

CARTILAGE INTERSTITIAL FLUID LOAD SUPPORT IN UNCONFINED COMPRESSION

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INTRODUCTION

Under physiological conditions of loading, articular cartilage is subjected to both compressive strains, normal to the articular surface, and tensile strains, tangential to the articular surface. Previous studies have shown that articular cartilage exhibits a much higher modulus in tension than compression. Theoretical analyses have suggested that this tension-compression nonlinearity enhances the magnitude of interstitial fluid pressurization during loading in unconfined compression, above a theoretical threshold of 33% of the average applied stress. The first hypothesis of this experimental study is that the peak fluid load support in unconfined compression is significantly greater than the 33% theoretical limit predicted for porous permeable tissues modeled with equal moduli in tension and compression [1]. The second hypothesis is that the peak fluid load support is higher at the articular surface side of the tissue samples than near the deep zone, because the disparity between the tensile and compressive moduli is greater at the surface zone.

MATERIALS AND METHODS

Cylindrical cartilage samples (6 mm diam.) were harvested from two different sources. In Group I, eleven samples were harvested from visually normal or mildly fibrillated regions of the femoral condyles and retropatellar surface of six human cadaver knee joints (4 males, 2 females, ages 30-61; thickness 1.49 ± 0.23 mm). In Group II, eleven samples were harvested from the femoral surface of three 2-4 month-old healthy bovine calf knee joints (thickness 1.96 ± 0.07 mm) obtained from a local abattoir. The testing apparatus [2] employed a stepper micrometer to provide displacement actuation, connected in series with a LVDT for measuring displacements, and a uniaxial load cell (load limit 44.50 ± 0.09 N) for measuring the total reaction force across the loading platens. Measurements of interstitial fluid pressure were performed using a piezoresistive microchip pressure transducer (NPC1210-100G-3N, Lucas NovaSensor, Fremont, CA; range 0-690 kPa gauge pressure). To measure the load supported by interstitial fluid pressurization, a 4.78 mm diam. \times 1.50 mm deep recess was created at the bottom of the testing chamber to hold a free-draining

porous filter flush with the bottom surface of the testing chamber. A 1 mm diameter hole below the center of the porous filter communicated with the 1 mm diam. \times 1 mm deep well of the piezoresistive pressure transducer.

Two types of displacement control tests were employed in this study. In each group, ten specimens were loaded by imposing a relatively rapid ramp displacement on the top platen to reach 20% compression in 100 s, immediately followed by unloading at the same rate. This triangular displacement profile test (Experiment 1) was used to determine the peak fluid load support achievable in these specimens during the loading phase. For the remaining eleventh specimen in each group, a more standard stress-relaxation test was performed (Experiment 2), where the top platen was displaced by 10% of the specimen thickness over 500 s, then maintained constant until stress-relaxation equilibrium was reached. This test was used to verify whether the fluid load support does indeed subside to zero at equilibrium, as observed in previous studies. In both groups, each of the eleven samples was tested twice, once with its articular surface facing the pressure transducer, and once with the surface nearer to the deep zone facing the transducer. The ordering of these two tests on each specimen was randomized.

RESULTS

The raw data for the total load, W , and the load supported by interstitial fluid pressurization, W^p , from a typical human specimen in Experiment 1 is presented in Figure 1a as a function of time, starting from the completion of the rapid tare loading. Both W and W^p rise in magnitude with a concave profile during the compressive phase of loading, and reduce in magnitude during the unloading phase. When plotting the same data as W^p against W during the compressive phase of loading, the observed response is very nearly linear, Figure 1b. The slope of this linear response, dW^p/dW , which is a measure of how much W^p increases with increasing W irrespective of their initial tare values, can thus be used to calculate the fluid load support for this test. Standard linear regression was employed to evaluate this slope. For

Group I (human specimens), the mean±standard deviation of the peak fluid load support was 79%±11% at the articular surface and 69%±15% near the deep zone, whereas for Group II (bovine specimens) it was 94%±4% at the articular surface and 71%±8% near the deep zone. All four values were significantly greater than 33%. In both groups, the fluid load support was significantly greater at the articular surface than near the deep zone, with $p < 0.05$ for Group I and $p < 0.0001$ for Group II. The fluid load support during the stress-relaxation test (Experiment 2) is shown in Figure 2, confirming that it is initially elevated but subsides to zero at equilibrium.

DISCUSSION

The objective of this study was to verify from experiments that interstitial fluid pressurization supports most of the load transmitted across articular layers even in the case of unconfined compression, despite earlier theoretical predictions that placed a ceiling of 33% on the fluid load support in this testing configuration. The results support this hypothesis, along with the hypothesis that the fluid load support is greater at the articular surface than near the deep zone, for both human and bovine tissue. As derived in our earlier studies [3,4], the peak fluid load support is dependent on the ratio of the tensile and compressive moduli of the tissue according to the formula

$$\left. \frac{W^p}{W} \right|_{\max} = \frac{1}{1 + 2 \frac{H_{+A} - \lambda_2}{H_{+A} - \lambda_2}}$$

where H_{-A} is the aggregate modulus in compression and H_{+A} is the corresponding modulus in tension; λ_2 is the “off-diagonal” modulus ($\lambda_2 < H_{-A}$). In the limiting case of $H_{+A} - \lambda_2 / H_{-A} - \lambda_2 = 1$, this formula predicts a peak fluid load support of 33%; however, when $H_{+A} - \lambda_2 / H_{-A} - \lambda_2 \gg 1$, the peak fluid load support approaches 100%. The physical explanation behind this theoretical prediction is premised on the recognition that cartilage is subjected to both compressive strains (in the axial loading direction, normal to the articular surface) and tensile strains (in the radial and circumferential directions, tangential to the articular surface), in unconfined compression as well as in physiological loading conditions. If the solid matrix compressive modulus is much lower than the tensile modulus, the tissue will strongly resist radial and circumferential expansion upon rapid axial loading, and the only way that the tissue can maintain its volume is for the interstitial fluid to pressurize substantially to prevent large compressive strains in the axial direction. The peak fluid load support of 94% observed in this study suggests that the ratio of $H_{+A} - \lambda_2$ to $H_{-A} - \lambda_2$ is greater than 30 to 1 near the articular surface of bovine cartilage, whereas 79% fluid load support suggests a ratio of 7.5 to 1 near the articular surface of human patellofemoral joint cartilage. Near the deep zone, the corresponding ratios are 4.9:1 in bovine cartilage and 4.4:1 in human cartilage. The depth-dependent differences in the ratio of tensile to compressive moduli are consistent with literature findings that demonstrate a decreasing tensile modulus and an increasing compressive modulus from the superficial to the deep zone of cartilage [5,6,7,8]. These results confirm that the disparity in tensile and compressive moduli of cartilage is an essential functional property of the tissue, without which the interstitial fluid could not contribute as significantly to load support and dynamic stiffening. The greater interstitial fluid load support at the articular surface represents a functional optimization which promotes a low friction coefficient for cartilage.

ACKNOWLEDGMENTS

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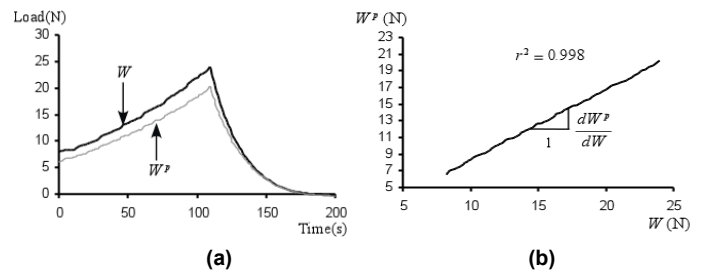


Figure 1: Typical response of a specimen in Experiment 1. (a) Total load W and fluid load W^p versus time; (b) W^p versus W .

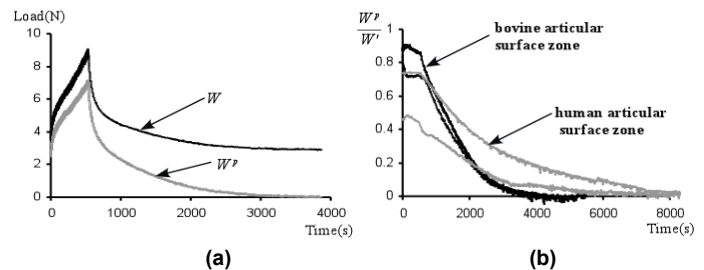


Figure 2: (a) Typical response of a specimen in Experiment 2. (b) Fluid load support for all tests of Experiment 2.