Published in final edited form as:

J Biomech. 2015 November 5; 48(14): 3945–3949. doi:10.1016/j.jbiomech.2015.09.029.

The Friction Coefficient of Shoulder Joints Remains Remarkably Low Over 24 h of Loading

Brian K. Jones, Krista M. Durney, Clark T. Hung, and Gerard A. Ateshian*

Columbia University, Departments of Mechanical Engineering, Biomedical Engineering, and Orthopaedic Surgery, New York, NY, USA

Abstract

The frictional response of whole human joints over durations spanning activities of daily living has not been reported previously. This study measured the friction of human glenohumeral joints during 24 h of reciprocal loading in a pendulum testing device, at moderate (0.2 mm/s, 4320 cycles) and low (0.02 mm/s, 432 cycles) sliding speeds, under a 200 N load. The effect of joint congruence was also investigated by testing human humeral heads against significantly larger mature bovine glenoids. Six human joints and six bovine joints were tested in four combinations: human joints tested at moderate (hHCMS, $n=6$) and low speed (hHCLS, $n=3$), human humeral heads tested against bovine glenoids at moderate speed (LCMS, $n=3$), and bovine joints tested at moderate speed (bHCMS, $n=3$). In the first half hour the mean $\pm$ standard deviation of the friction coefficient was hHCMS: 0.0016 ± 0.0011 , hHCLS: 0.0012 ± 0.0002 , LCMS: 0.0008 ± 0.0002 and bHCMS: 0.0024 ± 0.0008 ; in the last four hours it was hHCMS: 0.0057 ± 0.0025 , hHCLS: 0.0047 ± 0.0017 , LCMS: 0.0012 ± 0.0003 and bHCMS: 0.0056 ± 0.0016 . The initial value was lower than the final value ($p < 0.0001$). The value in LCMS was significantly lower than in hHCMS and bHCMS ($p < 0.01$). No visual damage was observed in any of the specimens. These are the first results to demonstrate that the friction coefficient of natural human shoulders remains remarkably low (averaging as little as 0.0015 and no greater than 0.006) for up to 24 h of continuous loading. The sustained low friction coefficients observed in incongruent joints (~ 0.001) likely represent rolling rather than sliding friction.

Keywords

Articular cartilage; friction and lubrication; interstitial fluid pressurization; diarthrodial joint biomechanics; human shoulder joint

Introduction

Many studies have reported measurements of the friction coefficient in articular cartilage, dating back to the 1930s, as reviewed previously (Ateshian and Mow, 2005). The great majority of these studies tested cartilage against an artificial surface, typically glass or metal,

*Corresponding Author: Columbia University, Department of Mechanical Engineering, 500 West 120th Street, MC 4703, New York, NY 10027, USA. Tel.: +1 212 854 8602; fax: +1 212 854 3304. ateshian@columbia.edu (Gerard A. Ateshian).

Conflict of Interest

The authors have no conflicts to report for this study.

and showed that the friction coefficient rises over time under a constant applied load, due to loss of interstitial fluid pressurization, as reviewed recently (Ateshian, 2009). For cartilage explants a few millimeters in diameter, this effect occurs with a characteristic time constant on the order of 1 h (Forster and Fisher, 1996). Friction coefficients may increase 100-fold over this time period, from initial values of ~ 0.003 to equilibrium values of ~ 0.3 (Carter et al., 2007; Forster and Fisher, 1996).

At such elevated levels of friction, significant tissue wear has been observed after only 1 h (Oungouliau et al., 2013; Oungouliau et al., 2015). In contrast, activities of daily living may routinely last several hours per day, yet there is no evidence such prolonged activities lead to cartilage pathology. In fact, it is well established that joint activity and motion are required for healthy cartilage homeostasis (Kim et al., 1994; Kiviranta et al., 1988).

The friction response of whole diarthrodial joints or excised cartilage surfaces may differ markedly from the explant experiments described above if the contact region migrates over one or both cartilage surfaces. In the migrating contact configuration, low friction coefficients (~ 0.02 – 0.05) have been measured for up to 8 h between cartilage explants (Caligaris and Ateshian, 2008; McCann et al., 2009b; Northwood and Fisher, 2007). This sustained low-friction has been attributed to fluid load support, which remains elevated when the migrating velocity is significantly greater than the velocity of interstitial fluid flow (Ateshian, 2009).

Studies of the frictional response of human joints have been typically conducted for no more than 1 h of loading in saline or synovial fluid (Caligaris et al., 2009; Charnley, 1960; Clarke et al., 1975; McCann et al., 2009a; McCann et al., 2008; O'Kelly et al., 1978; Roberts et al., 1982; Unsworth et al., 1975), exhibiting average friction coefficients in the range of ~ 0.01 – 0.08 .

Since the frictional response of whole joints over durations spanning activities of daily living has not been reported previously, the objective of this study was to measure the frictional response of human glenohumeral joints for 24 h of continuous loading, at low and moderate sliding speeds. The effect of joint congruence was also investigated by testing human humeral heads against mature bovine glenoids, which have significantly larger radii of curvature than human counterparts. Whole bovine shoulder joints were also tested as controls.

Materials and Methods

Eight fresh frozen human shoulder joints were obtained from a tissue bank (6 male, 2 female, ages 29–59, median age 49), and six mature bovine joints were obtained from a local abattoir. Glenoids and humeral heads were excised and potted in quick-set epoxy (Hardman Epoweld 8173) within polyethylene or polystyrene dishes (Fisher) (Figure 1). All joints were visually normal, with no evidence of cartilage degeneration.

Samples were mounted in a custom pendulum testing apparatus, with the humeral head mounted on the pendulum and the glenoid serving as the fulcrum (Figure 2). All samples were tested for 24 h while immersed in a solution of physiological buffered saline, protease

inhibitor (ethylenediaminetetraacetic acid) and biocide (Proclin 950, Sigma) (Figure 3). The glenoid was mounted on a 6-axis load cell (JR3, Woodland, CA) secured in a rigid aluminum frame. The pendulum was raised and lowered using a winch driven by a stepper motor (Oriental, Torrance, CA). A custom LabVIEW (National Instruments, Austin, TX) program controlled the motor motion and acquired load cell data at a frequency of 20 Hz. The weight of the pendulum was 200 N, which represents 1/3 of body weight on average and is representative of glenohumeral joint contact forces for raising the arm against gravity (Poppen and Walker, 1978). At this load magnitude, load cell electrical noise limited the minimum measurable friction coefficient to ~ 0.0005 . The stepper motor position served as a proxy for the pendulum angular position θ .

The pendulum motion was adjusted to produce a sliding speed of either 0.2 mm/s (moderate) or 0.02 mm/s (low) at the articular surfaces of the joint, corresponding to cycle periods of 20 s or 200 s, respectively. The total testing duration was 24 h, resulting in approximately 4,320 cycles and 432 cycles of loading at the moderate and slow sliding speeds, respectively. As a point of reference, pedometer data indicate that activities of daily living involve $\sim 4,500$ walking steps in a control population (Mansi et al., 2014).

The friction coefficient μ was calculated by recognizing that the work of lifting and lowering the pendulum was conservative, whereas the work done by friction at the articular surfaces was dissipative. Denoting the force vector measured with the load cell at each pendulum position θ by $\mathbf{f}_F$ on the forward stroke and $\mathbf{f}_B$ on the backward stroke, it was thus possible to evaluate the friction (tangential) force as $\mathbf{f}_T = (\mathbf{f}_F - \mathbf{f}_B)/2$, and the contact (normal) force as $\mathbf{f}_N = (\mathbf{f}_F + \mathbf{f}_B)/2$. The friction coefficient was then evaluated from $\mu = |\mathbf{f}_T|/|\mathbf{f}_N|$ at each θ , and averaged over all θ for each reciprocal cycle. A more detailed description of this calculation is presented in the Supplementary Material section.

Six human shoulder joints were tested at the moderate sliding speed (hHCMS test group), three human joints at the low sliding speed (hHCLS), and three human humeral heads were tested against bovine glenoids (LCMS), with each test lasting 24 h. One human humeral head was reused in all three groups, and two human humeral heads were reused in two groups. Prior to retesting, reused specimens were allowed to recover in an unloaded configuration for 24 h in a fresh batch of bath fluid at 4 °C. Three bovine shoulder joints were also tested at the moderate sliding speed (bHCMS).

Human shoulder osteochondral samples were fixed in 10% formalin for 24 hours and subsequently decalcified in RDO Gold (Apex Engineering Products). Samples were embedded in paraffin blocks and 5 μm thickness sections were obtained. Sections were mounted on glass slides and stained in 0.1% Safranin-O and 0.1% Fast Green. Slides were imaged using an Olympus BX60 microscope and Olympus DP72 camera.

Results

The time-history of the friction coefficient μ for the four test groups is reported in Figure 4, by binning results over consecutive time point (0, 0.5, 1, 4, 8, 12, 16, 20, and 24 h) for the purpose of evaluating means and standard deviations of μ over all samples. The mean and

standard deviations of μ for the initial (0 – 0.5 h) and final (20 – 24 h) bins are presented in Figure 5. For all test groups, the friction coefficient in the initial response was $\mu=0.0015\pm0.0009$ (mean $\pm$ standard deviation); in the final response it was $\mu=0.0046\pm0.0025$. There was no visual evidence of damage to any of the joints after completion of all tests (Figure 6).

A two-way analysis of variance was performed to examine the effect of time (initial versus final bin) and testing configuration (hHCMS, hHCLS, LCMS, bHCMS), with $\alpha=0.05$ and post-hoc testing of the means with Bonferroni correction. For all three testing configurations, the final μ was significantly higher than the initial μ ($p<0.0001$). The friction coefficient of low-congruence hybrid joints (LCMS) was significantly lower ($p<0.01$) than that of natural human joints (hHCMS) and bovine joints (bHCMS) at the same speed.

Discussion

The objective of this study was to examine the friction coefficient of shoulder joints under continuous loading to 24 h, for the purpose of determining how the frictional response may evolve under sustained activities of daily living. While the friction coefficient of cartilage has long been recognized as being very low, there have been two potentially confounding factors that have constrained our understanding of the frictional response under physiologically relevant conditions: (1) Depending on testing conditions, the friction coefficient may rise considerably over time to values associated with rapid cartilage wear in less than a day of testing. Since no prior in vitro studies have been conducted over a full day of testing, this in vitro observation has not been fully reconciled with our understanding of in vivo conditions, where negligible damage is expected to occur within a single day's activities. (2) As reviewed in the Introduction, friction studies of cartilage rubbing against glass have often produced coefficients averaging as low as ~ 0.003 in the short-term response, whereas studies of cartilage-against-cartilage have shown minimum friction coefficients averaging ~ 0.01 . This apparent three-fold discrepancy does not reconcile with our expectation that the frictional response of cartilage-on-cartilage should be lower than that of cartilage against other materials.

The results of this study resolved the first of these issues by demonstrating that the average friction coefficient of natural human shoulder joints remained extremely low (below ~ 0.006) for up to 24 h of loading (Figure 5, hHCMS and hHCLS groups). This outcome was not affected statistically by the sliding speed, with both tested speeds falling on the low end of physiological conditions. This rather astounding result was even more remarkable for showing that the temporal increase had in fact achieved a steady value ~ 14 h into the test, suggesting that longer testing would not likely cause further increases.

These results may be interpreted in light of earlier theoretical explanations of sustained interstitial fluid pressurization under large (Carter et al., 2007) and migrating contact areas (Ateshian and Wang, 1995; Caligaris and Ateshian, 2008). In particular, the high congruence of the native human shoulder joint produced a large contact area that created a very long path for pressurized interstitial fluid to escape the loaded regions. For a path length $a \approx 10$ mm (radius of glenoid articular contour), a tensile cartilage modulus $H_A \approx 10$

MPa and a cartilage hydraulic permeability $k \approx 10^{-4} \text{ mm}^4/\text{N}\cdot\text{s}$, the time constant $\tau = a^2/H_A k$ for loss of fluid pressurization would be $\tau \approx 10^5 \text{ s}$ (28 h) in the absence of migrating contact (zero sliding speed) (Ateshian et al., 1994). With a sliding speed of $V = 0.2 \text{ mm/s}$, the Peclet number ($Pe = Va/H_A k$) would be $Pe \approx 2000$, which is significantly greater than unity, thus sustaining interstitial fluid pressurization (Ateshian and Wang, 1995). This non-dimensional number would remain significantly above unity ($Pe \approx 200$) even with $V = 0.02 \text{ mm/s}$, which explains the lack of statistical difference in friction between hHCLS and hHCMS. However, these conditions may not have completely precluded loss of interstitial fluid pressurization; indeed, it is most likely that the observed increase in μ over 24 h (Figure 4) was produced by a slight loss of interstitial fluid load support over that time span.

Moreover, the minimum friction coefficient achieved over the first half hour of testing averaged 0.0015 ± 0.0009 in the natural human joints, half the lowest reported average values for bovine cartilage plugs against glass. To the best of our knowledge, this is the lowest value reported to date for whole human joints, which may be attributed to two interacting mechanisms: (1) For highly congruent contact of healthy native articular layers, the interstitial fluid load support is nearly equal to unity (100% fluid load support). (2) When cartilage contacts cartilage, the fraction ϕ of the contact area where solid matrix contacts solid matrix (the solid-to-solid contact area fraction) is very small; if the porosity of the superficial zone cartilage is $\phi^w \approx 0.9$ (90% water content), then $\phi = (1 - \phi^w)^2 \approx 0.01$ or 1%. Furthermore, if fluid becomes trapped and pressurized between the articular layers, as may occur in the glenoid fossa in some positions of the human glenohumeral joint (see Fig. 6F of (Soslowsky et al., 1992b)), the value of ϕ would decrease further (Soltz et al., 2003), possibly by a factor of 2. Accordingly, the minimum friction coefficient could be as small as 0.5% of the equilibrium value μ_{eq} achieved when interstitial fluid pressure has subsided (Ateshian et al., 1998; Krishnan et al., 2004). Given that $\mu_{eq} \approx 0.13$ in human cartilage-on-cartilage studies (Caligaris et al., 2009), these estimates would be consistent with the lowest values of μ reported here.

It is noteworthy that the low friction coefficients observed in this study occur even in saline which, unlike synovial fluid (SF), contains no boundary lubricants. As demonstrated in our previous study (Caligaris and Ateshian, 2008), SF can reduce the friction coefficient by a factor of ~ 2 over saline even in the presence of elevated interstitial fluid pressurization, indicating that boundary lubrication is effective at all times. However, interstitial fluid load support remains the dominant mechanism for reducing the friction coefficient (by a factor of ~ 60 in that earlier study) as also demonstrated here, where SF was not used.

The investigation of the effect of joint congruence was motivated by the epidemiological observation that congruence might be related to the incidence and rate of evolution of osteoarthritis (OA) in some joints. For example, the thumb trapeziometacarpal joint is much less congruent than the glenohumeral joint (Ateshian et al., 1992; Soslowsky et al., 1992a). The former is also known to develop OA earlier and more frequently than the latter (Kelsey, 1982). Similarly, iatrogenic factors that compromise joint congruence, such as meniscectomies, are known to increase the incidence of OA (Lanzer and Komenda, 1990). Therefore, the investigation of the frictional response as a function of congruence may help determine whether the incidence of OA is related to loss of interstitial fluid pressurization.

The very low, sustained value of μ observed in low congruence human-bovine hybrid joints (LCMS in Figure 4 and Figure 5) was initially unexpected, in light of the above discussion of the role of joint congruence for sustaining interstitial fluid pressurization. The results for whole bovine joints show that bovine cartilage does not exhibit superior lubrication to whole human joints (bHCMS versus hHCMS in Figure 5), precluding the explanation that low friction in the hybrid LCMS group resulted from using bovine cartilage. Therefore, upon further consideration, we believe that the relatively incongruent contact made it easy for the human humeral head to roll back and forth over the bovine glenoid, as the pendulum swung. (In contrast, the highly congruent native human and bovine joints experienced pure sliding contact, as ball-and-socket joints normally do, Figure 7.) Rolling friction is notoriously smaller than sliding friction (Johnson, 1985), as also suggested from contact mechanics of porous media (Ateshian and Wang, 1997), consistent with the findings for this group.

In conclusion, this study exposed two remarkable findings: The friction coefficient of whole human joints may average less than ~ 0.006 for up to 24 h of continuous loading in saline; in the first half hour of loading, this friction coefficient may average an astoundingly low value of ~ 0.0015 . These results shed new insights into the lubrication of whole joints over the range of normal activities of daily living.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

Research reported in this publication was supported by the National Institute of Arthritis and Musculoskeletal and Skin Diseases of the National Institutes of Health under Award Number R01AR043628. The content is solely the responsibility of the authors and does not necessarily represent the official views of the National Institutes of Health.

References

- Ateshian GA. The role of interstitial fluid pressurization in articular cartilage lubrication. *J Biomech.* 2009; 42:1163–1176. PMID: PMC2758165. [PubMed: 19464689]
- Ateshian GA, Lai WM, Zhu WB, Mow VC. An asymptotic solution for the contact of two biphasic cartilage layers. *J Biomech.* 1994; 27:1347–1360. [PubMed: 7798285]
- Ateshian, GA.; Mow, VC. Lubrication and wear of diarthrodial joints. In: Mow, VC.; Huiskes, R., editors. *Basic orthopaedic biomechanics & mechano-biology*. Philadelphia, Pa.; London: Lippincott Williams & Wilkins; 2005.
- Ateshian GA, Rosenwasser MP, Mow VC. Curvature characteristics and congruence of the thumb carpometacarpal joint: differences between female and male joints. *J Biomech.* 1992; 25:591–607. [PubMed: 1517255]
- Ateshian GA, Wang H. A theoretical solution for the frictionless rolling contact of cylindrical biphasic articular cartilage layers. *J Biomech.* 1995; 28:1341–1355. [PubMed: 8522547]
- Ateshian GA, Wang H. Rolling resistance of articular cartilage due to interstitial fluid flow. *Proc Inst Mech Eng [H]*. 1997; 211:419–424.
- Ateshian GA, Wang H, Lai WM. The role of interstitial fluid pressurization and surface porosities on the boundary friction of articular cartilage. *J Tribol.* 1998; 120:241–248.
- Caligaris M, Ateshian GA. Effects of sustained interstitial fluid pressurization under migrating contact area, and boundary lubrication by synovial fluid, on cartilage friction. *Osteoarthritis Cartilage.* 2008; 16:1220–1227. PMID: PMC2622427. [PubMed: 18395475]

- Caligaris M, Canal CE, Ahmad CS, Gardner TR, Ateshian GA. Investigation of the frictional response of osteoarthritic human tibiofemoral joints and the potential beneficial tribological effect of healthy synovial fluid. *Osteoarthritis Cartilage*. 2009; 17:1327–1332. PMID: PMC2753744. [PubMed: 19410031]
- Carter MJ, Basalo IM, Ateshian GA. The temporal response of the friction coefficient of articular cartilage depends on the contact area. *J Biomech*. 2007; 40:3257–3260. PMID: PMC2094001. [PubMed: 17490673]
- Charnley J. The lubrication of animal joints in relation to surgical reconstruction by arthroplasty. *Ann Rheum Dis*. 1960; 19:10–19. [PubMed: 13809345]
- Clarke IC, Contini R, Kenedi RM. Friction and wear studies of articular cartilage: a scanning electron microscope study. *Journal of Tribology*. 1975; 97:358–366.
- Forster H, Fisher J. The influence of loading time and lubricant on the friction of articular cartilage. *Proc Inst Mech Eng [H]*. 1996; 210:109–119.
- Johnson, KL. *Contact mechanics*. Cambridge [Cambridgeshire] ; New York: Cambridge University Press; 1985.
- Kelsey, JL. *Epidemiology of musculoskeletal disorders*. New York: Oxford University Press; 1982.
- Kim YJ, Sah RL, Grodzinsky AJ, Plaas AH, Sandy JD. Mechanical regulation of cartilage biosynthetic behavior: physical stimuli. *Arch Biochem Biophys*. 1994; 311:1–12. [PubMed: 8185305]
- Kiviranta I, Tammi M, Jurvelin J, Saamanen AM, Helminen HJ. Moderate running exercise augments glycosaminoglycans and thickness of articular cartilage in the knee joint of young beagle dogs. *J Orthop Res*. 1988; 6:188–195. [PubMed: 3278079]
- Krishnan R, Kopacz M, Ateshian GA. Experimental verification of the role of interstitial fluid pressurization in cartilage lubrication. *J Orthop Res*. 2004; 22:565–570. [PubMed: 15099636]
- Lanzer WL, Komenda G. Changes in articular cartilage after meniscectomy. *Clin Orthop Relat Res*. 1990; 41–48. [PubMed: 2406073]
- Mansi S, Milosavljevic S, Baxter GD, Tumilty S, Hendrick P. A systematic review of studies using pedometers as an intervention for musculoskeletal diseases. *BMC Musculoskelet Disord*. 2014; 15:231. PMID: 4115486. [PubMed: 25012720]
- McCann L, Ingham E, Jin Z, Fisher J. Influence of the meniscus on friction and degradation of cartilage in the natural knee joint. *Osteoarthritis Cartilage*. 2009a; 17:995–1000. [PubMed: 19328878]
- McCann L, Ingham E, Jin Z, Fisher J. An investigation of the effect of conformity of knee hemiarthroplasty designs on contact stress, friction and degeneration of articular cartilage: a tribological study. *J Biomech*. 2009b; 42:1326–1331. [PubMed: 19380137]
- McCann L, Udofia I, Graindorge S, Ingham E, Jin Z, Fisher J. Tribological testing of articular cartilage of the medial compartment of the knee using a friction simulator. *Tribology International*. 2008; 41:1126–1133.
- Northwood E, Fisher J. A multi-directional in vitro investigation into friction, damage and wear of innovative chondroplasty materials against articular cartilage. *Clin Biomech (Bristol, Avon)*. 2007; 22:834–842.
- O'Kelly J, Unsworth A, Dowson D, Hall DA, Wright V. Study of the role of synovial fluid and its constituents in the friction and lubrication of human hip joints. *Eng Med*. 1978; 7:73–83.
- Oungoulian SR, Chang S, Bortz O, Hehir KE, Zhu K, Willis CE, Hung CT, Ateshian GA. Articular cartilage wear characterization with a particle sizing and counting analyzer. *J Biomech Eng*. 2013; 135:024501. PMID: 3660847. [PubMed: 23445072]
- Oungoulian SR, Durney KM, Jones BK, Ahmad CS, Hung CT, Ateshian GA. Wear and damage of articular cartilage with friction against orthopedic implant materials. *J Biomech*. 2015
- Poppen NK, Walker PS. Forces at the glenohumeral joint in abduction. *Clin Orthop Relat Res*. 1978:165–170. [PubMed: 709928]
- Roberts BJ, Unsworth A, Mian N. Modes of lubrication in human hip joints. *Ann Rheum Dis*. 1982; 41:217–224. [PubMed: 7092334]
- Soltz MA, Basalo IM, Ateshian GA. Hydrostatic pressurization and depletion of trapped lubricant pool during creep contact of a rippled indenter against a biphasic articular cartilage layer. *J Biomech Eng*. 2003; 125:585–593. [PubMed: 14618917]

- Soslowsky LJ, Flatow EL, Bigliani LU, Mow VC. Articular geometry of the glenohumeral joint. Clin Orthop Relat Res. 1992a;181–190. [PubMed: 1446435]
- Soslowsky LJ, Flatow EL, Bigliani LU, Pawluk RJ, Ateshian GA, Mow VC. Quantitation of in situ contact areas at the glenohumeral joint: a biomechanical study. J Orthop Res. 1992b; 10:524–534. [PubMed: 1613626]
- Unsworth A, Dowson D, Wright V. Frictional behavior of human synovial joints — 1. Natural joints. J Lubr Technol 97 Ser F. 1975:369–376.

Figure 1.
Samples mounted in epoxy: human humeral head (left), human glenoid (middle), and bovine glenoid (right).

Figure 2. Schematic of friction testing apparatus and diagram of humeral head showing contact forces on the forward (f_F) and reverse (f_B) cycles of pendulum motion, and their resolution into normal (f_N) and tangential (f_T) components. (Fluid bath not shown.)

Figure 3. Human shoulder joint during friction test. Fluid bath contained physiological buffered saline, protease inhibitor, and biocide.

Figure 4.

Time-history of the friction coefficient μ for the four test groups (hHCMS and hHCLS in left panel, LCMS and bHCMS in right panel). The mean and standard deviations of μ were evaluated by first binning the data over consecutive time intervals (0, 0.5, 1, 4, 8, 12, 16, 20 and 24 h).

Figure 5.

Means and standard deviations of the friction coefficient μ for the four test groups (hHCMS, hHCLS, LCMS, bHCMS), from the Initial [0 – 0.5] h and Final [20 – 24] h binned intervals.

† for significant differences between LCMS and (hHCMS, bHCMS) ($p < 0.01$); across all groups, initial values were lower than final values ($p < 0.0001$).

Figure 6.

Representative histological section of a human glenoid articular layer following 24 h of friction testing, stained with Safranin-O (red, showing the distribution of charged proteoglycans) and Fast Green. No damage to the articular surface was apparent.

Figure 7. Schematic representation of pendulum motion with congruent (left) and incongruent (right) contact configurations. As the pendulum is lifted against gravity by a steel cable (see Figure 2), a perfectly congruent ball-and-socket joint only allows pure sliding whereas an incongruent joint may allow pure rolling.