Effect of Acetabular Cup Abduction Angle on Wear of Ultrahigh-Molecular-Weight Polyethylene in Hip Simulator Testing

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Abstract

The effect of acetabular component positioning on the wear rates of metal-on-polyethylene articulations has not been extensively studied. Placement of acetabular cups at abduction angles of more than 40° has been noted as a possible reason for early failure caused by increased wear.

We conducted a study to evaluate the effects of different acetabular cup abduction angles on polyethylene wear rate, wear area, contact pressure, and contact area. Our in vitro study used a hip joint simulator and finite element analysis to assess the effects of cup orientation at 4 angles (0°, 40°, 50°, 70°) on wear and contact properties. Polyethylene bearings with 28-mm cobalt-chrome femoral heads were cycled in an environment mimicking in vivo joint fluid to determine the volumetric wear rate after 10 million cycles. Contact pressure and contact area for each cup abduction angle were assessed using finite element analysis. Results were correlated with cup abduction angles to determine if there were any differences among the 4 groups.

The inverse relationship between volumetric wear rate and acetabular cup inclination angle demonstrated less wear with steeper cup angles. The largest abduction angle (70°) had the lowest contact area, largest contact pressure, and smallest head coverage. Conversely, the smallest abduction angle (0°) had the most wear and most head coverage.

Polyethylene wear after total hip arthroplasty is a major cause of osteolysis and aseptic loosening, which may lead to premature implant failure. Several studies have found that high wear rates for cups oriented at steep angles contributed to their failure. Our data demonstrated that larger cup abduction angles were associated with lower, not higher, wear. However, this potentially "protective" effect is likely counteracted by other complications of steep cup angles, including impingement, instability, and edge loading. These factors may be more relevant in explaining why implants fail at a higher rate if cups are oriented at more than 40° of abduction.

Bearing surface wear continues to be a major source of implant failure after total hip arthroplasty. Premature implant failure represents a marked burden on both patients and surgeons. Kurtz and colleagues[1] projected that, over the next 2 decades, the number of revisions will double from about 50,000 annually. In the setting of metal-on-polyethylene bearings, generation of particulate wear debris has been recognized as a primary cause of aseptic loosening and osteolysis leading to implant failure in the United States and Europe.[2-9] Osteolysis is commonly observed with polyethylene wear rates of more than 0.1 mm/y. However, the cause of polyethylene wear is multifactorial; implant, surgical, and patient-related factors are involved.[6] Understanding how these influence wear and prosthesis failure is imperative if the revision burden is to be reduced.

The characteristic patterns of polyethylene wear were initially described by Dowling and colleagues,[10] who demonstrated wear primarily in the superior aspect of the acetabular cup. More recently, malpositioned components, particularly cups placed at more than 50° of abduction, have been implicated in poor clinical outcomes and early failure from impingement, recurrent dislocation, and increased wear.[11] This has been observed in the high rates of volumetric wear and premature failure of metal-on-metal articulations, likely caused by edge loading, increased friction with large-diameter heads, and corrosion at the taper junction.[12-14] Earlier, Schmalzried and colleagues[15] reported similar findings with ultrahigh-molecular-weight polyethylene (UHMWPE) bearings. They

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observed significantly more ischial osteolysis when cups were implanted at more than 50° of abduction. Several other studies[16-20] have provided support for this finding; however, inherent in these studies are multiple variables (eg, cup migration, anteversion, creep) that, in addition to abduction angle, can effect wear. Few investigators have tried to assess the effects of various cup orientations (while controlling for such factors) to determine whether there is a correlation between increasing cup abduction and polyethylene wear. A further consideration is that newer materials (eg, highly cross-linked polyethylene) have different wear properties. Thus, although early-generation materials with lower fatigue strength and higher wear potential may have been affected by varying cup orientation, this might not be the case for modern materials, which have much higher fatigue and wear resistance. All these variables can be predictably controlled and studied in an in vitro setting.

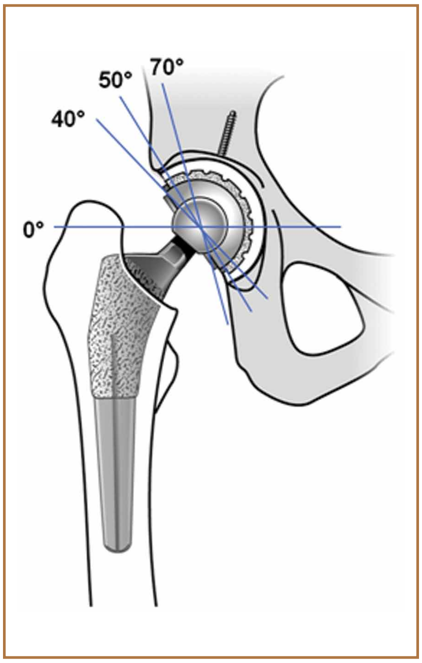

Figure 1. Schematic diagram of 4 acetabular cup abduction angles during hip simulator testing.

We conducted a study to assess the effects of various acetabular component abduction angles on wear rates using finite element analysis (FEA) and an in vitro mechanical testing simulator that mimics in vivo conditions. Variables assessed in this study were effect of changes in acetabular cup abduction on volumetric wear rates; macroscopic assessment of wear location; location and distribution of contact pressure; maximum contact pressure recorded for each cup orientation; and mean contact area of each cup orientation.

Materials and Methods

This study of metal-on-polyethylene wear properties as a function of acetabular cup abduction angle was conducted in 2 parts. Part 1 was an in vitro analysis using a hip joint simulator that mimicked in vivo conditions to quantify polyethylene wear and describe the distribution of wear within the cup. Part 2 was an FEA performed for each acetabular cup position to determine the location and distribution of contact pressure, maximum contact pressure, and mean contact area.

Hip Joint Simulation

The effect that acetabular cup abduction has on polyethylene wear was measured using a hip joint simulator (MTS, Eden Prairie, Minnesota) with 40 commercially available highly cross-linked polyethylene femoral liners (Trident; Stryker Orthopaedics, Mahwah, New Jersey) and matched-diameter (28-mm) cobalt-chrome femoral heads. The polyethylene liners had an inner diameter of 28 mm and were machined from GUR 1050 compression-molded polyethylene to the final dimensions. All the inserts were vacuum-packaged after a nitrogen gas flush and were sterilized by gamma irradiation (30 kGy) in a manner identical to that used for implants in the clinical setting. This material, referred to as *conventional UHMWPE*, was melted, irradiated, and used to magnify any wear effect.

The acetabular cups were mounted in titanium acetabular shells, which were then seated in UHMWPE fixtures using titanium bone screws. Matched-diameter cobalt-chrome femoral heads were mated with the inserts.

All cups were weighed, fixed, and positioned superiorly to their matched femoral heads at a neutral version angle. Cup orientations were divided into 4 groups of angles of inclination from the horizontal plane: 0°, 40°, 50°, and 70° (10 per group; **Figure 1**). As the joint force in vivo is about 10° to 15° medial to the superior direction,[21] cup inclination angles of 0°, 40°, 50°, and 70° simulate and correspond to in vivo conditions of about 10° to 15°, 50° to 55°, 60° to 65°, and 80° to 85° of abduction, respectively, because of the vertical load path used by the hip joint simulator. Although the exact clinical implication of this small difference may be difficult to interpret, it is relevant to cup device locking mechanism stress response.

The standardized orbital-bearing–type hip joint simulator applies a biaxial rocking motion to the femoral head by a rotating block inclined at 23°—thereby imparting combined flexion–extension, abduction–adduction, and internal–external rotation.[22] A physiologic load based on the method originally described by Paul[23] was applied to each bearing couple with a hydraulic actuator. The minimum and maximum loads used during the simulation were 50 N and 2450 N, respectively. Testing was conducted at 1 Hz for 3 million cycles using a joint fluid analogue consisting of alpha calf fraction serum (Hyclone Labs, Logan, Utah) diluted to 50% with a pH-balanced 20-mmol solution of deionized water and ethylenediaminetetraacetic acid, or EDTA (protein level, ≈20 g/L).[24,25] The serum was changed every 500,000 cycles.

All samples were analyzed at 500,000 cycle intervals, during which they were cleaned using nonabrasive techniques and weighed. The weight at the end of each cycle period was compared with the initial weight and converted to volume, which was then plotted as a function of cycle count. All inserts were also macroscopically analyzed for any gross damage or areas of deformation. Any surface damage, such as cracking or pitting, was recorded. Three of us examined and inspected all the liners using a standard microscope at ×4 magnification. All 3 researchers had prior experience in examining liners.

In all cases, their agreement was complete.

Finite Element Analysis

Static FEA (**Figure 2**) was conducted for each abduction angle studied on the hip simulator. The components used for the FEA model were of the same size and design as those used in the simulator study. All analyses were done using Workbench v11.0 (Ansys, Canonsburg, Pennsylvania).

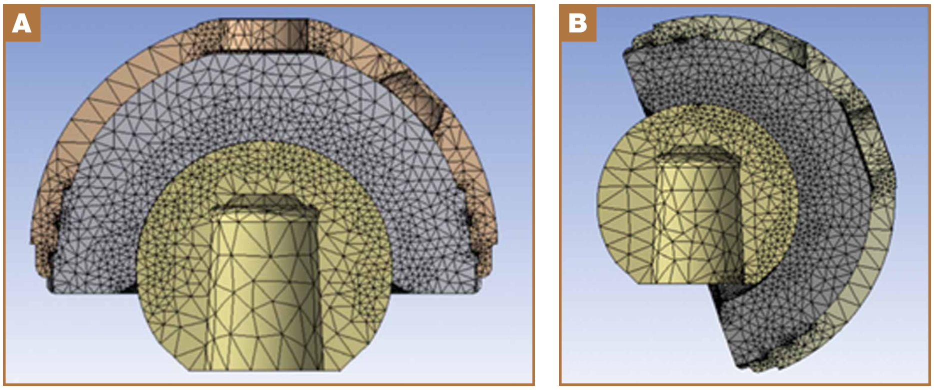

Figure 2. Finite element analysis models of (A) 0° and (B) 70° of abduction.

The models were meshed with 10-noded solid tetrahedral elements, and the size of the contact elements between the femoral head and the acetabular cup was set at 1 mm. The material properties of the digital polyethylene liner were modeled using experimental stress–strain data obtained from testing polyethylene cylinders. The femoral component and the acetabular shell were modeled as rigid bodies to reduce computation time. This minor simplification is justified by the significantly higher stiffness of cobalt-chrome relative to polyethylene; the higher stiffness causes almost no deformation of the metal bearing surface during articulation in vivo. Similarly, the outside of the acetabular shell was also rigidly fixed.

The contact between the metal femoral head and the polyethylene cup was modeled as a frictionless interface to allow for further simplification of the model, as FEA addressed only contact pressure and not wear. A frictionless support was applied to all symmetric surfaces, and a remote displacement was applied to the femoral head to constrain its rotation. The model was loaded to a maximum load of 2450 N, which was estimated using the Paul loading curve.[23] Contact stresses (measured in megapascals) using this load were calculated between the polyethylene liner and the femoral head, and the contact area (measured in square millimeters) on the polyethylene liner was calculated by quantifying the number of elements in contact with the femoral head.

Data and Statistical Analysis

All study data were collected and then extracted to an Excel spreadsheet (Microsoft Corporation, Redmond, Washington). Mean volumetric wear, mean maximum contact pressure, and mean contact area for each acetabular cup abduction angle were calculated, and an analysis of variance was performed to determine if a significant difference existed between these groups. Significant differences in mean reported values were set at $P < .05$.

Results

Results of the in vitro hip joint simulator test demonstrated a strong correlation between increased acetabular cup abduction and volumetric polyethylene wear measured at 1 million cycles ($R^2 = 0.05$; **Figure 3**). The relationship between abduction angle

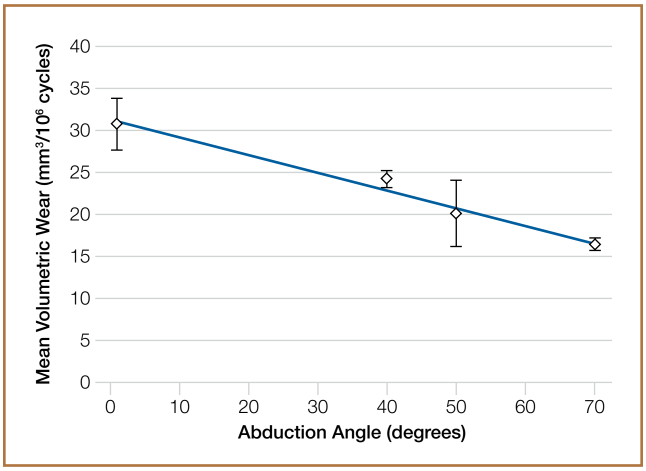

Figure 3. Wear rates for 4 abduction angles after 3 million cycles of loading. Vertical error bars represent SDs. Correlation coefficient (R^2) of linear regression line is 0.98.

and volumetric wear was inverse, with higher acetabular cup abduction angles associated with significantly ($P < .001$) less volumetric wear. The most wear was found in cups oriented at 0° (mean [SD], 30.8 [3.1] $mm^3/10^6$ cycles), the least in cups oriented at 70° (mean [SD], 16.3 [0.7] $mm^3/10^6$ cycles; **Table 1**), which corresponds to a 47% lower wear rate.

Macroscopic inspection of the polyethylene cups revealed

Table 1. Volumetric Wear as Function of Abduction Angle[a]

	Mean Volumetric Wear, $mm^3/10^6$ cycles	
Abduction Angle, °	Mean	SD
0	30.8	3.1
40	24.2	1.0
50	20.1	4.0
70	16.3	0.7

[a]All *P*s < .001.

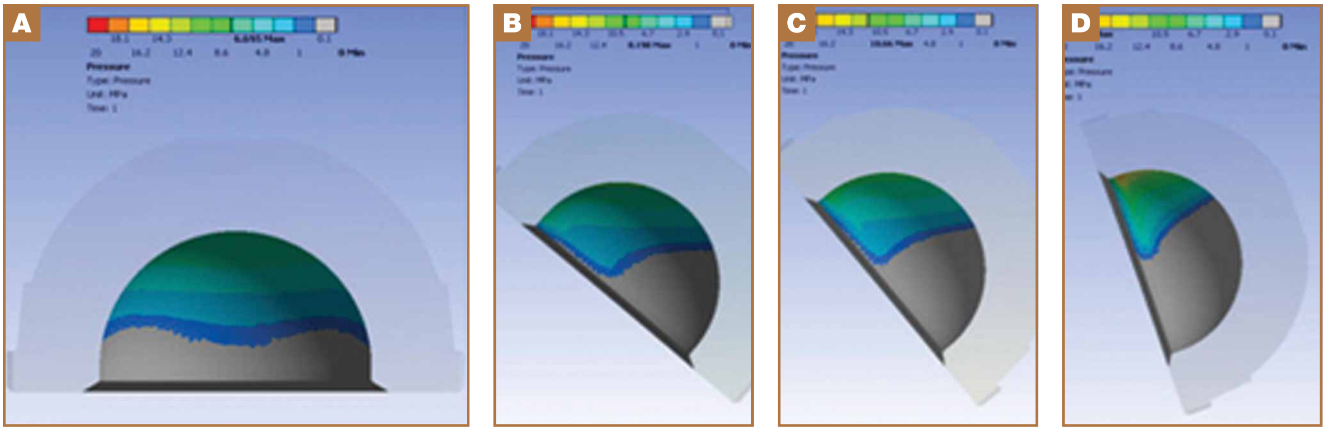

Figure 4. Graphical display of finite element analysis results for contact pressure at each abduction angle studied: (A) 0°, (B) 40°, (C) 50°, (D) 70°. Green shading denotes area of higher contact pressure (megapascals), and blue shading denotes area of lower contact pressure.

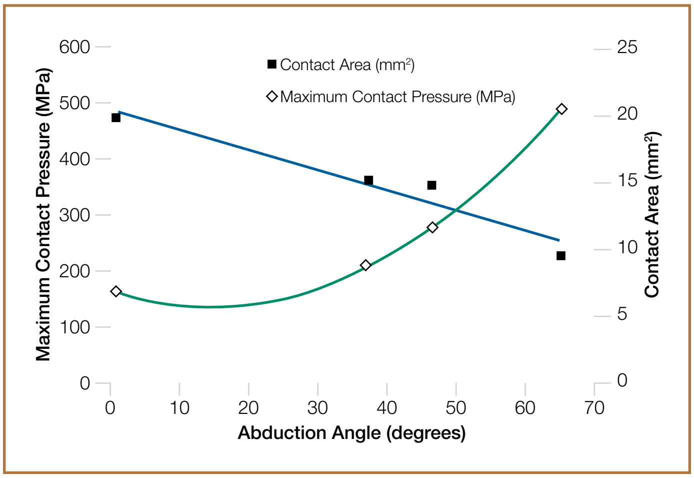

Figure 5. Maximal contact pressure (megapascals; primary axis) and contact area (mm^2) for 4 abduction angles computed using finite element analysis. Correlation coefficient (R^2) of linear regression line for contact area is 0.94; R^2 of second-order polynomial is 0.99.

wear scars and larger areas of polishing on inserts placed at lower abduction angles (0°, 40°) and smaller areas of polishing in cups placed at higher abduction angles (40°, 70°). On visual inspection, there was no evidence of fissuring, cracking, or edge loading on any of the cups.

FEA of maximum contact pressure and contact area demonstrated increasing contact pressure with higher abduction angles with corresponding small contact areas (**Figure 4**). The highest contact pressure was found at 70° abduction (20.4 MPa) and the lowest at 0° abduction (6.8 MPa). Conversely, the smallest contact area was at 70° abduction (227 mm^3), the largest at 0° abduction (474 mm^3) (**Figure 5, Table 2**).

Discussion

Particulate wear debris is a well-known factor that may lead to aseptic loosening or osteolysis and premature implant failure. Results from several studies have suggested that cup abduction angles of more than 40° or 50° may lead to increased polyethylene wear and may subsequently increase the risk for prosthesis loosening.[4,5,7,10,16,21,26] In the present study, we used a controlled biomechanical test to determine the relationship between abduction angle and volumetric wear: Cup abduction was sequentially increased with a constant load while confounding factors (eg, cup migration, differences in cup anteversion) were eliminated. Further, FEA was performed to investigate changes in contact pressure and contact area as a function of abduction angle.

The limitations of this study, commonly observed in in vitro studies, include the difficulty in precisely mimicking the in vivo environment. The lubrication used was bovine-derived, which though not identical to human joint fluid is a close analogue. Another limitation is the theoretical duration of wear with 10 million cycles, which would correspond to about 3.5 years of daily walking by the average active adult (mean age, 74 years).[27] This may have prevented us from observing edge-related damage and wear in the polyethylene. However, unlike in vivo studies, our in vitro study controlled several confounding variables that influence clinical wear: patient activity, patient weight, impingement, cup anteversion, and cup migration.

Table 2. **Finite Element Analysis Results: Maximum Contact Pressure and Contact Area by Abduction Angle**

Abduction Angle, °	Maximum Contact Pressure, MPa	Contact Area, mm^2
0	6.8	474
40	8.7	362
50	11.5	353
70	20.4	227

Our results appear to contradict earlier findings of more wear with larger abduction angles in vivo.[15,17,19,21,28] There may be several explanations for the discrepancy. Many prior studies treated the presence of osteolysis as a surrogate for polyethylene wear but did not actively quantify the wear using radiographic methods, making it difficult to correlate changes in abduction angle to quantitative changes in wear rates. Another explanation may be the presence of dynamic instability and impingement in patients with high abduction angles. It is therefore possible that osteolysis secondary to particulate wear is not caused by debris originating from the femoral head–acetabular cup interface but is instead caused by edge loading secondary to subluxation of the femoral head or by impingement of the neck on the acetabular rim. The results of this in vitro analysis, which specifically controlled for impingement and subluxation, suggest that factors other than cup abduction may lead to excessive wear and subsequent osteolysis. Although we cannot explain the variability in the SDs of the measurements at 0° and 50°, this argues favorably for testing larger numbers of specimens.

Several investigators have reported in vivo results similar to our in vitro results—where cup abduction angle is not correlated with increasing wear and may be associated with other factors.[29,30] Edge wear has been extensively studied in both metal-on-metal and ceramic-on-ceramic articulations but may be just as common in metal-on-polyethylene articulations. In a systematic review of the literature, Harris[31] found that, though edge loading was widespread in polyethylene bearings, linear or volumetric wear rates were not affected by the loading amount. Goosen and colleagues[30] analyzed the wear characteristics of 93 hips at long-term (9-year) follow-up and found no relationship between cup abduction angle and wear rates but did note higher wear rates in patients with osteolysis. It is therefore possible that the implant failures caused by aseptic loosening or osteolysis in some studies with cups placed outside the optimum orientation were due more to the effects of impingement, subluxation, third-body wear, and patient-related factors.[32] Our data suggest that the higher implant failure rate observed in cups implanted at steep angles likely resulted from dynamic factors and not from higher rates of volumetric wear. However, it is important to note that, though we may have found a lower wear rate with high abduction angles, any benefit of lower wear is likely counteracted by increased hip instability, impingement, and edge loading at these cup orientations. Additional well-controlled studies are needed to demonstrate the effect that factors other than creep and adhesive wear have on implant failure rates for acetabular cups placed outside the recommended 40° of abduction.

Although our finding contrasts with the expectation of increased wear with higher cup angle, it agrees with a finding from a basic science study of polyethylene material response to stress.[33] The relationship may be different for highly cross-linked polyethylene. It is important to note that this potentially "protective" effect for wear is likely counteracted by other complications of steep cup angles—including impingement, instability, and edge loading. These factors, which may contribute to device breakage, may be more relevant in explaining why implants fail at a higher rate when cups are oriented at more than 40° of abduction. This type of device integrity failure may have been generally reported as "wear," though true wear appears reduced. Another factor may relate to oxidative degradation of polyethylene—a factor not studied here.

Conclusion

Polyethylene wear after total hip arthroplasty is a major cause of osteolysis and aseptic loosening, which may lead to premature implant failure. Although wear is a multifactorial phenomenon, placement of acetabular cups at steep angles has been noted as a possible cause of early failure. Several investigators have found that high rates of wear at these cup orientations contributed to implant failure. Our data demonstrated that higher cup abduction angles were associated with lower, not higher, wear. This association likely resulted from less head coverage, and therefore a smaller volumetric surface where wear can occur.

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