FRICTIONAL RESPONSE OF BOVINE ARTICULAR CARTILAGE BEFORE AND AFTER REMOVAL OF THE SUPERFICIAL TANGENTIAL ZONE

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INTRODUCTION

Articular cartilage serves as the bearing material in diarthrodial joints. It provides for effective joint lubrication over several decades of loading. In previous studies it has been hypothesized that interstitial fluid load support contributes significantly to the reduction of the frictional coefficient due to load transfer from the solid to the fluid phase of the tissue [1-4]. Other authors have shown that the low frictional coefficient of articular cartilage can be attributed to surface proteins found in the superficial layer [5-6], phospholipids [7], or other boundary lubricants such as LGP-I from the synovial fluid [8]. Furthermore, a distinct upper superficial surface layer, 70-300 nm thick has been identified and shown to play a role in joint lubrication [9-11]. The specific aim of this study was to investigate the role of the surface layer in dictating the frictional response of bovine articular cartilage. Friction tests with sliding of bovine articular cartilage against glass in the configuration of unconfined compression creep were carried out before and after removal of the surface layer in order to investigate the differences in the frictional response of the tissue.

MATERIALS AND METHODS

Specimen Preparation

Twenty one cartilage plugs (diam. 8mm) were harvested from the humeral head of bovine shoulder joints obtained from a local abattoir (4 joints, ages 4-6 months). Using a sledge microtome (model 1400; Leitz, Rockleigh, NJ), tissue was removed from the deep zone to produce samples of thickness 1.1 ± 0.112 mm within the control group (n=11) and 1.1 ± 0.125 mm within the treated group (n=10). All specimens were stored at -20°C in physiological buffered saline (PBS) solution. On the day of testing, each sample was thawed and equilibrated in PBS solution. In order to obtain a uniform cylindrical cross section, 4.85mm plugs were further cored out from the samples. Two frictional tests were performed on each sample. For the first test, the articular surface was left intact in both groups. For the second test, control group specimens were left intact, whereas the surface layer was microtomed for all specimens within the treated group to obtain a final thickness of 0.973 ± 0.167mm. In order to control for any changes potentially induced by specimen preparation and handling, samples from the control group were similarly frozen within the embedding matrix on the microtome stage and later thawed and equilibrated in PBS before being tested.

Frictional Testing Apparatus

Sliding motion of cartilage against glass was provided by a computer controlled translation stage (Model PM500-1L, Newport Corporation, CA). Normal loads were prescribed via a voice coil actuator (Model LA 17-28-000A; BEI Kimco Magnetics Division, CA), connected in series with a linear variable differential transformer to measure creep displacements (HR100, Shaevitz sensors, VA). All loads were measured with a multiaxial load cell mounted on the translation stage (Model 20E12A-M25B,JR3 Inc., CA). Data acquisition and control was performed using a personal computer equipped with a data acquisition card. The entire specimen and glass surface were immersed in 0.15M PBS solution mixed with protease inhibitors (Complete protease inhibitor cocktail tablets, Roche Applied Science, IN). Tests were performed at room temperature.

Loading Protocol

All frictional tests were conducted between cartilage and glass, under the configuration of unconfined compression creep with a prescribed load of W =1.35N, with intermittent sliding over logarithmic time increments (range of translation ±2mm; sliding velocity 1mm/s). The test was terminated upon reaching creep displacement equilibrium, or after 10,000 seconds, whichever occurred first. The normal force, frictional force and creep displacement were monitored throughout the test. At the end of the first test sufficient time was allowed for the sample to equilibrate before being frozen for storage at -20°C until the day of the second test. The protocol for the second test was identical.

RESULTS

In all tests, the frictional coefficient, μ_{eff}, was observed to be time-dependent, achieving its minimum value μ_{min} immediately upon loading and eventually reaching an equilibrium value μ_{eq} (Figs. 1,2). In the control group (Fig. 1), μ_{min}=0.007 ± 0.006 and μ_{eq}=0.49 ± 0.097 over a loading duration of 9,026 ± 1,617s in the first test, and μ_{min}=0.018 ± 0.007 and μ_{eq}=0.332 ± 0.068 over 9,419 ± 3,007 s in the second test (mean ± std. deviation). In the treated group (Fig. 2), μ_{min}=0.008 ± 0.007 and μ_{eq}=0.46 ± 0.084 over 9,734 ± 1,157 s in the first test, prior to surface removal, and μ_{min}=0.047 ± 0.02 and μ_{eq}=0.131 ± 0.034 over 9,406 ± 2,492 s in the second test, after surface removal. A two-way ANOVA (α=0.05) with repeated measures between the first and second tests, and Bonferroni posthoc test of the means, demonstrated the following results: No statistical differences were found in μ_{min} for the control group between the first and second test (p=0.16). Significant differences were found in μ_{min} in the treated group, between the first and second test (p<0.0001). No differences were found in μ_{min} between the control and treated groups in the first test (p=1.00), but differences were found in the second test (p<0.0001). Statistical differences were observed in μ_{eq} for the control group between the first and second test (p=0.0001). Significant differences were found in μ_{eq} in the treated group, between the first and second test (p<0.0001). No differences were found in μ_{eq} between the control and treated groups in the first test (p=1.0), but differences were found in the second test (p<0.0001). No differences were found in the equilibrium modulus between control and treated groups or between the first and second test (p>0.15, E_Y = 0.208 ± 0.052 MPa).

DISCUSSION

The time dependent frictional response of intact articular cartilage reported in the current study agrees with previous findings in the literature [1-4,10]. It was found that repeated testing of control group specimens leads to a significant decrease in μ_{eq}, suggesting a wear process induced by the continuous loading configuration. Removal of the surface zone led to a dramatic decrease in μ_{eq}, which suggests that the articular surface does not possess a uniquely efficacious boundary lubricant that can promote lower values of μ_{eq}. It is also possible that the microtoming process reduced the natural surface roughness of the tissue, contributing to the decrease in μ_{eq}. Conversely, removal of the surface zone led to an increase in μ_{min}, from 0.018 to 0.047. Coupled with the finding on μ_{eq}, this result suggests that interstitial fluid pressurization, which has been implicated in the low friction characteristic of cartilage [1-4,10], is compromised by the loss of the surface zone. This interpretation is consistent with our understanding that fluid pressurization is enhanced by an increasing ratio of tensile to compressive stiffness of cartilage [12], and the knowledge that this ratio is greatest at the articular surface. These results increase our insight into the role of the articular surface zone in cartilage friction.

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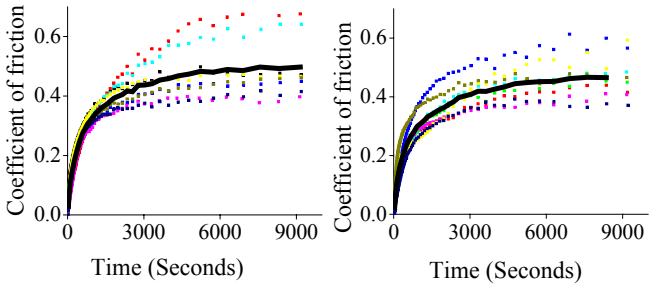

Figure1: Variation of μ_{eff} for (a)control and (b)test group; test1

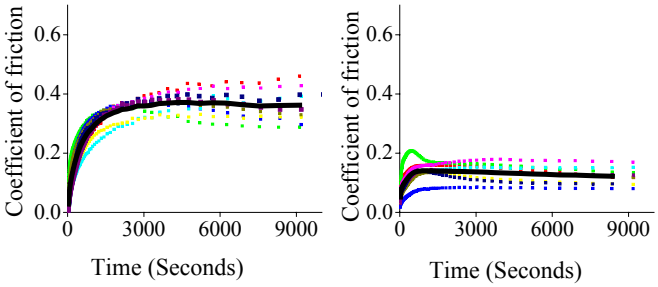

Figure2: Variation of μ_{eff} for (a)control and (b)test group; test2