Rate Dependent Deformation Response of Articular Cartilage

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Editor's Note: For a closer look at Edward's poster abstract, be sure to check out his short video presentation in the October digital version of TLT (available at **www.stle.org**).

INTRODUCTION

Osteoarthritis (OA) is a leading cause of severe disability in the United States. OA is characterized by the failure of articular cartilage, which in many cases provides a low friction, low wear interface in the body's joints over decades of use. Many researchers have shown that cartilage's load bearing and lubrication properties can be attributed to its biphasic structure. Interstitial fluid pressure can carry a substantial portion of the contact force, and the friction coefficient is linearly proportional to the portion of the load supported by the elastic matrix.

In a typical experiment, fluid pressure drops and friction increases as fluid vacates the contact over time. Sustainable lubrication was first discovered in 2008 when the traditional glass flat counterbody was replaced by a glass sphere.[1] This mechanism is likely important for retaining fluid pressure in non-conforming diarthrodial joints where one cartilage surface always migrates along the other. The cartilage mechanics community has overwhelmingly studied the theoretically simpler problem of uniaxial compression. This study probes the mechanical response of cartilage to rate-controlled indentation by spherical probes in an effort to relate non-equilibrium mechanics to its functionality.

METHODS

The results shown here were carried out on a plug of bovine cartilage, submerged in a phosphate buffered saline solution obtained from the femoral condyle of 12-20-month-old steer. A custom indentation device, schematically shown in Figure 1, was developed for low force, variable speed indentations with high resolu-

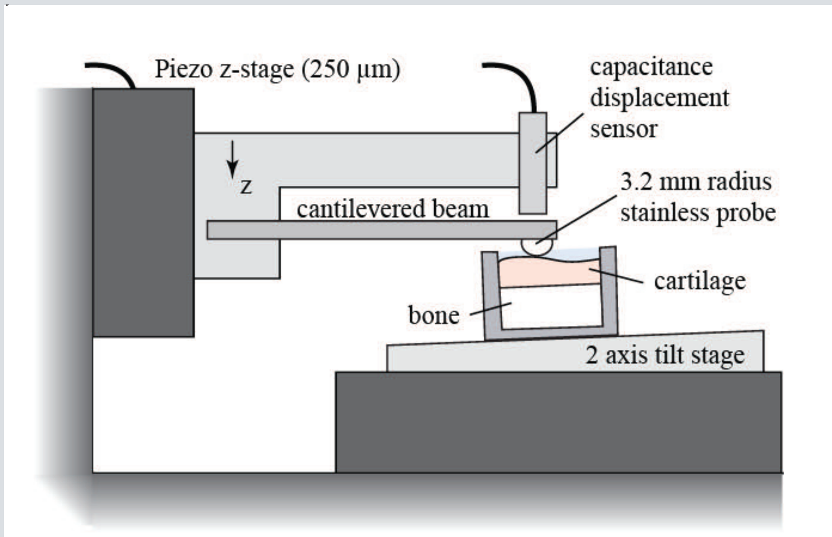

Figure 1 | **Schematic of custom indentation device.**

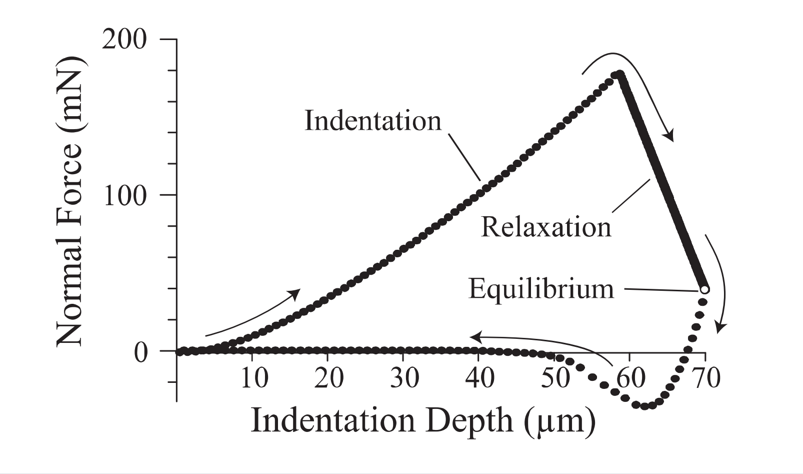

Figure 2 | **Example test consisting of indentation, relaxation and equilibrium. The slope of the relaxation segment is equal to the load cell spring constant.**

tion of indentation depth and force.

Figure 2 shows the basic testing procedure. The test consists of indentation, relaxation (loss of interstitial fluid pressurization) and equilibrium before unloading.

To determine the material response, the force-indentation data is fit to a modified Hertz equation (rigid sphere penetrating an elastic half-space). An effective modulus can be calculated as:

$$E' = \frac{3}{4}\frac{F_n}{R^{0.5}\cdot\delta_s^{1.5}} \qquad \text{(eq. 1)}$$

Where E' is an effective modulus, F_n is the normal force, R is the radius of the probe, and δ_s is the indentation depth.

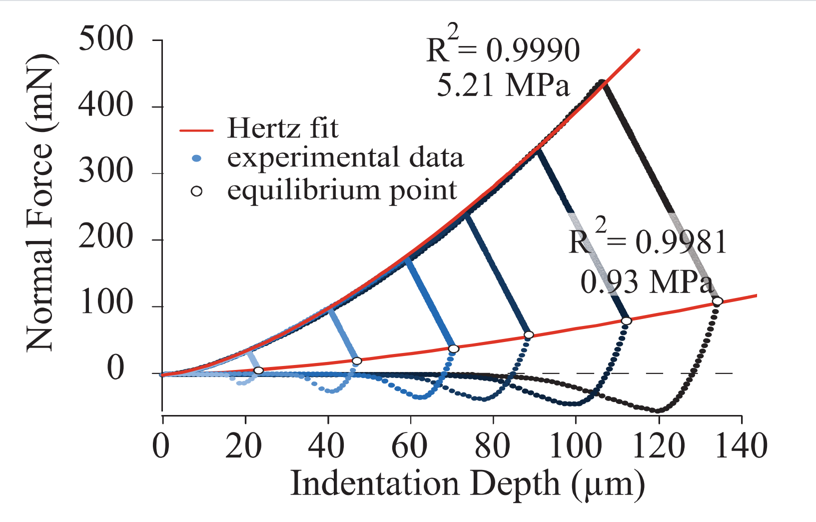

Figure 3 | **Indentation at 8.4 μm/s and equilibration at varying commanded depths compared to Hertzian fits.**

RESULTS

Six depth-varying indentations were made at 8.4 μm/s to study the depth/force and equilibrium/non-equilibrium relationships (*see Figure 3*). Both the equilibrium and indentation responses are in excellent agreement with Hertzian fits (*see equation 1*) despite cartilage being a heterogeneous non-linearly viscoelastic material. The equilibrium and effective moduli are 0.93 and 5.21 MPa, respectively. The tissue is effectively 5.6 times stiffer during indentation than it is in the absence of fluid pressure. Fluid pressure contributions to load support are necessary for carrying physiological loads while reducing friction and matrix stresses.

Subsequently, the indentation response was analyzed for a range of indentation rates from 580 nm/s to 21 μm/s. Prior studies have shown that fluid pressure and matrix stress are independent to an excellent approximation.[2] Consequently, the fluid load contribution at a given indentation depth is the difference of the total force and the equilibrium force. The fluid load fraction, *Wp/W*, is given by:

$$\frac{W_p}{W} = \frac{F_n - F_s}{F_n} = \frac{E' - E_0}{E'} \qquad \text{(eq. 2)}$$

Where, *Fs* is the force from matrix strain, and *Eo* is the equilibrium modulus. Since modulus is essentially constant with depth, the fluid load fraction is constant with depth for indentation at a given set of conditions.

Both the effective modulus and fluid load fraction increase monotonically with increasing indentation rate (*see Figure 4*). At 580 nm/s, load support exceeds 50%. At 21 μm/s, interstitial fluid supports up to 85% of the normal load. Fluid pressurization significantly increases joint load capacity while simultaneously reducing the normal and friction forces on the matrix. The unique biphasic structure provides improved functionality in response to increased tribological intensity (impact, sliding). Localized structural damage that impedes functionality (i.e., fluid load support) may be a mechanical mechanism that leads to propagation of damage in osteoarthritis.

CONCLUSIONS

- The prescribed contact conditions can

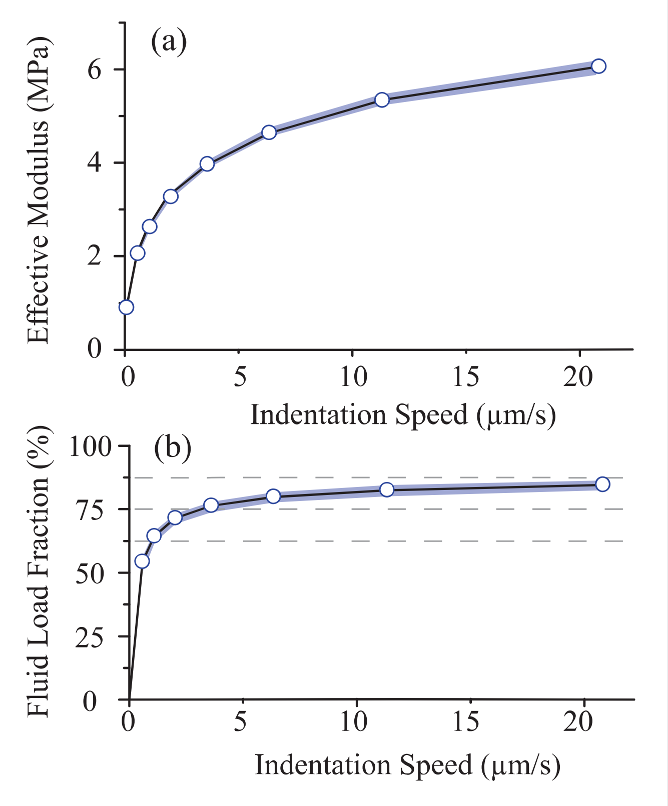

Figure 4 | **(a) Effective modulus and (b) fluid load fraction versus indentation rate. Shaded regions depict uncertainty.**

be described by Hertzian mechanics for both indentation and equilibrium responses.

- Fluid pressurization can support substantial portions of the load; which can decrease friction by the same proportion (according to Krishnan et al.).
- Sliding in-vivo promotes continuous deformation that leads to fluid pressurization and effective lubrication.

ACKNOWLEDGMENTS

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