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Mechanisms governing the inelastic deformation of cortical bone and application to trabecular bone

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Abstract

To understand the inelastic response of bone, a three-part investigation has been conducted. In the first, unload/reload tests have been used to characterize the hysteresis and provide insight into the mechanisms causing the strain. The second part devises a model for the stress/strain response, based on understanding developed from the measurements. The model rationalizes the inelastic deformation in tension, as well as the permanent strain and hysteresis. In the third part, a constitutive law representative of the deformation is selected and used to illustrate the coupled buckling and bending of ligaments that arise when trabecular bone is loaded in compression.

Keywords: Bone; Cortical; Trabecular; Inelastic deformation; Hysteresis

1. Introduction

Many studies of both cortical and trabecular bone have demonstrated their propensity to inelastic strain [1–4] (Fig. 1). Parallel to the long axis, cortical bone exhibits inelastic responses that differ in tension and compression [1–6] (Fig. 1a). In tension, it yields, followed by (linear) hardening up to a failure strain of order 2.5%. The inelastic strain has been attributed to the development of diffuse microcrack arrays [7–9]. Simultaneous measurements of the axial and transverse strains have revealed that such deformation involves dilatation [10], consistent with a role of microcracks. The strains have been characterized by means of a damage model with internal state variables [11]. In compression, cortical bone also yields, but at higher stress than in tension. It strain hardens rapidly to a peak, then softens and fails at strains of about 1.5%. The softening has been attributed to the formation of shear bands [7]. Such behavior is not unique, but akin to the deformation characteristics of nacre [12] and fibrous oxides [13].

When compressed, trabecular bone exhibits extensive inelastic deformation (Fig. 1b), often attaining strains exceeding 60% before failure [4]. A peak stress occurs at strains of 5–10%. The deformations preceding the peak dictate the overall load capacity. Yield surface determinations [14] suggest that inelastic mechanisms are involved. The characteristics of synthetic foams with topology similar to trabecular bone may provide benchmarks. The load capacity of elastomeric/polymer foams is dictated by the elastic buckling of its ligaments [15], while that for metal foams is controlled by yielding [16]. The consequence is that different scaling formulae relate the stress maximum, σ_{max}, to the properties of the ligaments. For isotropic, open cell polymer foams [1]: $\sigma_{max}/E \approx 0.05f^2$, where E is the Young's modulus for the material in the ligaments and f is the volume concentration of the solid in the foam. For isotropic metal foams [16]: $\sigma_{max}/\sigma_Y \approx f^{3/2}$, where σ_Y is the uniaxial yield strength of the material in the ligaments. The differences in these formulae simply underscore the need to confirm that the mechanism governing the load capacity of trabecular bone is inelastic.

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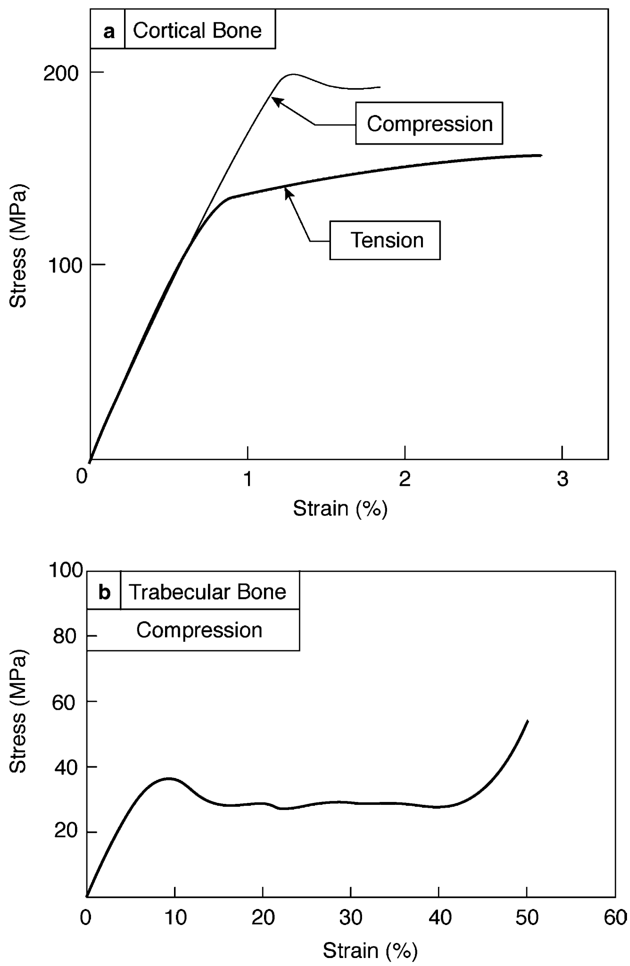

Fig. 1. (a) Schematic of the tensile and compressive stress/strain curves for cortical bone along the axis of a long bone [1]. (b) Schematic of a compressive stress/strain curve for trabecular bone [1].

The present study has three parts. (i) A mechanical probe is used to ascertain the hysteresis associated with the inelastic strains, as well as the elastic modulus evolution. (ii) Motivated by these measurements, a model is presented that highlights the relative roles of the constituent phases (collagen and mineral platelets) and of the interfaces between them. (iii) A constitutive law is invoked that incorporates key aspects of the inelastic deformation. The criterion for the choice is that the law already exists within a commercial finite element code such that, upon calibration, it can be applied immediately to large-scale problems, such as the cell level response of trabecular bone [1].

2. Materials and methods

The cortical bone used in this study was obtained from bovine femurs. Cylindrical, mid-diaphyseal sections (40 mm long) were extracted from fresh femurs (exact age not known, presumably between 2 and 7 years old). Rectangular specimens were cut on a longitudinal/radial plane at the posterior sector of the femur, with a low-speed diamond blade saw under continuous irrigation. The four rectangular surfaces were mechanically ground using silicon carbide media, and then polished with diamond suspensions, followed by 0.05 μm alumina suspension on a precision machine (MULTIPOL 2 Precision Polishing Machine, Ultra Tec, Santa Ana, USA). The final specimens were 38 mm long, 3.85 mm wide (across the thickness of the bone), and 3.05 mm (four specimens) or 1.52 mm (six specimens) thick. The specimens were stored in water at 4 °C for 1–15 days prior to testing.

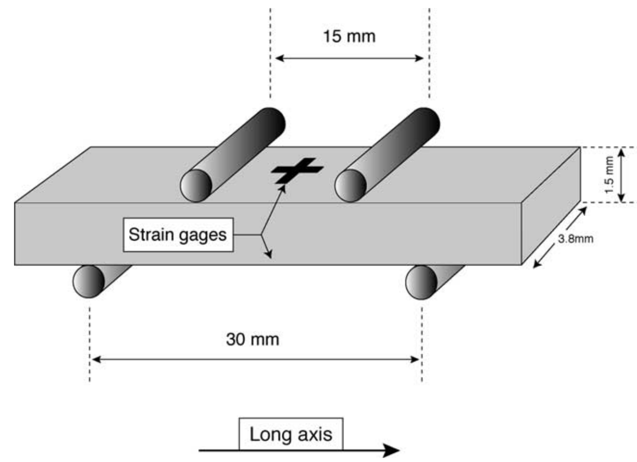

Fig. 2. Schematic of the instrumented flexure test showing the locations of the strain gages.

Four-point bend tests (Fig. 2) were conducted using a servo-hydraulic testing machine at a cross-head speed of 0.5 mm/min. The inner loading span was 15 mm and the outer support span 30 mm (Fig. 2). The upper and lower loading surfaces were chosen to be perpendicular to the lamellar boundaries of the cortical bone to minimize the structural variations across the thickness of the specimens and therefore, to allow comparison between the compressive and tensile behavior. To measure the strains on both tensile and compressive surfaces, strain gages (from either Omega Engineering or Vishay Micro-Measurements) were applied to both surfaces (one on each side) using M-Bond 200 adhesive following degreasing and cleaning. In some cases, 0°/90° strain gages were used for measurement of the transverse strains in addition to axial strains. Specimens were either monotonically loaded to failure, or repeatedly unloaded/reloaded at incrementally higher plastic strains, in order to measure hysteresis. The specimens were kept hydrated during testing by using an agarose gel (Sigma–Aldrich, USA). The inelastic deformation has been reproducibly established using six specimens.

After mechanical testing, the specimens were examined with a stereomicroscope for macro-scale damage. For this purpose, some specimens were stained with basic fuchsin [17]. The stained specimens were embedded in a mounting medium. Using a low-speed diamond saw, thin slices were cut along the longitudinal direction of the specimens, with each slice either parallel or perpendicular to the compressive (upper loading) surface. The slices were polished to