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DEPTH-DEPENDENT TENSILE AND COMPRESSIVE PROPERTIES OF HUMAN PATELLOFEMORAL JOINT CARTILAGE

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INTRODUCTION

It is well established that cartilage composition and structure vary through the depth of the tissue. Concurrently, it has been observed that the mechanical properties of cartilage also vary through the depth. The tensile stiffness of normal cartilage in the superficial tangential zone (STZ) has been reported to be significantly greater than in the deep zone (e.g., [1]), by up to a factor of 20. In a series of recent studies [2-4], significant inhomogeneity of cartilage has also been reported in compression, with the compressive modulus smallest in the STZ and highest at the deep zone. Interestingly however, the inhomogeneity in both tensile and compressive properties has seldom been reported in the same tissue sample. Much interest has been focused on the disparity in tensile and compressive properties of articular cartilage in recent studies [5,6], and on the role that inhomogeneity might play in the mechanical response of the tissue [7].

The objective of the current study was to simultaneously measure the tensile and compressive properties of human patellofemoral joint cartilage as a function of depth, by performing unconfined and confined compression stress-relaxation experiments on cylindrical plugs of cartilage serially sliced through the depth. Because unconfined compression subjects cylindrical samples to tensile stresses in the radial and circumferential directions, it can be used to extract tensile properties, while confined compression yields compressive properties of the tissue.

MATERIALS AND METHODS

Specimen Preparation

Visually normal cylindrical cartilage samples (diam. 4.78mm) were harvested from the articular layers of the anterior condyles of the femur and the center of the patella from three fresh frozen human cadaver knees (2 males, 1 female, ages 37, 44, and 61; thickness 2.32±0.16 mm). Using a sledge microtome, about 0.2 mm of tissue was removed from the deep zone to produce a surface parallel to the articular side. Each sample was then serially sliced into four layers starting from the deep zone; the first three slices were approximately of the same thickness (0.50±0.09 mm), while the most superficial slice consisted of the remainder (0.81±0.20 mm). All specimens were stored at –80^{0}C prior to testing. On the day of testing, each slice was thawed at room temperature and equilibrated in PBS solution. The testing protocol, consisting of unconfined and confined compression stress relaxation tests, followed that of our recent study [6].

Testing Apparatus

The testing apparatus consisted of a stepper micrometer for displacement actuation (Model 18500, Oriel Instruments, Stratford, CT, step size 0.7-1.2μm) connected in series with a linear variable differential transformer (LVDT) for measuring displacements (HR 100, Schaevitz Sensors, Hamptons, VA) and a uniaxial load cell for measuring the reaction force (Sensotec, Columbus, Ohio model GM). Data acquisition and control was performed via a personal computer equipped with a data acquisition card (National Instruments), running the Labview software. All tests were conducted in 0.15M physiological buffered saline (PBS) solution at room temperature.

Loading Protocol

For unconfined compression with impermeable platens, a tare load equivalent to 12 kPa was first applied; following equilibrium, a stress relaxation test was performed by ramping the top platen at constant velocity until 10% tissue strain was achieved within 500 seconds. The reaction force was monitored as a function of time until equilibrium was achieved (~4,300 s). At the completion of the test, the specimen was unloaded and allowed to recover for as much time as the test duration. For confined compression experiments, the specimen was placed in a confining chamber, 4.78mm in diameter, with an impermeable bottom surface. Loading was applied on the top surface via a porous indenter 4.76mm in diameter. An identical loading protocol as for unconfined compression was used.

Determination of Material Properties

The tensile and compressive properties of the tissue were obtained by curvefitting the experimental data with the biphasic-CLE theory [6]. This theory extends the biphasic model of Mow et al. [8] by incorporating the conewise linear elasticity theory of Curnier et al. [9] to describe the tension-compression nonlinearity of the tissue. For cubic symmetry, its material constants consist of the aggregate moduli in compression (H_{-A}) and tension (H_{+A}), the shear modulus (μ), the off-diagonal modulus (λ_2), and the axial (k_z) and radial (k_r) permeabilities of the solid matrix. Least-squares curvefitting of the confined compression test was performed first to yield the values of H_{-A} and k_z. Then, curvefitting of the unconfined compression test provided the values for H_{+A}, λ_2 and k_r. The shear modulus cannot be determined from these tests and is not reported here.

RESULTS AND DISCUSSION

Twenty-four specimens were tested in total for these experiments. The depth-dependent results for H_{-A} and λ_2 are presented in Figure 1 for both femur and patella; H_{+A} is presented in Figure 2, and k_z, k_r in Figure 3. H_{-A}, is observed to be smallest in the most superficial zone and largest in the deep zone, consistent with the studies of Schinagl et al. [3] on bovine articular cartilage. However, the concavity of the depth-dependent variation is opposite to that observed in that earlier study. λ_2 also achieves its smallest value in the most superficial zone, increasing with depth. Conversely, H_{+A} is largest in the superficial zone and decreases with increasing depth; this result is also consistent with previous reports of the depth-dependent tensile modulus of human cartilage [1]. The ratio of tensile to compressive moduli is largest in the most superficial zone smallest in the deepest zone. This ratio regulates the magnitude of interstitial fluid load support in cartilage [6], with greater fluid load support achieved at higher ratios. Poisson's ratio in compression is given by $\nu_- = \lambda_2 / (\lambda_2 + H_{+A})$; based on the data of Figures 1 & 2, ν_- is smallest near the surface zone and increases (nearly linearly) with depth, a result consistent with direct measurements [4]. Finally, it is interesting to observe that tissue permeability exhibits less of a dependence with depth than the elastic properties of the matrix. Depth-dependent measures of the permeability have seldom been reported in the literature.

The results of this study provide a useful set of depth-dependent material properties, in tension and compression, for human patellofemoral joint articular cartilage. Differences in the properties of femoral and patellar cartilage [10] were not addressed specifically in this study but may be investigated with a larger sample population. Knowledge of the material properties of cartilage is essential for the determination of stresses within the tissue. In future studies, it would be of interest to determine whether the inhomogeneity in cartilage properties is meant to produce a more homogeneous distribution in tissue stresses through the depth, under physiologic loading, as has been suggested for other biological soft tissues.

ACKNOWLEDGMENTS

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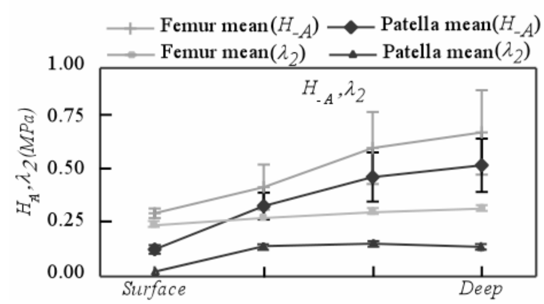

Figure 1

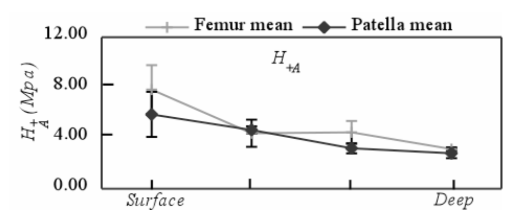

Figure 2

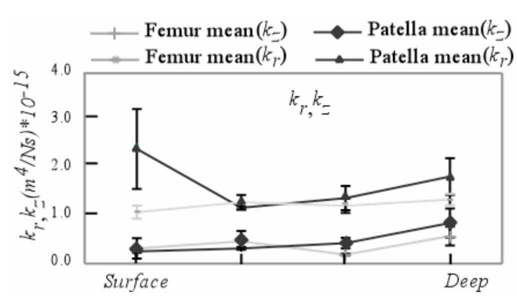

Figure 3