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Validation of a new multiscale finite element analysis approach at the distal radius

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ABSTRACT

High-resolution peripheral computed tomography is commonly used to evaluate mechanical behavior of the distal radius microstructure using micro-finite element (FE) modeling. However, only a 9 mm section is considered and boundary conditions (BCs) are usually simplified (platen-compression), and may not represent physiologic loading. Regardless, these methods are increasingly being used for clinical evaluations. Our goal was to develop and validate a novel multiscale solution that allows for physiologically relevant loading simulations (such as bracing during a fall), and show that mechanical behavior in the distal radius is different under platen BCs. Our approach incorporated bone microstructure together with organ-level radius geometry, by replacing matching continuum regions with micro-FE sections in user-defined regions of interest. Multiscale model predicted strains showed a strong correlation and a significant relationship with measured strains ($r = 0.836$, $p < 0.001$; slope = 0.881, intercept = −12.17 με, $p < 0.001$). Interestingly, platen BC simulated strains were almost 50% lower than measured strains ($r = 0.835$, $p < 0.001$), and strain distributions were clearly different. Our multiscale method demonstrated excellent potential as a computationally efficient alternative for observing true mechanical environment within distal radius microstructure under physiologically accurate loading.

1. Introduction

Computed tomography (CT) and finite element (FE) analysis are established tools used to investigate mechanical behavior at the distal radius under various conditions. Typically, the FE analyses involve organ- (macro) level bone geometry acquired from CT images at clinical-resolution (voxel size 234 × 234 × 625 μm or greater), to determine in situ strains [1-3], fracture strength [4-6], and the efficacy of fracture fixation methods [7-9]. The relatively lower resolution of the CT images provides insufficient structural detail through the cross-section. To account for this limitation, the FE analyses are restricted to a continuum assumption with inhomogeneous material properties related to density distribution, derived from voxel intensities of the images [10-12]. Depending on the application, distal radius continuum FE models provide useful information on cortical strains and integral whole bone mechanics. Bone however, is a complex structure consisting of cortical (low porosity) and trabecular (highly porous) regions each playing a unique role in whole-bone mechanics [13,14]. The microstructure of the trabecular region is particularly important to take into consideration because of its increased susceptibility to bone loss and osteoporosis [15,16], and its age-related changes in anisotropy [17].

High-resolution peripheral quantitative CT (HR-pQCT) is able to resolve bone microstructure (82 × 82 × 82 μm voxel size), making the non-invasive evaluation of cortical and trabecular microstructure at the human distal radius possible [18,19]. Due to the significantly longer scan times (~20 times > clinical CT), standard HR-pQCT protocols only consider a volume of interest typically 9.0 mm in length (110 transverse slices). This scan length sufficiently captures the clinically relevant region of the distal radius where fractures typically occur. HR-pQCT has been applied to study age-related changes, effects of diseases, and outcomes of various treatments on bone microstructure [20-24]. Also, HR-pQCT data are used to generate micro-FE models incorporating microstructure to evaluate distal radius mechanical behavior [25-30]. From the images, voxels of the cortical and trabecular bone regions are directly converted to an FE mesh and assigned homogeneous material properties. However, the micro-FE analyses only model the 9.0 mm thick section of the distal radius and the boundary conditions (BCs) usually simulate a simplified uniaxial compression test between two platens. Essentially, the proximal surface of the bone section is fixed and a uniform axial displacement (or load) is applied to the distal surface [31]. Though these simulations do not represent physiologic conditions, measures from these analyses are increasingly being used as outcomes for clinical trials [22,23,32,33].

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While simplified platen BCs may only result in a systematic variation in parameters compared to physiologic BCs, the effect of these BCs on estimated bone mechanical behavior has not been determined. Furthermore, physiologically relevant BCs are important to understand true mechanical environment within bone, which is associated with structural adaptation [34].

Ideally, FE simulations of the radius would involve micro-FE meshes of the organ-level geometry (incorporating microstructure), non-linear inhomogeneous material properties, simulated under physiologic BCs (such as axial loading through the extended wrist corresponding to most activities of daily living). However, due to time, current computational constraints, and impracticalities associated with acquiring large volumes of HR-pQCT data, these analyses are not possible using data from living human subjects. To address these limitations, a "compromise" approach would involve a micro-macro-level multiscale analysis where only selected regions of interest within organ-level geometry (continuum-FE) incorporate microstructure (micro-FE).

Hence, the goals of this study were twofold. (1) To develop and validate a novel multiscale approach to solve continuum and micro-FE models simultaneously under physiologic boundary conditions. Strain was chosen as an outcome for validation because it is an important mechanical stimulus associated with bone adaptation [35,36]. (2) To compare results of distal radius micro-FE loading simulations performed using physiologic BCs (multiscale models) and simplified platen BCs, versus experimental results. We hypothesized that platen BC simulations would exhibit different mechanical behavior at the distal radius region compared to physiologic BC simulations.

2. Methods

2.1. Specimens

Data were acquired from 10 fresh-frozen cadaveric left forearms with the hand intact (4 males, 6 females; mean age 64 years, range 32–89). The specimens were thawed to room temperature for image acquisition, experimental setup and mechanical testing. While thawed, the specimens were kept moist using a saline solution.

2.2. Image acquisition

All images were acquired using an XtremeCT HR-pQCT scanner (Scanco Medical, Switzerland). Two sets of images were acquired per specimen; clinical-resolution images of a 11 cm region of the distal forearm and wrist in 60° extension ($246 \times 246 \times 246$ µm voxel size, 126 mm field of view, 60 kV, 0.9 mA), and high-resolution images of a 9.0 mm region of the distal radius ($82 \times 82 \times 82$ µm voxel size, 126 mm field of view, 60 kV, 0.9 mA). The high-resolution image region corresponded to the strain gage attachment sites (distal set; see Section 2.3). All CT images were calibrated using a phantom with known calcium hydroxyapatite (HA) equivalent concentrations provided by the manufacturer.

2.3. Specimen preparation and mechanical testing

All soft tissue was dissected off the forearm proximal to the wrist joint capsule, leaving the interosseous membrane intact (Fig. 1). A transverse osteotomy was performed 14 cm proximal to the Lister's tubercle, and the proximal most 7 cm of bone was potted in polyurethane (7 cm exposed). For strain gage attachment, the periosteum was removed and the bone surface was sanded and cleaned with isopropyl alcohol. Three stacked rosettes (C2A-06-031WW-120, Micro-Measurements, Raleigh, NC) were attached circumferentially (anterior, posterior-lateral, posterior-medial) at the distal radius proximal to the Lister's tubercle (Fig. 1B and C). Three additional gages were attached 3 cm proximal to the distal gages in a similar configuration. The location of the distal gages (predominantly trabecular region) was in the vicinity of the clinically relevant site for distal radius fractures, and the proximal gages (predominantly cortical region) provided a greater range of strain measurement. The gages were attached using cyanoacrylate glue and coated with polyurethane.

The experimental setup replicated a fall configuration (Fig. 1A). The wrists were fixed in 60° extension (relative to forearm long axis) using a custom fixture and the potted ends were unconstrained to minimize frictional shear. The specimens were loaded in compression to 300 N at a rate of 0.1 mm/s using a uniaxial materials testing machine (ElectroPuls E1000, Instron, Norwood, MA). The target force of 300 N was chosen on the basis of achieving a range of periosteal strain magnitudes associated with structural adaptation (1000–2000 $\mu\varepsilon$), extrapolated from animal models [37,38]. Force, displacement and strain data were collected simultaneously at 80 Hz. Only 14 analog channels were available for data acquisition, therefore, data were collected from 4 gages per trial (3 channels per gage: $4 \times 3 = 12$ channels; plus 2 channels for force, displacement). A total of 10 trials were performed, allowing for 5 repeat trails per gage (approximately 3 min loading intervals), with data from two gages acquired throughout to assess for variability.

2.4. Modeling

Fig. 2 summarizes the workflow of the modeling procedures involving continuum only models (to determine contact BCs for the multiscale analyses), multiscale models (continuum+micro-FE), and micro-FE only models (for comparison of BCs).

2.4.1. Continuum only model analyses

For time and computational efficiency, cartilage contact BCs for the multiscale model analyses were determined from the corresponding continuum only model analyses. The procedures for creating the continuum models were based on prior validated methods, which were shown to have a root mean squared coefficient of variation of 0.3% for strain measures [1]. Briefly, to acquire model geometries the radius, scaphoid and lunate bones were segmented from the clinical-resolution images in Mimics (Materialise, Leuven, Belgium) using a fixed minimum cortical density threshold ($\rho_{HA} = 0.210$ g/cm^3) for all specimens. The scaphoid and lunate were included in the models for physiologically accurate forearm loading (Fig. 3). Cartilage was created by expanding the articular radius surface to be in contact with the scaphoid and lunate (~1.4 mm cartilage thickness). The bone and cartilage geometries were meshed in 3-matic (Materialise, Leuven, Belgium) using 10-node tetrahedral elements [113,228 ± 26,021 nodes; 302,725 ± 71,671 degrees of freedom (mean ± SD)], with acceptable edge lengths determined from a convergence analysis (2 mm and 3 mm for cartilage and bone, respectively). The cartilage was modeled as a hyperelastic neoHookean deformable solid ($E = 10$ MPa, $\nu = 0.45$) [39], and the scaphoid and lunate were modeled as rigid non-deformable solids. Linear elastic, isotropic ($\nu = 0.4$), inhomogeneous material properties were assigned to the radius based on an established density–elasticity relationship [12]. CT Hounsfield units were converted to ρ_{HA} using the CT phantom calibration and linearly incremented into 200 bins. Elements were assigned a modulus value corresponding to the average ρ_{HA} in each bin (no change in outcome with > 200 elements based on a sensitivity analysis). To account for negative modulus values due to the presence of marrow fat, ρ_{HA} values < 0.01 g/cm^3 were assigned $\rho_{HA} = 0.01$ g/cm^3. To account for lower modulus values at the periosteal surface due to partial volume effects, material assignment was based on ρ_{HA} values restricted to 1 voxel within the segmented radius area. Elements outside this area (<0.3%) were