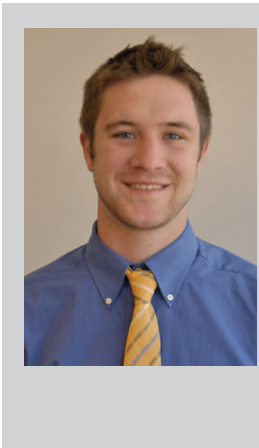


# Microtribology of Articular Cartilage: Effects of Sliding Speed and Contact Area

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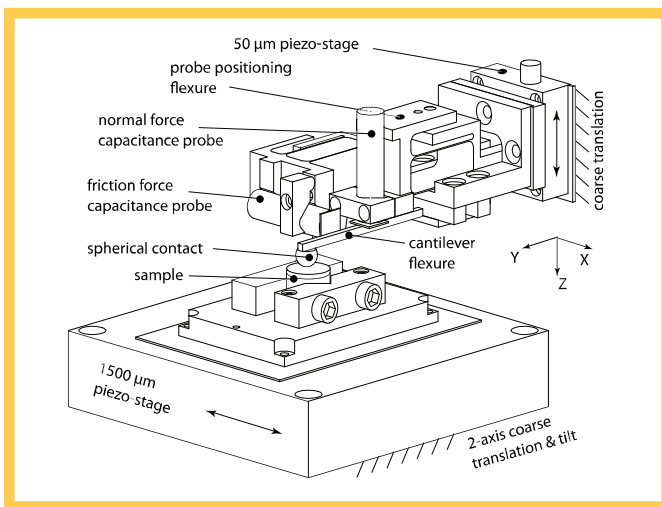


Figure 1 | Illustration of the custom microtribometer.

## INTRODUCTION

Osteoarthritis (OA), a degenerative joint disease associated with the degradation of articular cartilage, is a leading cause of chronic disability in the United States. Once OA has progressed to a symptomatic and visually detectable stage, the damage is irreversible and joint failure is imminent. In the earliest stages of OA, where treatment and rehabilitation are most effective, damage is highly localized, difficult to detect and difficult to study; the initiation and progression processes are poorly understood. Cartilage mechanics are highly sensitive to the poroelastic<sup>1</sup> structure, and typical tribological contacts on cartilage are insensitive to the effects of local damage on the local tribological response because of large contact areas. However, published data states that fluid pressurization in the cartilage matrix is not possible for microscale contacts.<sup>2</sup> This research targets an improved understanding of the tribological response of cartilage under microscale contacts to facilitate more fundamental studies of the OA progression processes.

## METHODS

All reported results were recorded from one cylindrical bovine cartilage plug ( $\phi 10\text{mm} \times 10\text{mm}$ , 1mm cartilage 9mm subchondral bone). One sample is used to eliminate the natural variation in biological samples. The results were determined to be representative of the population as they were consistent with more than 40 other samples tested over a four-month period. The tests were run submerged in phosphate buffered saline (PBS) solution.

Tests are carried out on a custom-built microtribometer shown in Figure 1. A horizontal piezo positioning stage (1500  $\mu\text{m}$  path) reciprocates samples, and a vertical stage (50  $\mu\text{m}$ ) lowers a transducer to apply loads. Forces are measured by use of a calibrated flexure fixed to the vertical stage. Deflections of the flexure are recorded by capacitance probes with sub-nanometer (nm) resolution. The probes used are  $\phi=1/4"$  and  $1/16"$  stainless steel spheres.

The measurement technique illustrated in Figure 2 enabled *in-situ* tracking of the contact area. Contact area is calculated from a measured displacement into the samples.

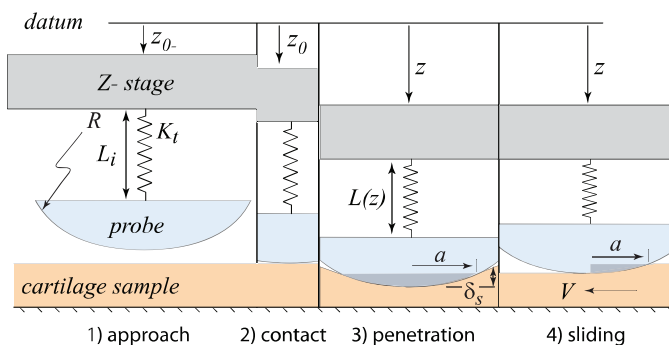


Figure 2 | Illustration of experimental procedure. The stage is extended its full range of  $z=50 \mu\text{m}$  at each condition. The cartilage is softer at slower speed, resulting in lower normal force and larger contact area.

The indentation depth is given by:

$$\delta_s = (Z - Z_0) - (L_i - L(Z)) \quad (1)$$

Where,  $Z$  is the displacement of the vertical piezo stage,  $Z_0$  is the displacement of initial contact, and  $L_i - L(Z)$  is the deflection of the flexure at  $Z$ . From  $\delta_s$ , a Hertzian contact model is used to determine the area in contact. Simple manipulation leads to a formula for  $a$  the contact half width as:

$$a = (R \cdot \delta_s)^{1/2} \quad (2)$$

Where,  $R$  is the radius of the contact probe.

An effective elastic modulus,  $E'$ , is defined here with a quasi-static application of the Hertz solution:

$$E' = (3/4) \cdot F_N \cdot R^{-0.5} \cdot \delta_s^{-1.5} \quad (3)$$

Where,  $F_N$  is the normal force. Calculation of the equilibrium  $E'$  with zero sliding speed is the elastic modulus of just the cartilage matrix  $H_a$ . The elastic force contribution with sliding,  $F_e$ , is calculated using equation 3 with  $H_a$  to solve for  $F_e$ . The force due to fluid pressurization,  $F_p$ , is then the difference of the total normal force and the elastic force. The fluid load fraction is simply the ratio of fluid force to the total normal force.

## RESULTS

Figure 3 shows that sliding has a significant effect on the effective stiffness of the cartilage surface. At zero sliding velocity, local fluid pressurization vacates the contact area and the force response is due solely to the deformation of the matrix.

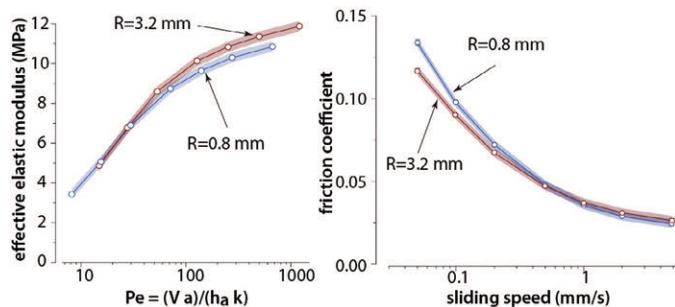


Figure 3 | Effective elastic modulus vs. Peclet number (shown on the left). Friction coefficient vs. sliding speed (shown on the right).

The effect of probe radius on friction coefficient is insignificant in comparison to the effects from variable sliding speed (Figure 3). The elastic force contributions are determined using static indentation measurements. The fluid supports a significant fraction of the normal force at all speeds independent of probe radius (Figure 4). Interestingly, the friction coefficient was proportional to the contact area (Figure 5); this suggests that the primary lubrication benefit of the fluid is to reduce the contact area. At slower sliding speeds, the

cartilage surface is effectively softer and a larger area of the matrix is brought into contact to support the load. The ratio of friction force to elastic normal force component is the interfacial friction coefficient; the interfacial friction coefficient is not particularly sensitive to probe size and is a function of sliding speed.

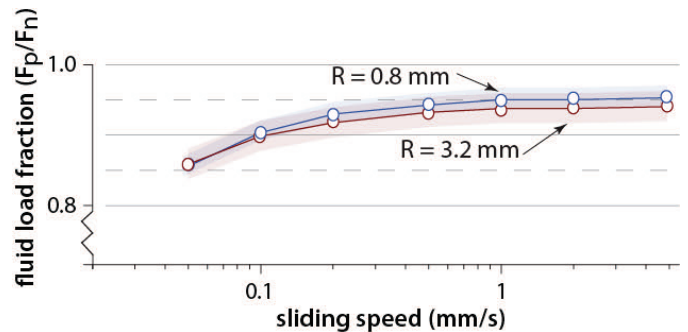


Figure 4 | Fluid load fraction vs. sliding speed for both probe radii.

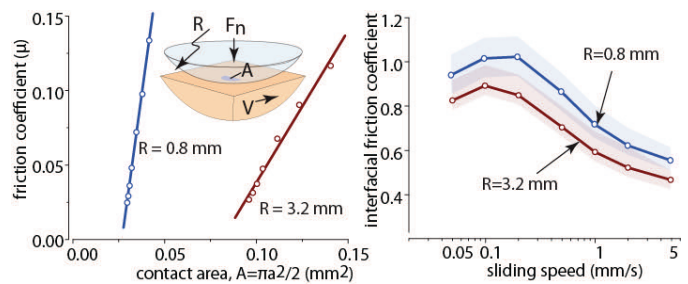


Figure 5 | Friction coefficient vs. contact area (shown on the left). Interfacial friction coefficient (normal force\*elastic load fraction/friction force) vs. sliding speed (shown on the right).

## CONCLUSIONS

1. At stationary equilibrium (e.g., standing), the soft matrix requires a large contact area to support the entire load.
2. Sliding contacts (as slow as 0.05 mm/s) induce sustainable fluid pressure, which increases stiffness and reduces contact area, normal stress and shear stress.
3. Sustained fluid pressurization can be achieved for a microscale contact, which enables future research on locally damaged cartilage with microscale contacts.

## ACKNOWLEDGMENTS

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