Fluid load support in the migrating contact area: How much migration is necessary?

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INTRODUCTION

It is well-established that cartilage achieves full fluid load support when Pe>>1 on an infinite track. In theory, cartilage maintains no fluid load support when the track length approaches zero, regardless of Pe. Understanding how inactivity and range of motion limitations might affect joint function and disease risk depends on the transition behavior between these two limiting states. Fortunately, all the work done on the MCA (migrating contact area) involves a finite track. Unfortunately, the transition mechanics have yet to be studied and the literature provides little insight into how load, contact area and range of motion restrictions might influence fluid load support of cartilage and general joint function. This study was designed specifically to quantify these effects and provide insight into their possible contributions to joint dysfunction and disease.

METHODS

Materials: Adult bovine stifles were obtained from a local abattoir. Full thickness ϕ19 mm osteochondral plugs (n=5) were extracted along the center line of the medial and femoral condyles from 3 joints. Samples were rinsed in phosphate buffered saline (PBS) and stored in protease inhibitor solution at 4°C. The samples were CNC-milled, mounted into a custom tribometer , and lubricated in a 1X PBS bath. Borosilicate glass spheres (ϕ 2.4, 3.9, 6.4mm) were used as the counter-body. The material properties of each sample were quantified once before tribological testing and once after. All tests were completed within 24hr of extraction to minimize the risk of enzymatic digestion-induced changes in material properties. Any sample whose equilibrium contact modulus fell outside our own definition of normal cartilage (0.1-1.5 MPa) or changed by more than 50% following tribological testing was discarded.

Tribological Testing: Equilibrium in the static indentation experiment served as the zero-track control for the sliding experiments and ensured zero fluid load support. Immediately following static equilibration, the lateral stage began reciprocating the cartilage sample relative to the probe at 1.5 mm/s (Pe>>1) over a 7 mm long track. In every case, the measured Pe number exceeded 100, which implies that the measured fluid load support was within 1% the maximal fluid load support of that system. Reciprocal sliding persisted at these conditions until the system reached a new dynamic equilibrium (<0.3 μm/min). The track length was then reduced to 4 mm and the system re-equilibrated. This procedure was repeated for 2, 1, 0.5, 0.2, 0.1, and 0.05 mm track lengths in that order. Following the final equilibration, sliding was terminated, the probe retracted, and the sample left to free-swell in the PBS bath for 10 min before being subjected to another indentation measurement to test for variations in material properties post-sliding. This variable track testing procedure was repeated for each sample at each of the three target loads (5, 20, and 100 μN) and probe sizes (ϕ 2.4, 3.9, 6.4mm) in random order.

In-situ characterization: Sample deformation was quantified directly and in real-time based on real-time measurements of force and stage position following determination of z_0 (surface location) and pre-calibration of stiffness by indenting a rigid substrate. The contact radius, diameter, and area are related to deformation through the geometry of the contact. Our previous in-situ observations showed that the contact geometry was best represented by simple Hertzian mechanics according to the following relationship, where a is the contact radius, δ_s is sample deformation and R is probe radius [1].

$$a = \sqrt{\delta_s * R} \tag{1}$$

The equilibrium (static) and effective (sliding) contact moduli are determined based on Hertzian contact mechanics using equation (2).

$$E_c = \frac{E}{1-\nu^2} = \frac{3}{4}\frac{F}{R^{0.5}{\delta_s}^{1.5}} \tag{2}$$

Finally, fluid load support is quantified using the effective modulus, E'_c, at a given sliding condition and the corresponding equilibrium modulus, E_{c0}, as described previously [2].

$$F' = \frac{F-F_s}{F} = \frac{\frac{4}{3}E'_c R^{0.5}{\delta_s}^{1.5} - \frac{4}{3}E'_{c0} R^{0.5}{\delta_s}^{1.5}}{\frac{4}{3}E'_c R^{0.5}{\delta_s}^{1.5}} \tag{3}$$

Analysis: The fluid load support (F') that develops in an] MCA of an infinite track is defined by equation (4) as described previously [4].

$$F' = \left[\frac{E^*}{E^*+1}\right] \cdot \left[\frac{Pe}{Pe+1}\right] \tag{4}$$

The first term is a material term that limits the maximum fluid load support; we will refer to this limiting value as $F'_{infinite}$. A linearly elastic biphasic material (e.g. hydrogel) is limited to a maximum of 50% fluid load support, whereas cartilage, which often has much higher tensile stiffness than compressive stiffness due to the nature of its collagenous structure, can approach 100% fluid load support. The second term is a speed term governed by the Pe number. When Pe << 1, fluid load support is negligible, when Pe >> 1, the fluid load support approaches $F'_{infinite}$, and when Pe = 1, the fluid load support is 50% $F'_{infinite}$. One can remove material effects by defining a dimensionless fluid load support ratio (F^*) as the ratio of F' to $F'_{infinite}$.

$$F^* = \frac{F\prime}{F\prime_{infinite}} \quad (5)$$

Finally, we define a dimensionless track length, S^*, as the ratio of track length, S, to contact diameter, d.

Post testing, the mean contact area (A) was determined based on the radius assuming a circular contact and the contact length (d)—which twice the contact radius (a). The contact area of the total dataset was sorted into three groups small (0.05-0.1mm^2), medium (0.3-0.5mm^2) and large (0.8-1.4mm^2) areas.

RESULTS

Fluid load support depended primarily on the migration length per unit contact length (S*) and maintained a maximal magnitude (F*=100%) at S* > 10 – in other words, any track longer than 10x the contact length is effectively an infinite migrating contact (Figure 1).

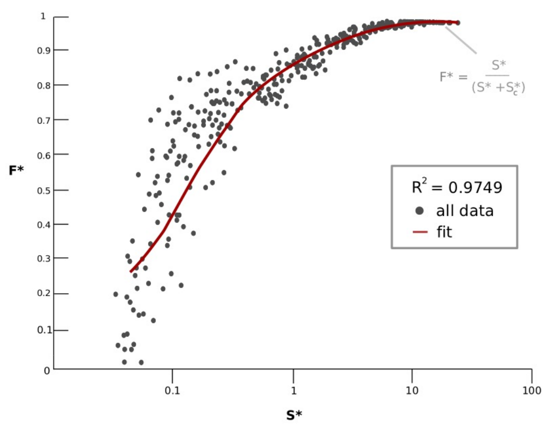

Figure 1: Plot of N=5 for entire dataset—varied load, track length and probe size.

At S* < 10, the fluid load fraction varied as a sigmoidal function of S*, falling to F* = 50% by S* = 0.1 on average. This value was defined as the transition migration length, S_c*. The data was fit to F* as a function of S* and S_c*.

$$F^* = \frac{S*}{S*+S_c*} \quad (6)$$

Using MATLAB, the results of each variable track-length experiment were fit individually to obtain the transition migration length, S_c*, for 1000 datasets of varying load, probe radius and contact area (Figure 2). Transition migration length (S_c*) did not depend significantly on probe radius, but had slight yet significant effects from load and contact area. As applied load increased, a larger S_c* was needed to achieve a fluid load support of at least 50%.

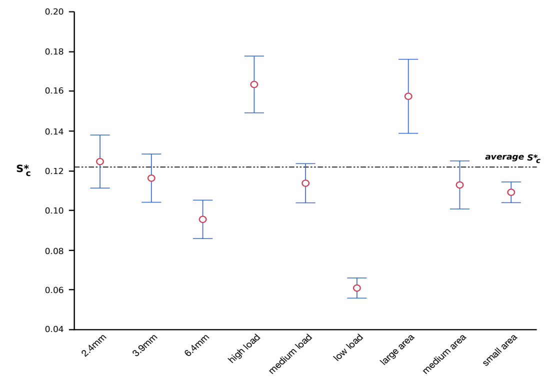

Figure 2: Transition migration length (S_c*) with 95% confidence interval for each experimental condition. The dashed line represents the average S_c* = 0.121.

DISCUSSION

When the migration length approached the contact length, the fluid load support of cartilage fell well below that predicted by the established mechanics of migrating contacts. Current accepted theory suggests that cartilage fluid load support is sustained under the condition that migration speed is fast enough to prevent the loss of interstitial fluid—Pe>>1. While fluid load support is a function of the Peclet number, it is also a function of track length. Based on our results, track lengths larger than 10 times the contact diameter (S*=10) have no significant effect on fluid load support and thus can safely be treated as infinite tracks. We propose a simple analytical correction (equation 6), based on the transition migration length (S_c*), should be used when S*<10. Although care must be used when extrapolating our results, the function describes all the data from this study (varying joints, samples, loads, contact areas) with R^2 > 0.97.

Clinically, the results suggest nominally static joints lose fluid load support like an SCA (stationary contact area), despite micro-motions with Pe>>1 (i.e. fidgeting or swaying back and forth). Thus, intentional full range of motion activity is necessary to prevent dehydration of joints.

Our results also showed a significance of applied load on fluid load support and its effects on transition migration length. Increasing applied loads corresponds physiologically with increasing bodyweight. With increased bodyweight, there is a higher dependence on track length. Therefore, in order to maintain fluid load support, one with a higher bodyweight would need to engage in more full range of motion activities compared to one of a smaller bodyweight. The heaviest and least active populations are likely at increased risk of dysfunction.

These results demonstrate that fluid retention and load support are impaired by reduced activity and reduced ranges of motion, especially given the relatively short tracks of most joints at full range of motion (S*<5, typically).

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