

Comparative Study of the Intrinsic Mechanical Properties of the Human Acetabular and Femoral Head Cartilage

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Summary: Biphasic creep indentation methodology and an automated indentation apparatus were used to measure the aggregate modulus, Poisson's ratio, permeability, thickness, creep and recovery equilibrium times, and percentage of recovery of normal articular cartilage in 10 human hip joints. These properties were mapped regionally to examine the mechanical factors involved in the development of site-specific degenerative lesions in the acetabulum and femoral head. The results indicate that there are significant differences between these properties regionally in the acetabulum and femoral head and between the two anatomical structures. Specifically, it was found that cartilage in the superomedial aspect of the femoral head has a 41% larger aggregate modulus than its anatomically corresponding articulating surface in the acetabulum. In addition, the superomedial aspect of the femoral head has the greatest aggregate modulus (1.816 MPa) within the hip joint. During sitting, the inferior portion of the femoral head is in contact with the anterior acetabulum, and the anterior acetabulum has a 53% greater aggregate modulus than the inferior femoral head. This area below the fovea on the femoral head has the least aggregate modulus (0.814 MPa) within the hip joint. These mismatches in the compressive modulus of opposing articulating surfaces may contribute to degeneration of cartilage in the superomedial acetabulum and the inferior femoral head. Our findings support the clinical observation that these areas are frequent sites of early degeneration.

Degeneration of articular cartilage usually occurs at certain sites in the femoral head and the acetabulum. In the femoral head, degenerative changes of the articular cartilage are more frequent in the inferior portion below and immediately around the fovea (12,13,22,31,34) and in the peripheral area near the edges (12,22,34). In contrast, the superior aspect of the femoral head, which is a predominantly weight-bearing area (20,22,31), usually is less affected (31,34). Its corresponding site in the triangular superior portion or dome of the acetabulum exhibits a greater prevalence of degeneration

(12,13,34). These degenerative changes of articular cartilage in the human hip are attributed to mechanical (1,23,35), biochemical (35), and pathological (35) factors.

Biomechanically, development of elevated stresses at various sites in the joint is believed to contribute to degenerative diseases (1,23). For example, in grossly normal human cadavers, maximum local stresses were found superiorly in the acetabular dome (1,10), the most common location of cartilage loss (1,12,13,34). The highest *in vivo* stresses in a human hip joint instrumented with a rigid endoprosthesis were in the superior and posterior portions of the acetabulum (23), which again correspond to sites of degenerative changes. *In vivo* stresses in cartilage of the hip joint vary both by region (1,23,42) and in magnitude (10,16,20). It has been suggested that regional variations in stress may be due to the geom-

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etry of cartilage (19,42) or that local stress concentrations may be influenced by differences in the mechanical properties of cartilage (8).

It is conceivable that regional differences in the intrinsic biochemical and biomechanical characteristics of cartilage of the human hip may contribute to local or early degeneration. Differences in the proteoglycan and water contents between the superior and anteroinferior portions of normal cartilage of the human femoral head have been reported (41). Slight regional variations also are known to occur in the glycosaminoglycan content of normal acetabular cartilage and between that of femoral head cartilage and acetabular cartilage (31). Regional and depth variations also have been shown to be present in the collagen content of human normal articular cartilage (27,38).

These variations in the biochemical composition of normal articular cartilage suggest regional variations of the intrinsic mechanical properties as well. Kempson et al. measured the "creep or 2 second modulus" in the human femoral head and reported that the stiffest cartilage is located in the superior portion extending to the anterior and posterior aspects, whereas the softest cartilage is at or near the fovea (26). Cameron et al. found a corresponding pattern of stiffness on the femoral head (15). Roberts et al., using the 2 second modulus technique of Kempson et al. (25), found a difference in compressive stiffness between the superior and anteroinferior portions of the femoral head (40,41). Roentgenographic measurements of deformations of femoral head cartilage demonstrated a lack of uniformity in cartilage compliance (3). We are not aware of any studies examining the mechanical properties of articular cartilage in the acetabulum or of any comparative studies of the mechanical properties of cartilage in the acetabulum and the femoral head.

Therefore, our objective was to compare regional variations in the intrinsic mechanical properties of normal articular cartilage in the human femoral head and matching acetabulum, as well as to perform pooled comparisons using structure, side, and sex as factors. A second objective was to compare the material properties at various anatomic positions in order to examine the hypothesis that the site-specificity of degeneration of hip cartilage is related to a disparity in these properties.

MATERIALS AND METHODS

Ten fresh-frozen human femoral heads, with corresponding acetabula, were obtained from the Mus-

culoskeletal Transplant Foundation (Holmdel, NJ, U.S.A.). The 10 hips came from five donors, ages 24 (man), 28 (woman), 29 (man), 47 (man), and 50 years (woman). All specimens appeared grossly normal and were stored at -80°C until dissection. At the time of sectioning, each specimen was thawed in a solution of normal saline with protease inhibitors (*N*-ethylmaleimide, 10 mM; benzamidine HCl, 5 mM; EDTA, 2 mM; and phenylmethylsulfonyl fluoride, 1 mM) at room temperature for 1 hour. The India ink staining technique (33) then was performed to examine gross morphology of all joint surfaces and to identify normal areas for indentation tests. Each femoral head and acetabulum was sectioned into eight and six osteochondral specimens, respectively (Figs. 1 and 2), on the basis of the anatomic position in which the superior portions of the femoral head and acetabulum were in contact. Accordingly, both the posterior and anterior aspects of the femoral head were in contact with those in the acetabulum. Sectioning was done under continual irrigation with normal saline solution containing protease inhibitors to maintain proper tissue hydration. It previously had been shown that edge effects on the stress and strain fields in articular cartilage under biphasic creep indentation are negligible for osteochondral specimens with a radius 3-4 times larger than the radius of the indenter tip (7). All osteochondral specimens were at least 15×15 mm, whereas the indenter tip had a radius of 1.0 mm. Each osteochondral specimen was wrapped in gauze moistened with the solution and was stored in double-sealed plastic bags at -80°C . At the time of indentation testing, each specimen was thawed again for 1 hour in normal saline solution with protease inhibitors. Each osteochondral specimen was exposed to two freeze-thaw cycles, which are not believed to affect the tissue's intrinsic material properties (6,24,30).

The osteochondral specimens were tested with a new automated indentation apparatus to obtain the *in situ* creep and recovery behavior of articular cartilage on the joint surface. Our design was based on technology developed for the creep biphasic indenter (37) for indentation testing of articular cartilage. At the time of testing, each specimen was mounted with cyanoacrylate cement in a bath of adjustable depth, containing normal saline solution with protease inhibitors. Care was exercised to test only normal surfaces, as determined with the India ink staining technique (33). A fiberoptic positioning system in line with the loading shaft emitted light that illuminated a circular area, 2.0 mm in di-

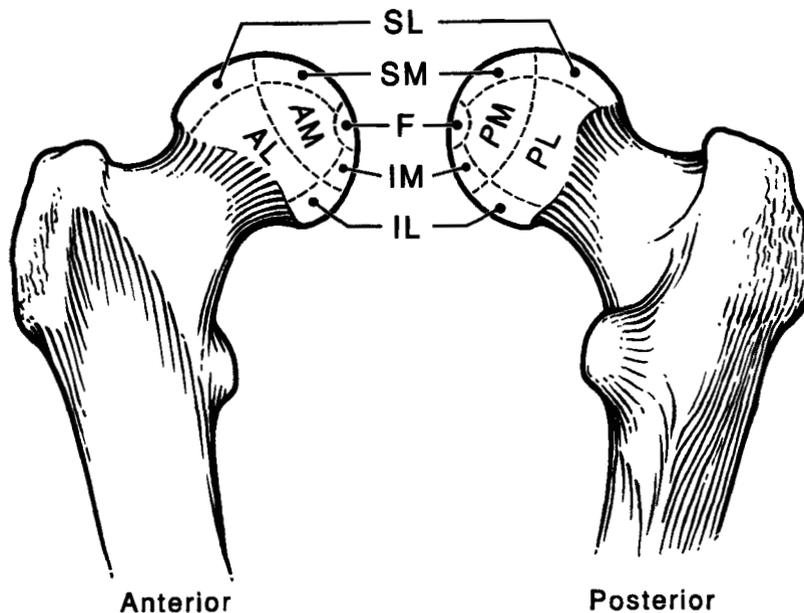


FIG. 1. Three-dimensional schematic depiction of test sites on the femoral head. A = anterior, S = superior, P = posterior, I = inferior, M = medial, L = lateral, and F = fovea.

ameter, in the desired portion of the articular surface. With the aid of a vertical and horizontal lead screw assembly and a spherical joint, each specimen was translated and rotated until the reflected light was collected by light receivers circumferentially located around the fiberoptic light source. This indicated that the loading shaft was perpendicular to the cartilage surface. Perpendicularity was achieved within 20 seconds. A 0.0687 N tare load was applied via a flat-ended, cylindrical, rigid, porous indenter tip, 2.0 mm in diameter, and computer-based data acquisition commenced. The slope of the tare load creep was monitored, and when the slope became smaller than 1×10^{-6} mm/sec, the computer sent a binary signal to a solenoid-controlled vacuum line, which released a 0.438 N test load onto the loading shaft and, thus, onto the articular surface. This test load was positioned accurately with a precise bevel gear, and it was released from a height of approximately 70 μ m. Deformation of the tissue was monitored with a linear variable differential transformer. Its output was collected by a computer board at a resolution of 0.3 mV, which corresponded to a 0.25 μ m deformation-resolution. Data points were collected and plotted on the screen every 2.5 μ m or every 100 seconds, whichever happened first. When the slope of the creep curve was smaller than 1×10^{-6} mm/sec, the solenoid automatically turned on the vacuum line, and the load was removed.

The tissue was allowed to recover, and when the

slope criterion was satisfied, data acquisition ceased automatically. Air bearings facilitated movement of the loading shaft and decreased frictional resistance to less than 9.81×10^{-4} N. The air supplied to the bearings initially was passed through a 5 μ m particle filter and two 0.1 μ m coalescing filters for complete

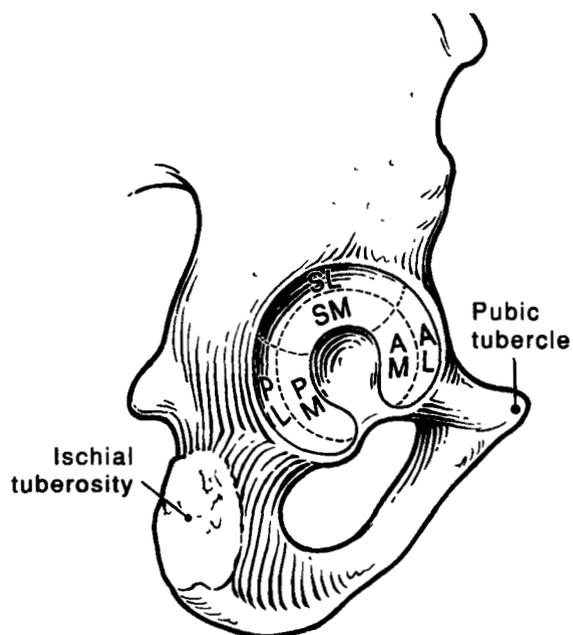


FIG. 2. Three-dimensional schematic depiction of test sites on the acetabulum. A = anterior, S = superior, P = posterior, M = medial, and L = lateral.

TABLE 1. *Intrinsic material properties (mean \pm SD) and thickness of articular cartilage*

	Aggregate modulus (MPa)	Poisson's ratio	Permeability ($\times 10^{-15}$)(m ⁴ /N·s)	Thickness (mm)
Femoral head (n = 10)				
Anteromedial	1.198 \pm 0.245	0.055 \pm 0.073	0.906 \pm 0.497	1.84 \pm 0.17
Anterolateral	1.067 \pm 0.561	0.055 \pm 0.069	0.940 \pm 0.409	1.23 \pm 0.28
Superomedial	1.816 \pm 0.868	0.058 \pm 0.074	1.002 \pm 0.576	1.66 \pm 0.13
Superolateral	1.054 \pm 0.742	0.058 \pm 0.069	0.814 \pm 0.262	1.11 \pm 0.23
Posteromedial	1.555 \pm 0.603	0.013 \pm 0.028	1.101 \pm 0.610	1.79 \pm 0.24
Posterolateral	1.169 \pm 0.628	0.039 \pm 0.060	0.781 \pm 0.450	1.11 \pm 0.23
Inferomedial	0.948 \pm 0.238	0.047 \pm 0.048	0.880 \pm 0.470	1.42 \pm 0.27
Inferolateral	0.679 \pm 0.162	0.045 \pm 0.064	0.899 \pm 0.444	1.03 \pm 0.16
Acetabulum (n = 10)				
Anteromedial	1.340 \pm 0.739	0.045 \pm 0.051	0.826 \pm 0.365	1.22 \pm 0.19
Anterolateral	1.147 \pm 0.495	0.034 \pm 0.054	0.710 \pm 0.362	1.26 \pm 0.15
Superomedial	1.284 \pm 0.650	0.055 \pm 0.075	1.133 \pm 1.114	1.11 \pm 0.34
Superolateral	1.072 \pm 0.546	0.011 \pm 0.022	0.983 \pm 0.609	1.83 \pm 0.45
Posteromedial	1.424 \pm 0.602	0.097 \pm 0.077	0.737 \pm 0.588	1.06 \pm 0.24
Posterolateral	1.147 \pm 0.444	0.019 \pm 0.025	0.819 \pm 0.440	1.09 \pm 0.23

removal of liquid and microparticles and was regulated to a pressure of 552 kPa. Once recovery equilibrium had been achieved, the test site was marked with India ink, and the sample bath was translated to an apparatus in which thickness was measured with a needle probe system (36,37). All data acquisition and control software was written with use of the Labview icon environment (National Instruments, Austin, TX, U.S.A.).

The creep indentation problem—in which a frictionless, porous tip is used to apply a step load—was solved by Mak et al. (29) using the linear biphasic theory (36). Subsequently, Mow et al. developed a numerical algorithm to compute the three intrinsic coefficients (37) required by the theory. This methodology was used in the past, along with the creep indentation experiment, to obtain the mechanical properties of cartilage (6,17,37).

Three separate four-way analysis of variance (ANOVA) procedures were performed to study the effects of the following independent factors—side (right or left), donors, quadrants (anterior, posterior, superior, or inferior), and location (medial or lateral)—on the dependent variables (aggregate modulus, permeability, thickness, creep equilibrium time, recovery equilibrium time, and percentage of recovery) for each structure (femur and acetabulum) individually and then again for both structures together. In addition, a fourth separate four-way ANOVA was performed to look at the effect of structure, side, donors, and location on the dependent variables. The Student *t* test was performed to compare these dependent variables between men

and women. Due to the non-normal frequency distribution of the values for Poisson's ratio, nonparametric Wilcoxon signed rank tests were performed for this dependent variable. A separate nonparametric Wilcoxon signed rank test also was performed to compare creep and recovery equilibrium times. The statistical significance level was set at $p < 0.05$ for all tests. Fisher's least significant difference multiple comparisons test of the means was applied when the F test in the ANOVA was significant. No further *post hoc* tests were run if the F test was not significant. Simple linear and polynomial correlation analyses also were performed, between all parameters measured, to identify any significant relationships.

RESULTS

The intrinsic material properties and thickness of articular cartilage in the femoral head and acetabulum are given in Table 1. For the aggregate modulus, variability due to donors, quadrants, and location was significant; for permeability, donor variability was significant; and for thickness, variability due to donor, structure, quadrant, and location was significant. Side was not found to be significant.

In the femoral head, the superomedial aspect had the largest aggregate modulus ($p < 0.05$) and the inferolateral portion had the smallest ($p < 0.05$). The anterior and inferior quadrants were, respectively, the thickest and thinnest areas ($p < 0.05$). In comparisons of specific sites, the thickest and thinnest were the anteromedial and inferolateral regions, respectively ($p < 0.05$). Medial specimens pooled together had a 39% larger aggregate modulus ($p < 0.05$) and

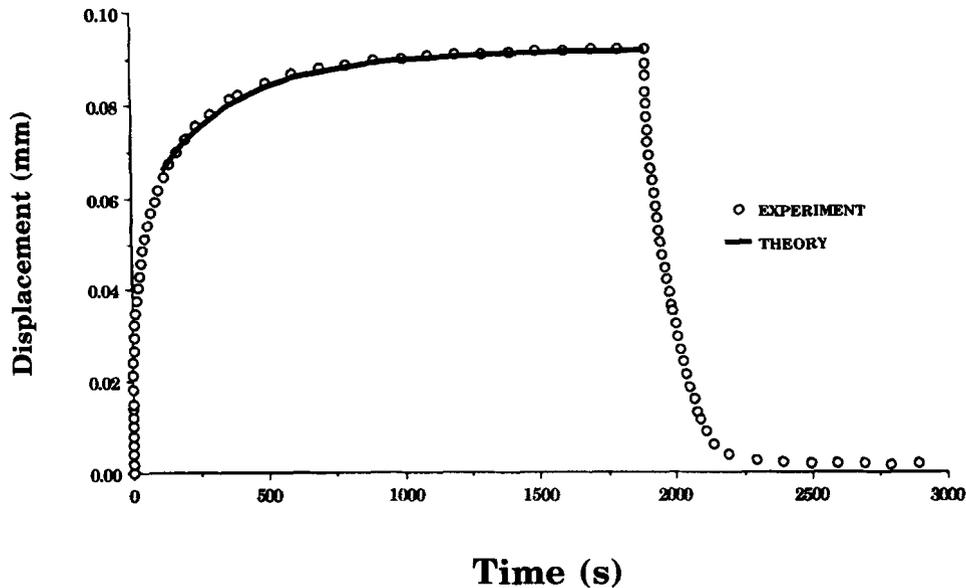


FIG. 3. A typical creep-recovery curve in the superomedial portion of the femoral head with a corresponding curvefit of the creep portion. The data are from the hip of a 29-year-old man.

49% greater thickness ($p < 0.05$) than the lateral specimens.

In the acetabulum, the superolateral cartilage was the thickest ($p < 0.05$). No significant regional variations between acetabular test sites were found in the aggregate modulus or in permeability. All medial specimens pooled together had a 20% greater aggregate modulus and 19% less thickness ($p < 0.05$), and Poisson's ratio was three times as large as that for the lateral sites ($p < 0.01$).

In comparisons of all sites in both the femoral head and acetabulum, the inferior and superior aspects of the femoral head had the smallest and largest aggregate moduli, respectively ($p < 0.05$). Similarly, the thickest cartilage was in the antero-medial site of the femoral head and the thinnest tissue was in the inferolateral site ($p < 0.05$). The superomedial aspect of the femoral head had a 41% larger aggregate modulus and was 50% thicker than the respective site on the acetabulum ($p < 0.05$). In addition, Poisson's ratio of the posteromedial aspect of the acetabulum was 0.084 larger than that of the posteromedial aspect of the femoral head ($p < 0.03$). The anterior acetabulum had a 53% larger aggregate modulus than the inferior femoral head ($p < 0.05$). The lateral specimens of the acetabulum pooled together were 24% thicker ($p < 0.05$) and Poisson's ratio was 0.028 smaller ($p < 0.05$) than for their counterparts on the femoral head. The medial specimens of the femoral head pooled together, however, were 48% thicker ($p < 0.05$) than those on the acetabulum.

The femoral specimens pooled together were 11% thicker ($p < 0.01$) than the acetabular specimens, although no differences were shown with regard to intrinsic material properties.

A typical creep-recovery curve of the superomedial aspect of the femoral head of a 29-year-old man is shown in Fig. 3. For all test sites, the average amount of recovery after removal of the test load was 93.4%. The mean (\pm SD) times for creep and recovery for all specimens were $3,394 \pm 1,679$ seconds and $1,727 \pm 906$ seconds, respectively. Thus, recovery was 51% faster than the creep indentation phase ($p < 0.001$). The rate of creep on load application was larger than the rate of recovery after removal of the load (Fig. 4). This may have been due to differences in permeability caused by tissue compaction under the load (6,37). As time increases, the rate of creep keeps decreasing and becomes smaller than the rate of recovery. Finally, recovery is achieved while creep keeps increasing. These differences in the kinetics of creep and recovery were observed consistently in all specimens.

Acetabular specimens achieved 94.7% recovery and femoral specimens achieved 92.0% ($p < 0.08$). In the femoral head, the inferolateral and superomedial sites took the longest time ($p < 0.05$) and shortest time ($p < 0.05$), respectively, to achieve recovery. In the acetabulum, the superolateral site took the longest to achieve recovery ($p < 0.05$). The lateral specimens of each anatomical structure took longer to recover than did each of their respective medial

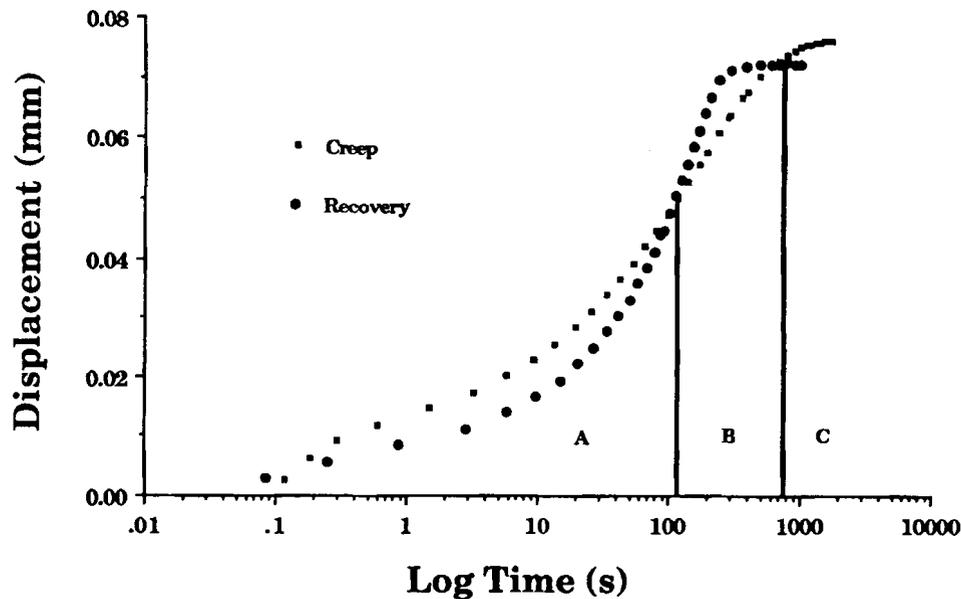


FIG. 4. There were significant differences between the rates of creep and recovery. Recovery equilibrium is achieved 51% faster than creep equilibrium. Region A indicates load apposition; region B, creep deceleration; and region C, recovery.

specimens ($p < 0.05$). Cartilage achieved creep equilibrium under the tare load in $1,049 \pm 1,008$ seconds. The experimental creep curves were fitted with the master solution via a nonlinear regression procedure (37), and the three intrinsic material properties of cartilage were obtained. The fitted creep curve (Fig. 3) is an example of the quality of curvefits observed for all specimens.

No significant correlations were found between thickness and aggregate modulus ($r^2 = 0.007$; $p > 0.3$), creep time ($r^2 = 0.001$; $p > 0.7$), recovery time ($r^2 = 0.0003$; $p > 0.8$), or percentage of recovery ($r^2 = 0.002$; $p > 0.6$). Certain relationships were found to be significant. The aggregate modulus had a high negative correlation with recovery time ($r^2 = 0.494$; $p < 0.001$), a moderate negative correlation with creep time ($r^2 = 0.315$; $p < 0.001$), and a low positive correlation with percentage of recovery ($r^2 = 0.127$; $p < 0.001$). Creep equilibrium time had a low positive correlation with recovery time ($r^2 = 0.198$; $p < 0.001$) and a moderate negative correlation with percentage of recovery ($r^2 = 0.385$; $p < 0.001$).

DISCUSSION

The cartilage in the inferior region of the femoral head and the dome of the acetabulum undergo early degenerative changes. To examine whether this site-specificity of cartilage degeneration in the hip is related to a disparity in the intrinsic material properties of normal tissue between articulating sur-

faces, we mapped these properties regionally in the human hip. The results indicate that regional variations in material properties, creep characteristics, and thickness of cartilage, especially in the femoral head, are significant. Our findings clearly indicate that sites of frequent degeneration correspond with significant mismatches in mechanical properties of cartilage.

For cartilage to maintain its normal mechanical and biochemical properties, it may be necessary for the tissue to be in recurrent contact with its opposing cartilage on a regular basis. In fact, articular cartilage in areas of habitual noncontact has been shown to correspond to areas of frequent degeneration when compared with cartilage in areas of habitual contact (12,18,22). In young people, the dome of the acetabulum is a habitually noncontact or intermittently loaded area due to incongruence (10,12), whereas cartilage of the superior femoral head comes into habitual contact with the anterior and posterior surfaces of the acetabulum. An increase in congruence occurs with age (3,12), due to either actual remodeling of the hip joint (11) or an increase in tissue compliance (3) that brings the dome into more frequent contact with the superior aspect of the femoral head. Recently, it was suggested that an intermittently loaded area has a smaller compressive modulus than that of a habitually load-bearing region (6). Thus, the superior femoral head may retain its "better" mechanical properties due to habitual use, whereas

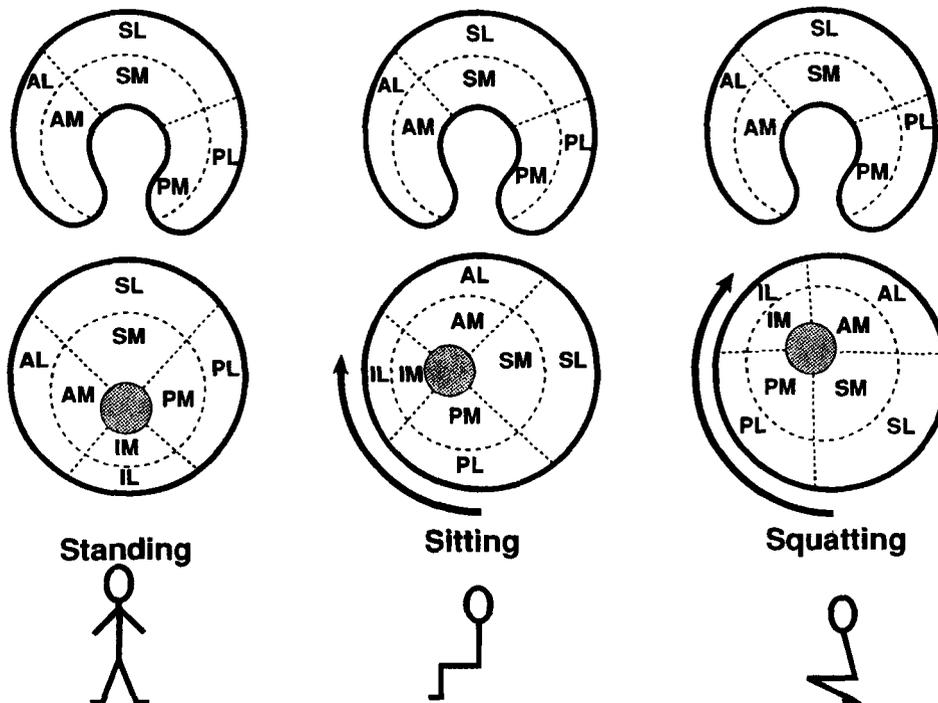


FIG. 5. The articulating relationships of femoral head and acetabular surfaces during standing, sitting, and squatting. A = anterior, S = superior, P = posterior, I = inferior, M = medial, and L = lateral.

dome cartilage has “inferior” mechanical properties due to habitual noncontact and undergoes abrasion with subsequent degeneration. The periphery of the femoral head, a noncontact area outside the rim of the acetabulum, also is a frequent site of degeneration (12,22,34), which may be explained by this theory.

The concept of use and disuse also is supported by the low frequency of osteoarthritis in populations in which the squatting position is very common (14,21). In this position, the femoral head undergoes sagittal plane rotation greater than the usual arc of 130°, bringing the superior aspect of the acetabulum into contact with segments of both the anterior and inferior femoral head. Thus, the squatting position increases the surfaces put into use both on the femoral head and acetabulum (21) and may prevent degeneration of articular cartilage.

A disparity in the mechanical properties between two articulating surfaces results in dissimilar strain fields; thus, one side is exposed to a more strenuous mechanical environment. As a result of larger strain fields, degenerative changes may be accelerated. In this study, the intrinsic material properties of cartilage were examined at sites that articulate during sitting and standing. An average person spends a significant amount of time seated with the hips

flexed 70-110°. During sitting, the anterior segment of the femoral head is in greatest contact with the superior segment of the acetabulum (Fig. 5). Additionally, the inferior segment of the femoral head is in contact with the anterior segment of the acetabulum, and the superior segment of the femoral head is in contact with the posterior aspect of the acetabulum. Our results indicate that these contact surfaces have similar compressive moduli, except for the anterior acetabulum, which has a significantly larger aggregate modulus than the inferior femoral head. During normal stance and gentle walking, the “stiff” superomedial portion of the femoral head is mostly in contact with the “soft” superomedial portion of the acetabulum (Fig. 5). The inferior aspect below the fovea and the superior portion of the acetabulum frequently are sites of early degeneration. Meachim proposed that loss of articular cartilage may be due to a type of “grinding mechanism” between two articulating joint surfaces (32). Our results suggest that a mismatch in the compressive modulus may indeed contribute to abrasion of the inferior femoral head and superior acetabulum, with subsequent degeneration of cartilage. Such a mismatch may affect joint lubrication and accentuate the detrimental effects of transarticular impact loads on cartilage degeneration (46). It also may influence the load-sharing mecha-

nism between the solid and fluid phases of articular cartilage (7).

The material properties and thicknesses of medial and lateral specimens of cartilage were significantly different. Articular cartilage from all medial sites in the acetabulum, pooled together, had a larger aggregate modulus and smaller Poisson's ratio, and was thinner than cartilage from lateral sites, which is in agreement with previous studies (42,43). In the femoral head, the medial sites had a larger aggregate modulus but were thicker than the lateral sites. A small Poisson's ratio indicates a propensity for more fluid transport through the solid matrix (6), while a larger aggregate modulus indicates decreased compliance under load (4). These variations in compressive modulus, thickness, and propensity for interstitial fluid flow may indicate differences in congruence between the outer periphery of the hip joint and the more centrally located aspects.

Regional studies have failed to show any specific differences in the distribution of lesions between the hips of men and women (13,34), even though it is generally accepted that men have a higher prevalence of primary osteoarthritis of the hip. It has been postulated that stresses in the hip joints of women are higher than those in men, due to the smaller radius of the femoral head in women (9); this suggests that a greater prevalence of osteoarthritis should be found in women. In the present study, there were no statistically significant differences between specimens from men and women in the mechanical properties of articular cartilage, although the power of our study to detect differences due to sex of the individual was low due to the small number of specimens. Further studies should be undertaken to examine variations in intrinsic material properties between the sexes.

Creep equilibrium time under the same tare load ranged from 270 to 6,730 seconds. The large standard deviation indicates that an *a priori* assumed fixed amount of time under the tare load may be insufficient for articular cartilage to achieve creep equilibrium. This also was true for test load creep and recovery equilibrium. Thus, equilibrium times for tare load creep, test load creep, and test load recovery should be determined with use of a slope-based criterion. Furthermore, friction that developed in the apparatus appeared to be negligibly small (approximately 0.2% of the applied load); therefore, it was not expected to have a significant effect on the determination of equilibria with use of the slope-criterion. Our results indicate that, as the aggregate

modulus increases, articular cartilage reaches faster creep and recovery equilibria; it also achieves a larger percentage of recovery. It has been shown that alterations in the biochemical composition of articular cartilage induce tissue degeneration, which corresponds to a loss of stiffness, strength, and other functional properties (4,5,28,39). Therefore, it is reasonable to use the aggregate modulus as a relative measure of the tissue's structural integrity. Indeed, healthier tissue may be more likely to recover fully after removal of the creep indentation load, whereas fibrillated tissue may not achieve 100% recovery, due to disruptions in its extracellular matrix.

As expected, no correlation between aggregate modulus and thickness ($r^2 = 0.007$; $p > 0.3$) was observed. Simon has shown that compressive stresses on the joint cartilage do not correlate with thickness (44,45). Akizuki et al. reported a lack of correlation between aggregate modulus and thickness in cartilage of the human tibial plateau (2), and Athanasiou et al. found a repeating pattern but no significant correlation (6). Our findings are in agreement with these prior investigations.

In this study, we discovered that the thickness and some of the material properties of articular cartilage in the human hip joint differ from those in the human knee joint. Specifically, for human acetabular and femoral head cartilage ($n = 140$), the aggregate modulus was 1.207 ± 0.606 MPa, Poisson's ratio was 0.045 ± 0.060 , permeability was $0.895 \pm 0.537 \times 10^{-15}$ m⁴/N·s, and tissue thickness at the test site was 1.34 ± 0.38 mm. For cartilage of the human knee ($n = 14$), the corresponding values are 0.604 ± 0.154 MPa, 0.060 ± 0.074 , $1.446 \pm 0.609 \times 10^{-15}$ m⁴/N·s, and 2.631 ± 1.04 mm (6). Thus, cartilage in the human hip joint is twice as stiff, less permeable, and half as thick compared with cartilage in the knee. These variations may be indicative of differences in *in vivo* stress distributions between the two joints, and careful comparative studies are needed to verify and elucidate these differences. It should be noted that, in the knee study, a predetermined amount of time was used as a creep and recovery equilibrium endpoint, whereas in the present study the slope-based creep equilibrium criterion was used.

This slope-based criterion represents a strength of the present study because, through its use, we eliminated the variability introduced by the arbitrary determination of equilibria. Other strengths include the use of specimens of cartilage from relatively young and disease-free individuals, a fiberoptic positioning system that obviates the subjective manual

determination of indenter perpendicularity, and the automated data-acquisition and control system. A limitation of the study is the inclusion of a relatively small number of specimens and the use of both left and right sides in the statistical analyses.

This study, which represents the first comprehensive examination of the biomechanical properties of articular cartilage in the human hip, contributes to a deeper understanding of the mechanics of hip cartilage and may help to elucidate the mechanical factors involved in the etiology of osteoarthritis of the hip. This knowledge may be incorporated in the design process of the artificial articulating surfaces of hip prostheses so that they can mimic normal joint functions more closely. New materials for partial replacement of hip cartilage that exhibit mechanical behavior similar to that of the adjacent or opposing cartilage also can be developed. Furthermore, in the design of osteotomies of the femoral head or acetabulum, the orthopaedic surgeon should account for the stiffness mismatch between the proposed articulating cartilage surfaces; an osteotomy that brings into contact surfaces of unequal stiffness may accelerate abrasion of the softer cartilage. Knowledge of the regional variations in the mechanical properties of articular cartilage in the hip joint gives us a better understanding of the *in vivo* and *in situ* behavior of tissue.

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