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The Temporal Response of the Friction Coefficient of Articular Cartilage Depends on the Contact Area

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Abstract

The hypothesis of this study is that the time constant for the transient increase in friction coefficient of articular cartilage under a constant load is proportional to the size of the contact area, as predicated by the dependence of the frictional response on interstitial fluid pressurization. This hypothesis is verified experimentally from measurements of the frictional response of bovine articular cartilage disks of three different diameters (4, 6 and 8 mm) against glass. At two different applied stresses (0.127 MPa and 0.254 MPa), the coefficient of determination of a linear regression of the time constant versus the contact area yielded $R^2 = 0.997$ ($p < 0.002$). The results of this study provide a cogent explanation for the expectation that the friction coefficient in situ will not achieve the elevated equilibrium values observed under common testing conditions.

Introduction

The frictional response of articular cartilage has been shown to be time-dependent, with the friction coefficient increasing monotonically under a constant load until reaching an equilibrium value (McCutchen, 1962; Wright and Dowson, 1976; Forster and Fisher, 1996). This temporal response has been associated with the concomitant creep deformation of the tissue (McCutchen, 1962), prompting the formulation of hypotheses relating the frictional response of cartilage to its interstitial fluid pressurization (McCutchen, 1962; Maciowski et al., 1994; Forster and Fisher, 1996; Ateshian, 1997; Ateshian et al., 1998). This hypothesis has been verified experimentally in our recent study (Krishnan et al., 2004) where the frictional response of bovine cartilage against glass was found to correlate negatively with the measured interstitial fluid pressure (high pressure = low friction, low pressure = high friction).

A natural question following this finding is whether the friction coefficient rises to elevated values under physiological conditions in situ, and if not, what prevents it from doing so. In a recent study we showed that dynamic loading of cartilage for up to one hour does not sustain elevated fluid pressurization, leading to the same increase in friction as under static loading (Krishnan et al., 2005); therefore, dynamic loading could not explain why the friction coefficient might stay low in situ. An alternative explanation is that loss of fluid pressurization can be delayed with larger contact areas, which impose a longer path for pressurized fluid to escape from the loaded region, as predicated by porous media models such as the biphasic theory (Mow et al., 1980; Armstrong et al., 1984). Therefore, the hypothesis of this study is that the time constant for the increase in friction coefficient under a constant load is proportional to the size of the contact area, a concept introduced in our earlier studies (Park et al., 2004; Krishnan et al., 2005). This hypothesis is tested experimentally by measuring the frictional response of bovine articular cartilage disks of different radii against glass.

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Materials and Methods

Cartilage samples were harvested from three 1-3 months old bovine shoulder joints obtained from a local abattoir on the day that the animals were sacrificed. Specimen preparation followed the protocol described in our earlier study (Krishnan et al., 2005). Cylindrical plugs were prepared with diameters of 4 mm (n=12, 1.5 ± 0.2 mm thick) and 6 mm (n=12, 1.6 ± 0.1 mm), and 8 mm (n=12, 1.5 ± 0.1 mm).

The friction testing apparatus was the same as described in our earlier study (Krishnan et al., 2005). The testing protocol consisted of measuring the time-dependent friction coefficient of the cartilage plugs against glass, in phosphate buffered saline containing protease inhibitors, in unconfined creep under a constant applied stress (0.127 MPa and 0.254 MPa, n=6 samples of each diameter for each stress), with a sliding velocity of 1 mm/s and reciprocal translation of ± 4.5 mm. Based on preliminary studies, tests were terminated after 18,000s for $\varnothing 4$ mm samples, 35,000s for $\varnothing 6$ mm samples, and 45,000s for $\varnothing 8$ mm samples. For all specimen sizes, 100 cycles of sliding were prescribed. The time lapse between consecutive sliding cycles was increased logarithmically to minimize wear at later time points. The time-dependent friction coefficient, μ_{eff} , was calculated from the ratio of the friction force to the normal force. The minimum value of μ_{eff} was denoted by μ_{min} and the equilibrium value by μ_{eq} . The characteristic time constant τ_μ for μ_{eff} to rise from μ_{min} to μ_{eq} was determined as the time required for μ_{eff} to rise from μ_{min} by $(1 - e^{-1})(\mu_{eq} - \mu_{min})$.

A linear regression analysis was performed on the raw data for τ_μ versus contact area. Based on the hypothesis of this study, which predicates a linear proportionality between these variables, the regression analysis did not include a constant. The correlation coefficient R was used to test the measure of the linear relationship. Two-factorial analysis of variance ($\alpha=0.05$) and Bonferroni post-hoc testing were used to detect statistical differences in τ_μ , μ_{min} , μ_{eq} and the equilibrium creep strain, ϵ_{eq} , for the three specimen diameters and two contact stresses (Mathematica 5.2, Wolfram Research Inc., IL).

Results

Average results for the normalized time-dependent friction coefficient, μ_{eff}/μ_{eq} , are presented in Figure 1 for 0.254 MPa applied stress. As expected, the friction coefficient rose monotonically from μ_{min} to μ_{eq} (also reported in Table 1), with the larger diameter samples requiring a longer time to equilibrate. Average and standard deviations of μ_{min} , μ_{eq} , τ_μ , and ϵ_{eq} , are presented in Table 1. Plots of the average values of τ_μ versus the specimen cross-sectional area, which is also the contact area between the cartilage and glass contact area, exhibit a strong and highly significant linear dependence (Figure 2). τ_μ ($p=0.004$), ϵ_{eq} ($p=0.043$), and μ_{eq} ($p<0.001$) increased significantly with applied stress, and τ_μ increased significantly with specimen diameter ($p<0.001$) (Table 1). μ_{min} was higher for the $\varnothing 4$ mm than the $\varnothing 6$ mm and $\varnothing 8$ mm specimens ($p=0.001$).

Discussion

The finding that the time constant τ_μ for the friction coefficient to reach equilibrium increases linearly with the contact area validates the hypothesis of this study (Figure 2). In our previous study (Krishnan et al., 2004), it was shown experimentally that the friction coefficient varies linearly with interstitial fluid pressurization, confirming earlier friction hypotheses in the literature (McCutchen, 1962; Forster and Fisher, 1996; Ateshian et al., 1998), and indicating that the time-constant τ_μ for the frictional response is the same as that of the flow-dependent viscoelasticity of cartilage.

From the biphasic theory (Mow et al., 1980), it is known that the gel time constant for the flow-dependent viscoelasticity of cartilage in unconfined compression of a cylindrical disk, or contact of spherical articular layers, is proportional to the square of the disk or contact area radius (Armstrong et al., 1984; Ateshian et al., 1994). Indeed, for a linear isotropic solid matrix, $\tau_\mu \propto a^2/H_A k$ according to theory, where a is the contact area radius, H_A is the aggregate modulus of the solid matrix, and k is the hydraulic permeability; when accounting for the tension-compression nonlinearity of cartilage, the aggregate modulus corresponds to the tensile modulus (Soltz and Ateshian, 2000). Therefore the outcome of this study validates this theoretical dependence, further confirming our understanding of the dependence of the cartilage frictional response on interstitial fluid pressurization.

In addition it was also observed that τ_μ increases with increasing contact stress (Figure 2). This outcome can be explained by the nonlinear response of the tissue; the hydraulic permeability k decreases exponentially with applied stress (Mansour and Mow, 1976) due to loss of tissue volume (Lai et al., 1981) as represented indirectly by ε_{eq} , whereas the tensile aggregate modulus H_A (which is closely related to the tensile Young's modulus) remains constant or increases moderately with applied stress (Woo et al., 1976; Roth and Mow, 1980; Akizuki et al., 1986). Thus, $H_A k$ is expected to decrease with increasing stress, producing higher values of τ_μ . Note that the change in contact area with increasing stress was minimal in this study; assuming that Poisson's ratio is ~ 0.1 (Wang et al., 2002), it is estimated from ε_{eq} in Table 1 that it increased by no more than 0.8%.

μ_{min} , which was observed to be higher in the smallest specimens, is regulated by the peak interstitial fluid load support (Krishnan et al., 2004). Since μ_{min} is typically achieved in the first few cycle of sliding, a faster decrease in fluid pressurization in the smaller samples in the early time response may explain the higher μ_{min} .

The time constants observed in this study are on the low end of expected values in situ, since the tested contact areas and applied stresses were relatively small. Indeed, for human joints, contact areas of 50 mm² (the high end of values tested here, Figure 2) are representative of finger and wrist joints, which typically experience stresses of 0-3 MPa (Tencer et al., 1988; Ateshian et al., 1995). For the patellofemoral joint, contact areas may be as high as 400 mm², with peak contact stresses of 4.5 MPa under normal activities (Ahmed et al., 1983; Huberti and Hayes, 1984). Similar values are observed in the tibiofemoral joint (Kurosawa et al., 1980; Ahmed and Burke, 1983). Therefore, it is expected that the time constants under physiological conditions are on the order of several hours, based on a reasonable extrapolation of the results of this study. The reason that larger specimens were not tested in this study is that the natural curvature of joints prevents the harvesting of sufficiently flat disks of larger diameters. The application of higher loads was precluded by the range of the multi-axial load cell used for measuring the friction and normal forces, which is partly constrained by the need to capture friction coefficients that vary over two orders of magnitude (Table 1).

The lubrication regime observed here is that of 'biphasic lubrication', a term coined by Forster and Fisher (Forster and Fisher, 1996), characterized by the role of interstitial fluid pressurization in regulating the frictional response. Alternatives such as fluid-film lubrication and boundary lubrication cannot predict the time-dependent behavior observed here, nor the linear relationship between τ_μ and the contact area.

The direct implication of this experimental study is that the friction coefficient of cartilage will remain low for longer durations when the contact area is larger. Thus, congruent joints, which achieve the highest possible articular contact area for given joint dimensions, promote lower friction for longer durations. The results of this study provide a cogent explanation for our expectation that the friction coefficient in situ will not achieve the elevated equilibrium values

observed under common testing conditions. They also suggest that loss of joint congruence, either due to trauma or surgical interventions such as meniscectomy, may lead to a more rapid rise in friction.

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Figure 1.

Average response of μ_{eff} for each of three specimen diameters (n=6 per group), for an applied stress of 0.254 MPa.

Figure 2.

Linear fit of τ_{μ} versus contact area, for both applied stresses. Fits were performed on the raw data, but plots show mean $\pm$ standard deviations for clarity. R is the correlation coefficient; the p -value represents the probability that the coefficient of *area* in the regression equation is not significantly different from zero.

Table 1

Average $\pm$ standard deviation of the minimum and equilibrium friction coefficients, $\mu_{\min}$ and μ_{eq} , the time constant for the frictional response, τ_{μ} equilibrium creep strain, ε_{eq}

	$\mu_{\min}$	μ_{eq}	τ_{μ} (S)	ε_{eq}
		Applied stress = 0.127 MPa		
4 mm	0.0156 ± 0.0097	0.30 ± 0.14	818 ± 467	0.36 ± 0.06
6 mm	0.0059 ± 0.0032	0.22 ± 0.06	1721 ± 433	0.36 ± 0.05
8 mm	0.0033 ± 0.0012	0.26 ± 0.12	2800 ± 1754	0.40 ± 0.03
		Applied stress = 0.254 MPa		
4 mm	0.0087 ± 0.0068	0.37 ± 0.10	1339 ± 181	0.40 ± 0.08
6 mm	0.0053 ± 0.0014	0.36 ± 0.05	2499 ± 747	0.40 ± 0.05
8 mm	0.0041 ± 0.0015	0.38 ± 0.03	4271 ± 822	0.42 ± 0.02