a high-resolution, sparse, 3-D matrix. The remaining entries in the high-resolution matrix are populated using bi-cubic interpolation (Amira, ThermoFisher Scientific). The result is a 3-D dataset with isotropic voxels (0.332 × 0.332 × 0.332 mm) that contains the combined imaging data from the coronal and sagittal scans, allowing for a reconstructed new axial view, which was not available from either orthogonal scan alone. The total acquisition time for the two orthogonal PD-weighted scans was approximately 60 min per specimen.

2.2.2. CT scanning

CT scans (GE HealthCare Technologies, Chicago, IL) were performed with scan parameters of 120 kV, 240 mA, slice thickness = 0.625 mm, pixel size = 0.625 mm (Fig. 1B) (Boston Medical Center, Boston, MA). Specimens were thawed at room temperature for 12–24 h and immersed in water inside a custom-fabricated capsule prior to scanning (Fig. S1). The femur capsule was placed on top of a commercially available, clinically used spine phantom (calibration rods of 0, 75, and 150 mgHA/ccm) and positioned in a supine orientation for scanning.

2.3. Finite element analysis

2.3.1. Segmentation and meshing

The proximal femur was semi-automatically segmented from the surrounding background and soft tissue (MITK-GEM 2017.5.0, SimTK, [26]). The segmentation of the proximal femur from the image data (Fig. 1C and D) resulted in a standard triangular language (STL) representation of the proximal femur geometry, which was later smoothed and refined to reduce the number of surface nodes (n = 10,000). A second-order tetrahedral volume mesh was then generated from the surface mesh using an advancing-front algorithm (MITK-GEM 2017.5.0, SimTK, [26]).

Dice similarity coefficient (DSC) was used to assess both intra- and inter-operator reproducibility and repeatability of the MRI segmentation workflow. Intra-operator repeatability was quantified through repeated segmentations by a single operator, while inter-operator reliability was assessed by having two trained operators independently conduct segmentation on a subset of specimens (n = 6). DSC was calculated for both a single mid-coronal slice and the whole volume (Fig. S2, S3 and Table S1, S2).

2.3.2. Mesh and bin size convergence

Mesh convergence analysis was performed on a single femur (48-year-old female) using both MRI- and CT-prescribed inhomogeneous linear elastic material properties described in the later sections. Seven mesh densities were evaluated: an extra-coarse, coarse, medium-coarse, medium, medium-fine, fine, and extra-fine mesh, corresponding to specified element side lengths of 6.5, 4.5, 3.5, 3.0, 2.5, 1.5, and 1.0 mm, respectively (Fig. S4). All mesh cases were simulated in a sideways fall scenario of 10° adduction and 15° internal rotation. Measures of whole-bone stiffness (N/mm), 90th percentile principal strains (μ ϵ), and failure load (N) were compared across mesh sizes. The medium mesh (element side length = 3.0 mm) provided a suitable balance between computational efficiency and accuracy, yielding stiffness and strain predictions within 5% of the extra-fine mesh for both MRI and CT models. Additionally, a bin size convergence analysis was performed to assess the influence of material discretization on predicted mechanical outcomes. Four binning resolutions (250, 500, 750, and 1000) were evaluated. Across all predicted mechanical measures, the differences between successive bin sizes were less than 2% (Table S4). Based on these results, the medium mesh size and a bin size of 500 were selected for all subsequent simulations.

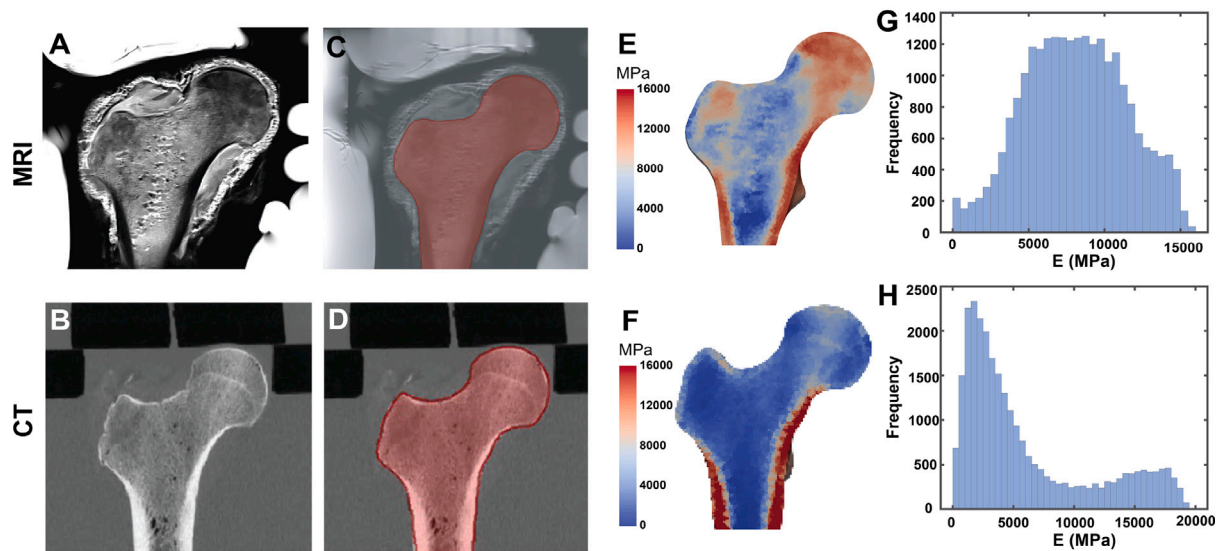


Fig. 1. Representative mid-coronal views of the same *ex vivo* cadaver femur acquired from MRI (top row) and CT (bottom row). The first column (A, B) shows the raw scans, second column (C, D) shows the grayscale image data segmentation, third column (E, F) shows the spatial distribution of inferred elastic modulus, and the fourth column (G, H) shows the histograms of elastic modulus values for the entire FE model.

2.3.3. Material mapping methods

MRI-based material mapping.

MRI-based material mapping followed recently proposed methods, where bone volume fraction (BV/TV) was approximated by an inverted linear scaling of the MRI grayscale values of pure marrow (assumed to be 0% BV/TV) and pure cortical bone (assumed to be 100% BV/TV) [11,14,27,28]. While previous methods relied on manually selecting regions of marrow and pure bone for each patient scan, we instead defined the upper 2.5% of gray values from the segmented bone volume as pure marrow (0% BV/TV) and the lower 2.5% as pure bone (100% BV/TV). This approach minimizes user intervention and the possibility for inter-operator variability.

Each element was assigned an elastic modulus (E in MPa) following the assumption of isotropic, linearly elastic material properties (Fig. 1E and G). The elastic modulus was assigned linearly proportional to the mean BV/TV value of voxels contained within each element, following a relationship previously implemented and evaluated in prior MRI-FE studies by Zhang et al. and Rajapakse et al. [12,29]:

$$E = (15000) * (BV/TV) \quad (1)$$

This validated formulation was retained for the primary analysis to permit direct comparison with earlier MRI-FE literature. However, to address the known non-linear relationship between bone density and elastic modulus [30], we performed a supplementary analysis using a complementary apparent-density power-law mapping approach. In this analysis, MRI-derived BV/TV values were converted to apparent density with the assumption that fully mineralized bone tissue has a density of 1.8 g/cm^3 , such that $\rho_{app} = 1.8(BV/TV)$ [31]. The resulting apparent tissue density values were then used to assign elastic modulus using the density-modulus relationship in Eq. (4). MRI and CT images provide intrinsically different intensity information; thus this analysis was included as a supplementary sensitivity analysis as the relationship between MRI intensity and bone density or apparent density remains unclear and unvalidated. This analysis is reported in the Supplementary Material (Supplementary Section 8.1). A minimum modulus value of 0 MPa and a Poisson's ratio of 0.3 were specified.

CT-based material mapping.

Mean grayscale values were measured from cylindrical volumes of interest (VOIs) within phantom rods of known CT densities (0, 75, and 150 mgHA/ccm) that were included in the scan field of view. These CT density values (ρ_{CT} in mgHA/ccm) were then converted to ash

density (ρ_{ash} in mgHA/ccm) and subsequently to apparent density (ρ_{app} in mgHA/ccm) using the following relationships [32]:

$$\rho_{ash} = \frac{\rho_{CT} + 0.09}{1.14} \quad (2)$$

$$\rho_{app} = \frac{\rho_{ash}}{0.6} \quad (3)$$

Then, each element was prescribed an elastic modulus (E in MPa) (Fig. 1F and H) following a density-modulus power relation [30]:

$$E = 6850 * \rho_{app}^{1.49} \quad (4)$$

A minimum modulus value of 0 MPa and a Poisson's ratio of 0.3 were specified.

Homogeneous material mapping.

To determine the respective contributions of the different material mapping methods versus image segmentations between MRI and CT models, a separate set of simulations was performed using a homogeneous Young's modulus (10 GPa) assigned throughout the entire models. By assigning identical homogeneous material properties to the MRI-FE and CT-FE models, the influence of heterogeneous material mapping was removed, allowing differences in model predictions to be attributed primarily to geometry and segmentation. This allowed us to assess whether the MRI segmentation sufficiently captured the femur geometry relative to the CT-based models.

2.3.4. Boundary conditions

Boundary conditions representative of a sideways fall were applied to each proximal femur model [8,33,34] (Fig. 2). The proximal femur was positioned with 10° adduction, referencing the femur shaft axis, and 15° internal rotation, referencing the axis passing through the femoral head center and mid-femoral neck center. The distal end of the femur was fully translationally constrained, while rotation was only allowed about an axis corresponding to the hinge joint at the distal end. Custom-molded polymethylmethacrylate (PMMA) pads were modeled at the femoral head and greater trochanter to prevent local bone crushing and facilitate uniform load distribution. The modulus of elasticity for PMMA was set to 3.1 GPa and Poisson's ratio to 0.3. The PMMA pad on the femoral head was placed within an aluminum cup with a modulus of elasticity of 70 GPa and a Poisson's ratio of 0.35. The interfaces between bone and PMMA were simulated using a frictional contact of 0.3 [35], while a tie constraint was prescribed

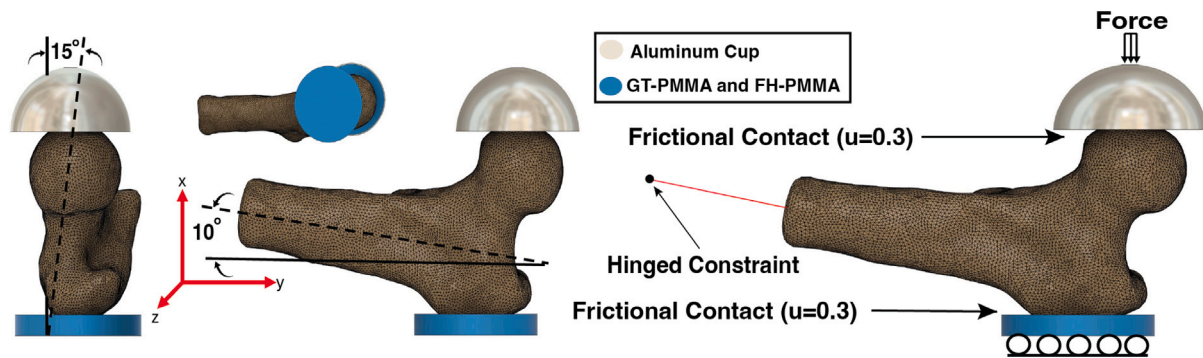


Fig. 2. Finite element (FE) model and boundary condition of a femur in a representative sideways fall scenario.

between the aluminum cup and the femoral head PMMA pad. The base of the PMMA pad at the greater trochanter was placed on a roller bearing, permitting free translation in the transverse plane while being constrained in the loading direction. A compressive load of 2000 N was applied to a reference point rigidly coupled to the top surface of the aluminum cup on the femoral head (Fig. 2). All FE analyses were performed with Abaqus v.15.0.0 (Dassault Systèmes Simulia, Johnston, RI).

2.3.5. FE-predicted mechanical parameters

From the sideways fall simulations, whole-bone stiffness was computed as the ratio between the applied load and the corresponding displacement of the reference point in the loading direction. Bone strength was assessed with an asymmetric maximum principal strain criterion [36–39] to calculate a node-based failure index (FI). To avoid local effects, strains were averaged within a 3 mm radius sub-region surrounding each node [38]. Then, the ratios of maximum and minimum principal strains to the respective strain limit (0.73% in tension or -1.04% in compression) [37] were calculated, and the greater of the two ratios was denoted as the FI. Failure of a given nodal sub-region was calculated by scaling the 2000N load until the failure index reached 1 for that sub-region. The FE-predicted failure load was defined as the load at which 0.1% of all elements within the femoral volume reached the failure criterion. The maximum and minimum principal stresses and strain at the failure load were recorded as well. Failure location was defined as the central node of the sub-region with the highest ratio of principal strain to strain limit. The analysis was done considering all nodes in the mesh volume, analogous to an estimation of failure initiation within the entire model. The identified failure location was then mapped back to the mesh for visualization using the MyMesh Python package [40].

2.4. Mechanical testing

Each cadaveric femora was thawed at room temperature for at least 16 h, cleaned of any remaining soft tissues, and cut 20 cm distal to the greater trochanter. The shaft of the proximal femur was potted in an aluminum cylinder using PMMA (Bosworth Fastray). The femoral head and greater trochanter were also potted using an aluminum spherical cap and a flat plastic cap, respectively, to prevent local crushing (Fig. 3). A black-and-white speckle pattern was applied to the anterior surface of the femur to facilitate measurement of surface strains via digital image correlation (DIC). A custom-fabricated hinge mount positioned the proximal femur with 10° adduction and 15° internal rotation to simulate a sideways fall configuration in the material testing machine (8874, Instron Inc., Norwood, MA) [33,41–44]. Displacement was applied to the femoral head at a controlled speed of 100 mm/s until either macroscopic failure of the specimen occurred or, in the case of two femora, until the force limit of the Instron was tripped. Load was recorded at 5000 Hz by a six-axis load cell (K6D68, ME-Measurement systems GmbH) and a data acquisition system (Interface

BX8-HD44 BlueDAQ). One of the specimens was excluded from quantitative analysis due to the failure of the system software to save the test data. Measures of the whole-bone stiffness and failure load were obtained from the load–displacement curve. Stiffness was computed using the region of the load–displacement curve with the greatest linear slope, representing the maximum elastic response. Failure load was subsequently defined as the force that corresponds to the end of the region used to calculate the slope.

DIC was performed using two high-speed cameras at 1500 frames s^{-1} (Baumer VLXT-06M.I 10 GigE cameras; Trilion Snap 3D software with DAQ and GOM Correlate Pro DIC software, Trilion Quality Systems, King of Prussia, PA). The cameras were equipped with 25 mm lenses and arranged at a convergence angle of 18.5° , capturing the anterior aspect of the femur within a measurement volume of $325 \times 255 \times 255$ mm. Illumination was provided by a high-intensity light source (GS01709, GSVITEC), enabling shorter exposure times and reduced motion blur. Calibration of the cameras was achieved using a certified Trilion Snap 175–005 coded panel at $20^\circ C$, resulting in a calibration deviation of 0.016 px and scale discrepancy of 0.003 mm, both well below recommended limits (0.05 px and 0.014 mm, respectively). The two camera pairs were synchronized via a common end-trigger signal, achieving temporal alignment within one frame. The load cell signal was synchronized with the DIC recordings using their respective acquisition frequencies. Strain fields were computed using a facet size of 19 pixels and a point distance of 16 pixels. Prior to DIC–FE comparison within GOM Correlate, the FE-predicted surface was co-registered to the DIC-derived experimental surface to ensure both datasets were expressed in a common coordinate system. Validation of the DIC system was achieved by running a uniaxial compression test on a polypropylene (PP) cylindrical pipe (Fig. S6). Comparison of strains measured by the DIC system with those obtained from traditional strain gauges showed strong agreement, with absolute differences ranging from 0.02–0.14% for maximum principal strain (E_1) and 0.05–0.07% for minimum principal strain (E_2) (Table S6). In addition, verification of the entire experimental setup was performed by running preliminary tests on composite foam sawbones femora (Sawbones® company) before conducting the cadaveric experiments (Fig. S7, Table S7).

2.5. Statistical analysis

Linear regression and percentage differences were used to compare the whole-bone stiffnesses and failure loads predicted by MRI- and CT-based models to the corresponding experimental values for the paired specimens. Paired t-tests were used to evaluate the difference between MRI- and CT-based predictions of whole-bone stiffness, failure load, and the 90th percentile minimum and maximum principal strains. The Holm-Bonferroni correction was applied to control the family-wise error rate across the mechanical measures within each pairwise comparison. All statistical analyses were performed in JMP Pro 15.0.0 (SAS Institute Inc., Cary, NC, USA). Significance level was

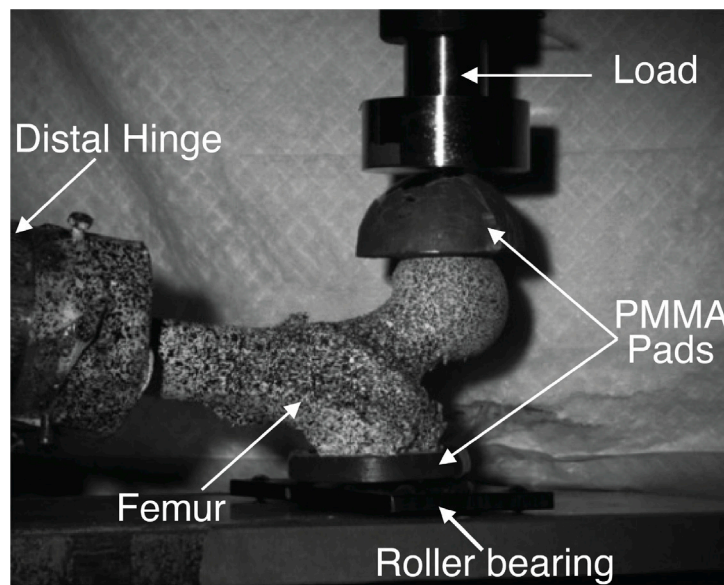


Fig. 3. Experimental set-up for mechanical testing in sideways fall configuration.

set to $\alpha = 0.05$. Predicted failure locations were mapped onto a canonical schematic top-view of the left proximal femur for qualitative comparison. Experimentally measured principal strain fields obtained from DIC at the point corresponding to maximum whole-bone stiffness were qualitatively compared to FE-predicted strain distributions from both imaging modalities to assess the spatial agreement and identify regions of strain concentration.

3. Results

The MRI-based FE predictions showed poor agreement with experimentally measured values of stiffness and failure load, whereas the CT-based predictions demonstrated strong correlation with experiment. The MRI-based predictions showed no correlation for either failure load ($R^2 = 0.16$, $p = 0.29$, $RMSE = 3868$) or whole-bone stiffness ($R^2 = 0.07$, $p = 0.50$, $RMSE = 1920$) (Fig. 4). In contrast, the CT-based models exhibited strong correlation for both (failure load: $R^2 = 0.60$, $p = 0.014$, $RMSE = 879$. Whole-bone stiffness: $R^2 = 0.56$, $p = 0.020$, $RMSE = 607$), consistent with the full-cohort analysis (Fig. S8). Relative to experiment, MRI-based models over-predicted stiffness and failure load with a mean paired percentage difference of 317% and 89%, respectively. CT-based models over-predicted stiffness by 144% and under-predicted failure load by 12%.

Both MRI- and CT-based FE models approximated the spatial distribution of experimentally measured principal strain patterns. Both models identified similar primary regions of high strain concentration: high compressive strains concentrated along the superolateral aspect of the femoral neck (Fig. 5A), while patterns of high tensile strains were observed in the inferomedial aspect of the neck (Fig. 5B). FE-predicted strain magnitudes were lower than those measured experimentally, and MRI-based models exhibited the lowest strain values, consistent with their higher predicted whole-bone stiffness.

All MRI-based FE predicted measures differed significantly from their CT-based counterparts, except when homogeneous material properties were assigned (Fig. 6, Table 2). Compared with CT, MRI-based models had higher median elastic modulus (+266%, $p < 0.001$), higher predicted stiffness (+72%, $p = 0.002$), higher predicted failure load (+135%, $p = 0.032$), and lower 90th-percentile minimum and maximum principal strains (−39% and −51%, respectively, both $p = 0.002$). Also, MRI-based model predictions did not correlate with those from the CT-based models (Fig. S9). Examination of histograms of element-wise elastic modulus for all specimens revealed that MRI-based models

had a lower maximum assigned modulus, with more elements in the mid-to-upper modulus range compared to CT-based models (Fig. 7A). Similarly, MRI-based models exhibited greater frequencies of low compressive and tensile strains and fewer high-strain values relative to the CT-based models (Fig. 7B–C). Paired, though qualitative, comparisons of internal strain distributions revealed that these differed between the two models, with CT-based FE models exhibiting high compressive and tensile strains within the greater trochanter that were not apparent in the MRI-based models (Fig. S10). In contrast, under homogeneous material assignment, MRI-derived predictions of stiffness, failure load, minimum principal strain and maximum principal strain measures were no different from their CT counterparts (Fig. 6, Table 2).

In the CT-based models, failure was predicted to occur in compression either on the surface of the femoral neck ($n = 6$) or internally ($n = 3$) (Fig. 8). The internal failures were within the greater trochanteric region ($n = 2$) or near the femoral head ($n = 1$). In contrast, MRI-based FE models consistently predicted failure to occur only at surface locations, with one failure occurring in tension.

4. Discussion

This study evaluated the feasibility of using PD-weighted MRI-based models of the human proximal femur to simulate a sideways fall loading scenario. We found no correlation between the MRI-based FE predictions of stiffness or strength and the experimentally measured values. In contrast, CT-based FE models demonstrated good agreement with experiments, consistent with prior literature [38,45,46]. Not surprisingly then, the FE-predicted mechanical measures from the PD-weighted MRI-based models differed from those from the CT-based models of the same femora, though the strain distributions predicted by both model types agreed qualitatively with the measured strain distributions. Strain concentrated at the superolateral and inferomedial regions of the femoral neck, with predominant compressive failure in the mid-neck and trochanteric fossa, as has been seen in prior reports [41]. When the contribution of the different material mapping methods was removed by assigning homogeneous material properties throughout the models, no differences between the MRI- and CT-based models were observed. Collectively, these findings indicate that PD-weighted MRI qualitatively captures the femoral geometry sufficiently to simulate the overall structural mechanics of sideways fall but is limited in predicting quantitative mechanical responses due to insufficiency of the BV/TV-modulus material mapping method.

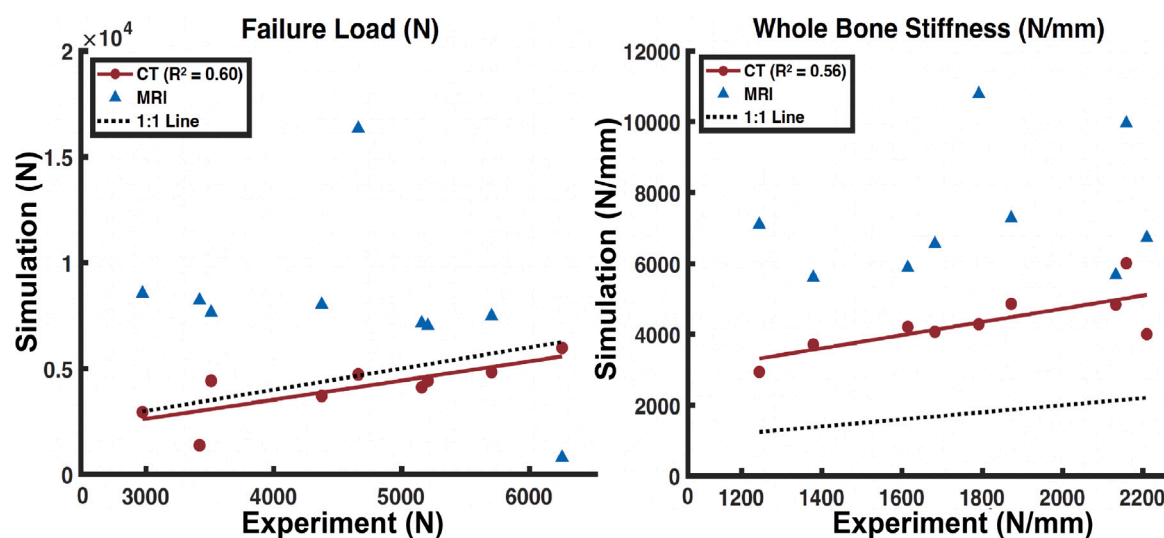


Fig. 4. Comparison of MRI- and CT-based model predictions versus experiment for failure load (A) and whole-bone stiffness (B) for the paired specimen. Linear regression is shown for CT-based models only.

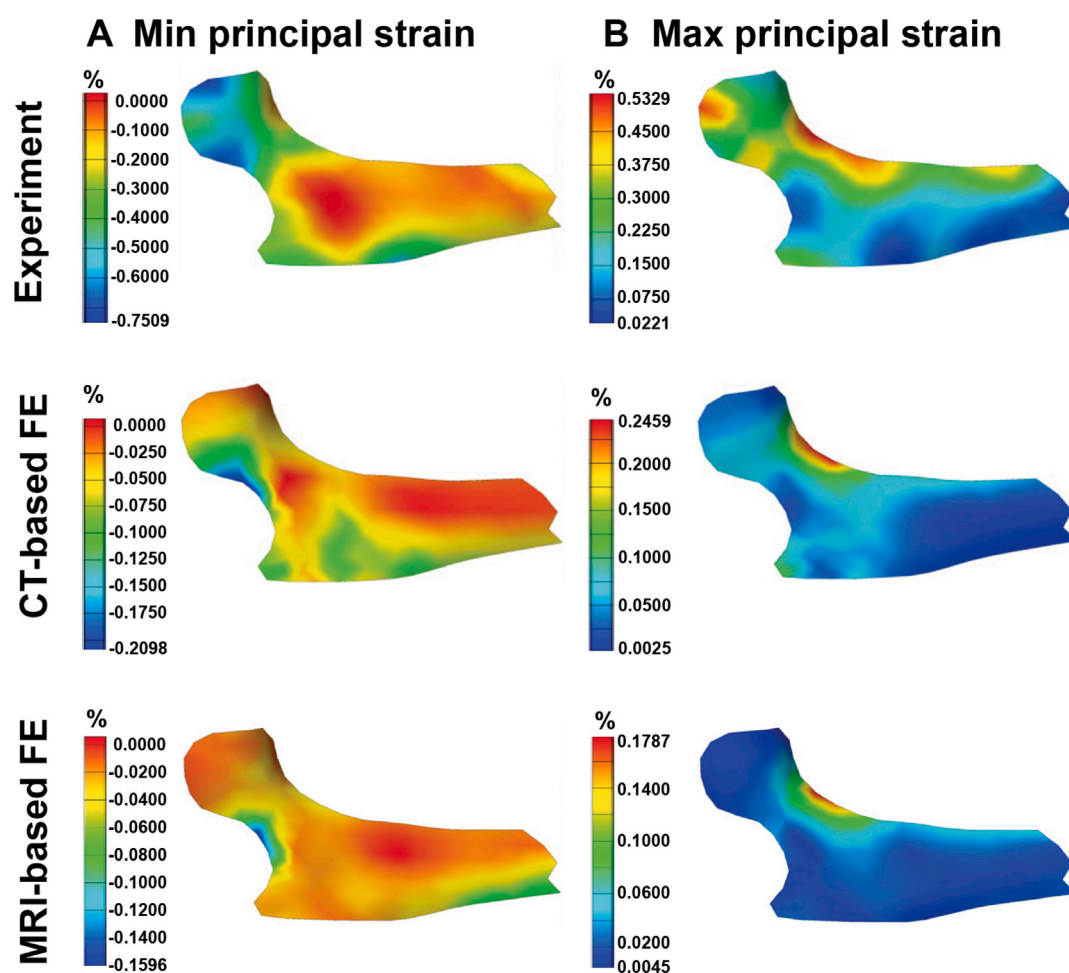


Fig. 5. Result of minimum principal strain (A) and maximum principal strain distribution (B) comparison between experiment (first row), CT-based (second row), and MRI-based (third column) FE simulation of sideways fall of the same representative specimen. Note the difference in scale bar for each measure.

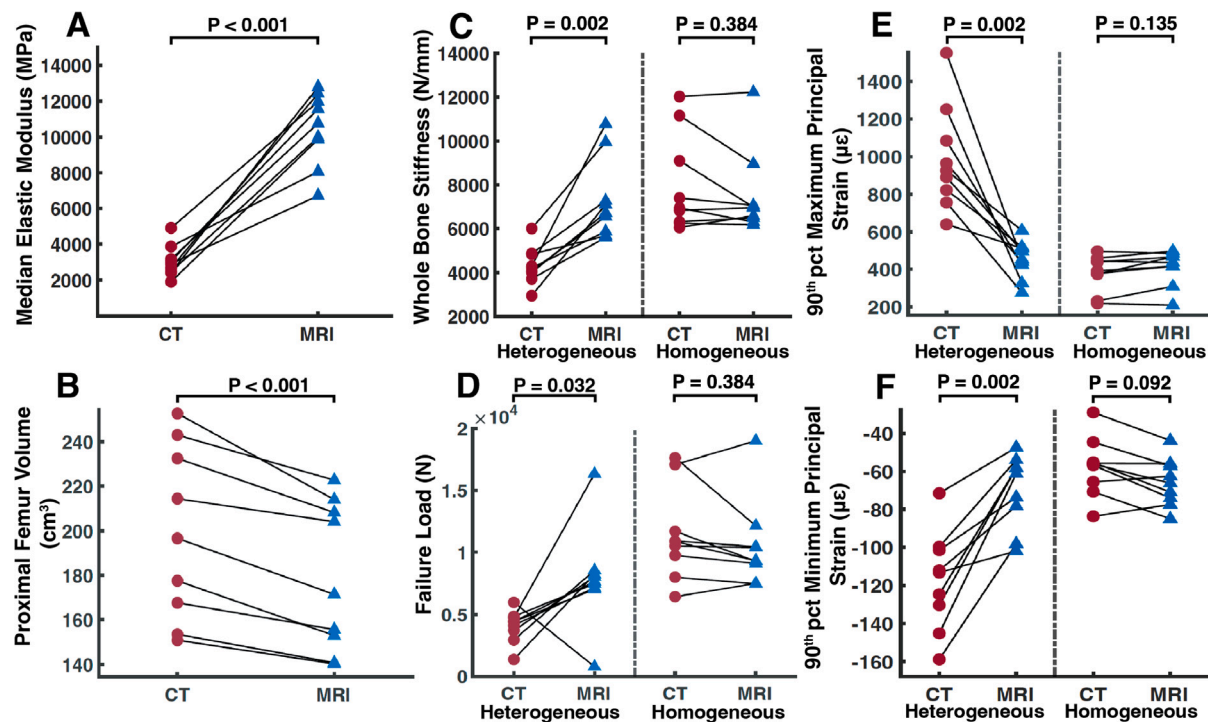


Fig. 6. Paired comparison of MRI- and CT-based model-derived median elastic modulus (A) and proximal femur volume (B), and their corresponding FE-predicted whole-bone stiffness (C), failure load (D), 90th percentile maximum principal strain (E), and 90th percentile minimum principal strain (F).

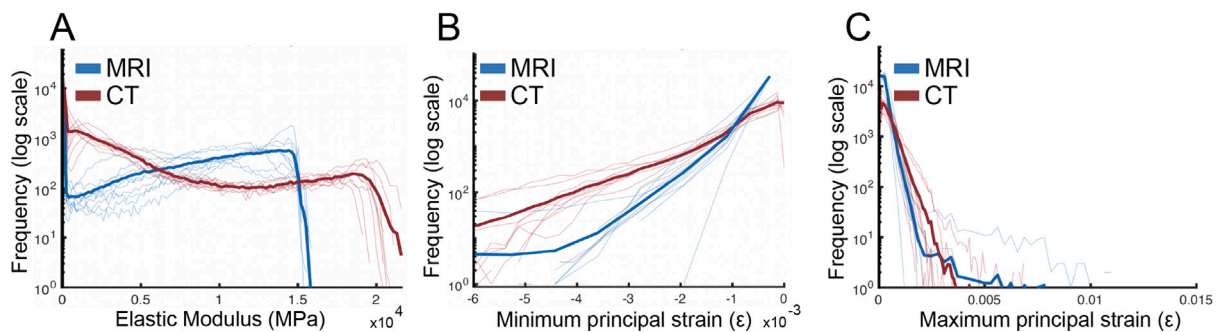


Fig. 7. Element distributions of elastic modulus (A), minimum principal strain (B), and maximum principal strain (C) for paired CT (red) and MRI (blue) specimens. Mean histogram denoted as a bold line.

Table 2

Comparison of MRI- and CT-based FE measures. Data are reported as mean $\pm$ stdev (range). Mean percent differences (% diff) and corresponding p-values from paired t-tests comparing MRI and CT models are shown separately for heterogeneous and homogeneous models.

Measures	MRI-based FE	CT-based FE	Heterogeneous		Homogeneous	
			% diff	p-value	% diff	p-value
Proximal femur volume [cm ³]	178.9 $\pm$ 33.3 (140.2 – 222.7)	198.7 $\pm$ 38.8 (150.8 – 252.7)	–9.8 (–15.3 – –4.8)	<0.001	–	–
Median elastic modulus [MPa]	10455 $\pm$ 2036 (6715 – 12777)	3056 $\pm$ 889 (1917 – 4912)	+266 (–424 – 108)	< 0.001	–	–
Whole-bone stiffness [N/mm]	7273 $\pm$ 1859 (5602 – 10781)	4326 $\pm$ 855 (2942 – 6008)	+72 (17 – 152)	0.002	–4.7 (–23 – 8.8)	0.384
90th pct min principal strain [μ ϵ]	–70 $\pm$ 19 (–102 to –48)	–117 $\pm$ 26 (–159 to –72)	–39 (–58 to –10)	0.002	+18 (–7 – 50)	0.092
90th pct max principal strain [μ ϵ]	451 $\pm$ 101 (276 – 606)	987 $\pm$ 277 (640 – 1551)	–51 (–74 to –20)	0.002	+9 (–5 – 33)	0.135
Failure load [N]	7912 $\pm$ 3935 (789 – 16336)	4057 $\pm$ 1302 (1374 – 5979)	+136 (–87 – 498)	0.032	–6 (–31 – 17)	0.384

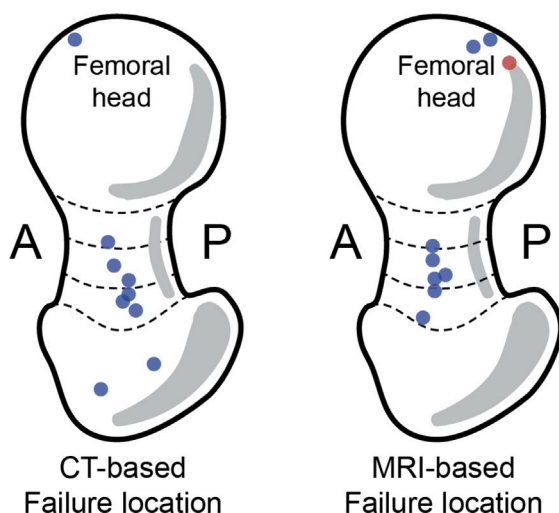


Fig. 8. MRI- and CT-based FE-predicted failure locations represented on a canonical schematic top-view of a left proximal femur (A: Anterior, P: Posterior) with color specifying failure mode (tension in red, compression in blue). Failure was predicted to occur predominantly in compression on the superolateral aspect of the proximal femur, with the exception of one CT-based model and three MRI-based FE models.

This study is the first to investigate the use of PD-weighted MRI scans to assess bone stiffness, strength, and failure location of the proximal femur under sideways fall. PD-weighted MRI has been of interest for assessment of bone because total water proton density has been shown to correlate well with bone microstructure [19–22,47]. While prior studies have investigated MRI-based FE modeling of the proximal femur [12,14,29], this is the first study to directly perform a paired comparison with the current standard, CT-based FE modeling, while also providing experimental validation for both modalities.

The experimental measurements and CT-based FE predictions reported here agree with prior literature, whereas the MRI-based FE results show mixed agreement with prior studies. The experimentally measured ranges of whole-bone stiffness and failure load match previously reported values from sideways fall tests [33,46,48–51], supporting their use to benchmark the FE models. The CT-based FE predictions and their strong correlation to experiments are consistent with many prior CT-based studies [33,38,44,52–54]. This agreement is expected given the well-established material mapping strategies and modeling frameworks underlying CT-based FE methods. In contrast, although a prior study [12] reported good agreement between MRI-based predictions and experimental measurements, no correlation was found here. In addition, the predicted whole-bone stiffnesses reported here are higher than those reported by Rajapakse et al. [12], but overlapped with another study, though with less extreme principal strains observed here [29]. A likely source of this discrepancy is our use of PD-weighted MRI, which depicts bone tissues differently compared to T2-weighted sequences used previously and can alter material property assignments and, in turn, mechanical response [55]. Differences in modeling approaches and boundary conditions may also drive the discrepancies. For example, Rajapakse et al. [12] used a voxel-based nonlinear FE formulation, which captures local compliance and yielding, thereby reducing apparent stiffness, whereas we used a linear elastic model. Similarly, Zhang et al. [29] fully constrained selected nodes at the greater trochanter, while we applied a roller contact constraint, which can affect predicted mechanical responses.

The inherent differences between PD-weighted MRI and CT imaging modalities, which necessitated distinct material mapping methods, drove the observed disparities in model predictions. This conclusion is supported by the results from the homogeneous material mapping

simulations, which showed no difference in failure load or stiffness between the two imaging modalities. Furthermore, the overestimation of the median elastic modulus in the MRI-based models ($p < 0.001$), as shown in Fig. 7, also contributed to the observed differences in mechanical response, as evidenced by the higher predicted whole-bone stiffness. Interestingly, the predicted failure locations and modes by MRI-based FE were consistent with those from the CT-based models and prior reports [41]. This is likely because failure patterns are primarily governed by femoral geometry and loading conditions, which were identical across both types of models. Although both models exhibited strain concentrations in similar regions of the femoral neck compared to experiment, slight spatial discrepancies were observed. These discrepancies could be attributable to experimental factors such as lighting conditions, speckle-pattern quality, or limitations of GOM Correlate in resolving femoral boundaries. Some outliers were observed in both models (Fig. 8). Although localized high compressive strain near the femoral head fovea has been previously reported [44,52], these outliers may have resulted from inadvertent inclusion of cartilage or soft tissue, which deforms more than surrounding bone, or from limitations of the strain-based linear failure criterion.

This work also had limitations. First, this study employed *ex-vivo* MRI scans, and the soft tissues surrounding the proximal femur *in-vivo* imaging were absent. Absence of soft tissues can lead to reduce signal-to-noise ratio, which would contribute to variations in grayscale values that further manifest as variations in assigned elastic modulus, thereby influencing FE predictions. Prior *ex-vivo* DXA studies have used soft-tissue surrogates such as rice bags to mimic soft tissue attenuation [56]; however, such approaches are neither directly transferable to MRI nor standardized. This study was also limited to a relatively small sample size. Nevertheless, the paired study design controls for inter-specimen variability, increasing statistical power and supporting the robustness of the observed differences. An additional limitation is that the boundary conditions assumed a simplified representation of sideways fall [34]. Even though other portrayals of sideways fall have been used [36,57], there is debate over which is most physiologically representative [58–60]. The conditions used in this study were deemed acceptable to allow for verification of results with prior FE modeling and mechanical testing studies of the proximal femur [33,34,38]. Mesh convergence was assessed using a representative mid-cohort specimen (48-year-old female) from which an element side length of 3 mm was selected. While this provides a reasonable estimate of mesh sensitivity, mesh requirements may differ in older specimens with thinner cortices. Future work could evaluate mesh sensitivity in older specimens and explore integration-point-based material mapping as a potential approach to better capture variations in material properties within elements [61]. Lastly, all simulations assumed linear elastic materials and consequently, assessment of bone strength was limited by the use of a linear failure criterion that extrapolates the elastic response. We note that this approach has been shown in prior studies to provide accurate predictions of failure load [38]. Further, we mitigated concern over the use of linear models by defining the measured failure load as the force that corresponds to the end region used to compute the linear slope.

In summary, although MRI-based FE modeling potentially offers a non-invasive, radiation-free assessment method, CT-based FE modeling remains the gold standard for image-based assessment of the mechanical behavior of the proximal femur. We found that PD-weighted, MRI-based FE-predicted mechanical measures differed from CT-based predictions and showed poor correlation to experiment, but they generally identified failure location and mode in a manner consistent with the CT-based models. The findings suggest that MRI-based segmentation is sufficient to capture patient-specific geometry of the femur, and that geometric features largely govern the failure location in the proximal femur during a sideways fall. However, the simple linear mapping of PD-weighted image intensity to modulus limits the accuracy of predicted femoral strength and stiffness. Thus, further investigation of MRI-based material mapping is necessary to expand opportunistic assessment via image-based FE modeling, in patient populations for whom MRI, rather than CT, is the primary modality of clinical screening.

CRediT authorship contribution statement

Benjamin D. Olowu: Writing – review & editing, Writing – original draft, Visualization, Validation, Software, Methodology, Investigation, Formal analysis, Data curation. **Joshua D. Auger:** Writing – original draft, Software, Methodology, Investigation, Conceptualization. **Elise F. Morgan:** Writing – review & editing, Supervision, Resources, Project administration, Funding acquisition, Conceptualization.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Appendix A. Supplementary data

Supplementary material related to this article can be found online at <https://doi.org/10.1016/j.bone.2026.117973>.

Data availability

Data will be made available on request.

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