

4

Hip Joint Tribology

Amir Kamali

According to the ASM International handbook, *tribology* is defined as the science and technology of interacting surfaces in relative motion and all practices related thereto. It includes the study of wear, friction, and lubrication.

Wear and Wear Mechanisms

Wear is defined as the progressive removal of material from contacting surfaces in relative motion. The wear mechanisms in metal-on-metal bearings are as follows:

Adhesive wear occurs by the transfer of material from one surface to another when two surfaces articulate against each other under load. The transferred material could break off and act as third-body particles resulting in abrasive wear.

Abrasive wear occurs when material is removed from a surface by hard asperities on the counterface and hard particles (third body) trapped between the two contact surfaces.

Corrosive wear occurs by the combination of mechanical wear and chemical reaction. Corrosion is the mechanism by which metal ions are released, and as this process is less understood than the other wear mechanisms, more details have been provided in Chapter 5.

It should be pointed out that in metal on metal bearings, all the above-mentioned wear mechanisms could occur simultaneously but at different rates.

Friction and Lubrication

Friction describes the force that opposes motion between articulating surfaces. Lubrication between the bearing surfaces of hip implants and its effect on friction generated during articulation is commonly illustrated by a Stribeck diagram, as shown in Fig. 4.1.

The Stribeck curve is traditionally depicted in three phases. When the thickness of the fluid film is less than or equal to

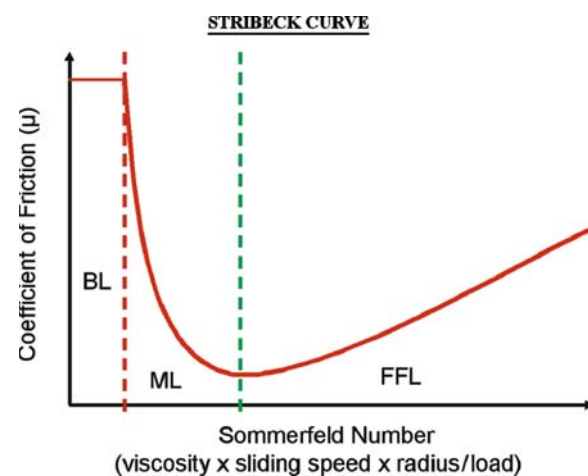


FIG. 4.1. Stribeck curve showing lubrication regimes and their effect on friction.

the average surface roughness of the articulating surfaces, boundary lubrication (BL) is achieved. In this phase, the asperities of the articulating surfaces are in contact at all times. As the thickness of the fluid film increases, the articulating surfaces become separated from each other. There is a transition stage called mixed lubrication (ML), where there is a combination of fluid film and boundary lubrication. As it can be observed in Fig. 4.1, the coefficient of friction continues decreasing until full fluid film lubrication (FFL) is generated, where the articulating surfaces are separated by the lubricant.

Tribology Testing Using Hip Simulators

Tribological testing has been carried out for decades in order to predict the longevity of different designs of hip prostheses. Hip wear simulators have been used extensively by researchers to determine the wear of implants under conditions that are considered to be relatively close to the normal walking cycle. However, *in vitro* hip simulator studies have consistently

reported wear rates that are lower than those reported in *in vivo* studies.

Because of the uninterrupted and identical motions per cycle in hip simulators, the joints will be operating in exaggerated lubrication conditions most of the time, which would protect the bearing surfaces. However, *in vivo*, the extensive range of motion (including stop-start motion), high force, and microseparation between the articulating surfaces would break down the fluid film lubrication in metal on metal bearings resulting in the implant operating in less favorable lubrication regimes and consequently generating higher wear than in *in vitro* samples.

Despite the differences between the *in vitro* wear simulator testing conditions and *in vivo* conditions of the metal on metal implants, the morphologic analysis of the components after wear simulation has shown comparable wear morphology/mechanisms to the clinically retrieved metal on metal hip implants, as shown in Fig. 4.2.

There are a number of different hip simulators available in the market, as shown in Fig. 4.3.

One of the most popular hip simulators that is currently used to predict the long-term performance of hip implants is the MTS machine (orbital type or biaxial rocking motion). In this hip joint simulator, the femoral head is mounted at an angle of 23 degrees to the horizontal, resulting in ± 23 degrees of flexion/extension and ± 23 degrees of abduction/adduction of the implant. The average sliding distance in an orbital-type hip simulator under the above-mentioned conditions has been reported by Wang et al. [1] to be $1.045D$, where D is the diameter of the femoral head. Hence, the sliding distance for a 50-mm femoral head per cycle translates to 52.3 mm. As the tests are normally carried out at 1 Hz, the sliding speed per cycles is going to be 52.3 mm/s. However, the average sliding distance in a natural hip joint on average is $0.67D$, which for a 50-mm femoral head, the average sliding distance per cycle and sliding speed translates to approximately 33.5 mm

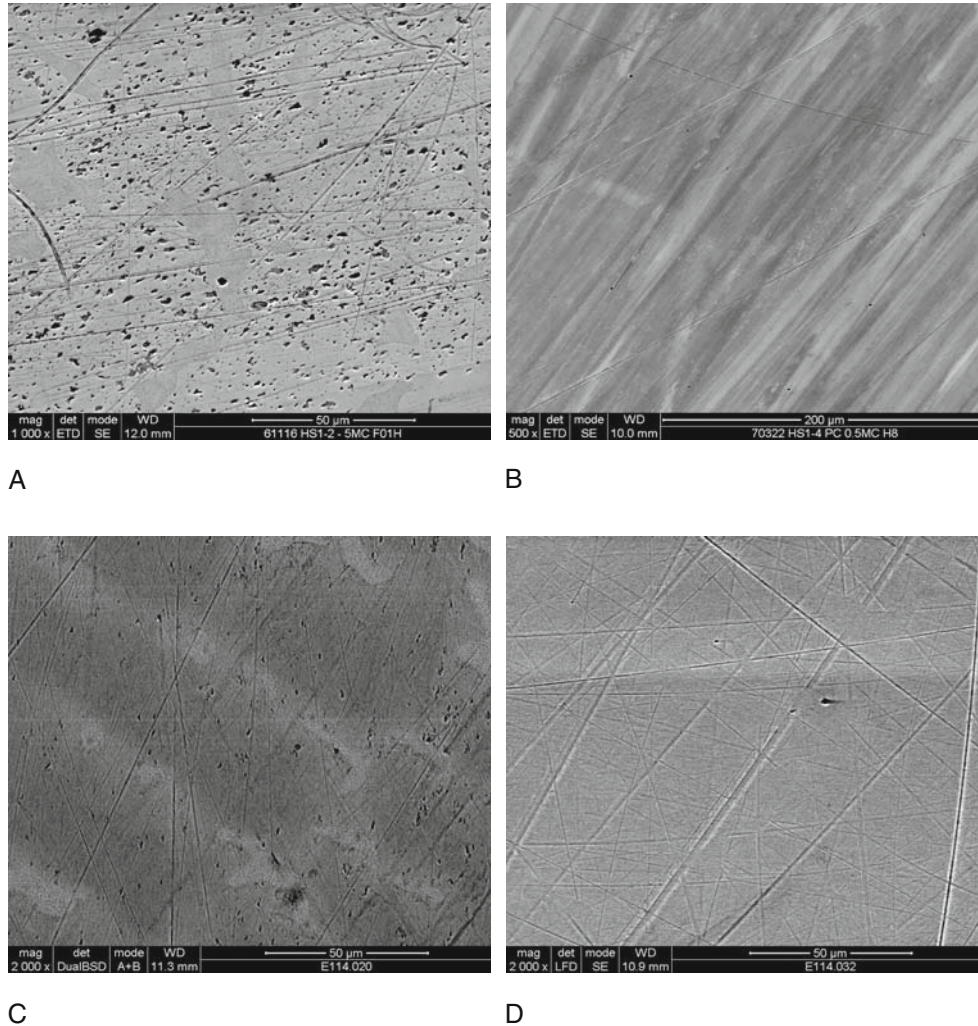


FIG. 4.2. Wear morphology/mechanisms of hip simulator tested and retrieved implants. (A) Hip simulator, corrosion wear. (B) Hip simulator, abrasive wear. (C) Retrieved implant, corrosion wear. (D) Retrieved implant, abrasive wear.

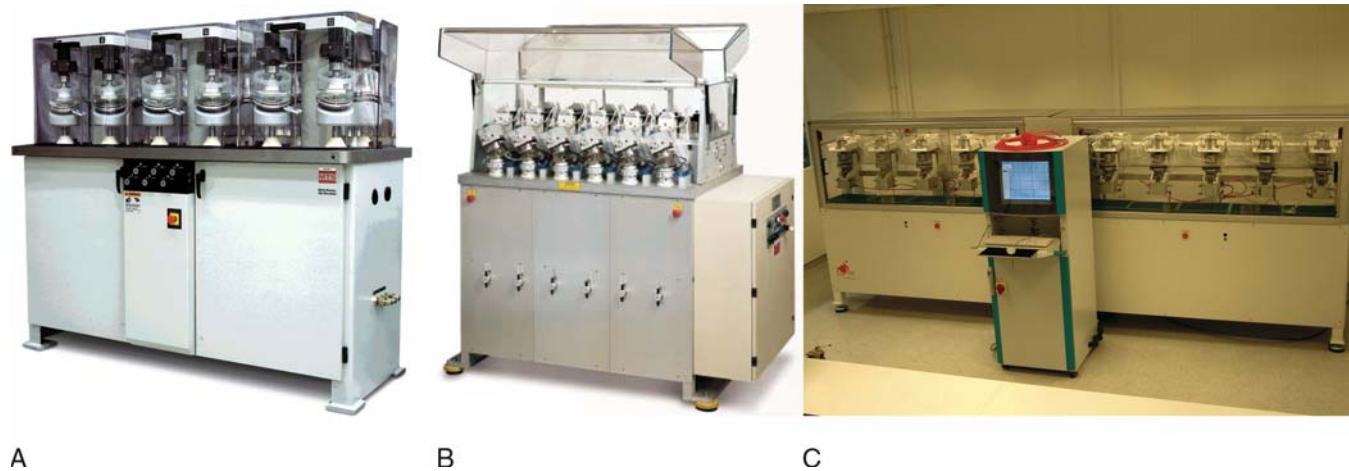


FIG. 4.3. A selection of the hip simulators available on the market: (A) MTS, (B) AMTI, (C) SimSol.

and 33.5 mm/s, respectively. Hence, the orbital-type hip simulator would produce a 56% increase in the sliding distance per cycle in comparison with that generated by the natural hip joint. The significant increase in sliding distance and sliding speed would improve the lubrication and consequently reduce the contact between the articulating surfaces.

A number of researchers have reported no significant difference between the wear generated by various CoCr alloy microstructures in hip joint simulator studies. However, as mentioned previously, the excessive fluid film lubrication generated between the articulating surfaces of metal on metal bearings in hip simulators would reduce the effect of material microstructure on wear of the implants, as the contact between the components is artificially minimized. The effect of material microstructure on wear is clear when the lubricant in the test is not separating the two articulating surfaces. This can be achieved by using pin-on-plate or pin-on-disk machines. These machines are used for material screening purposes, and it has been shown in a number of studies that as-cast high-carbon CoCr alloy microstructure is superior in terms of wear resistance to other CoCr alloy microstructures. In one of the more recent pin-on-disc studies, Kinbrum et al. [2] demonstrated that the microstructure and in particular the carbide volume fraction present in the material is critical to the tribological performance of metal on metal devices. The authors had a series of as-cast (high carbide), single heat treated (medium carbide), and double heat treated (low carbide) CoCr alloy pins and disks. The single and double heat treatments had been carried out in order to reduce the carbide volume fraction in the material. The results are presented in Fig. 4.4.

It can clearly be observed in Fig. 4.4 that the as-cast material has greater wear resistance when compared with single and double heat treated materials.

It should be pointed out that it is not advisable to compare wear results generated at different laboratories and/or hip simulators as factors such as the kinetics, kinematics, synchronization

between the load and motion, the fluid test medium and its degradation with time may differ greatly from one center to another, all of which would affect the wear results. Also, measurement techniques (i.e., gravimetric or volumetric wear measurements) are other factors that may influence the wear results.

It is also important to mention that one of the most significant factors affecting the wear of implants *in vitro* is the correct test setup and component alignment. In a series of hip simulator studies by Dowson et al. [3] and Isaac et al. [4], the researchers showed that low-clearance (83–129 μm) components generate significantly lower wear than do the high-clearance (254–307 μm) components. A series of pictures of the hip simulator stations has been published, showing that within the first 150,000 cycles of the test, the lubricant (new-born calf serum) for the high-clearance joints had gone black due to the high wear of the components. It was also reported that there were no significant changes in the color of the serum used in the stations with low-clearance joints.

Similar hip simulator studies have been carried out at Smith & Nephew's Implant Development Centre (IDC) on 50-mm

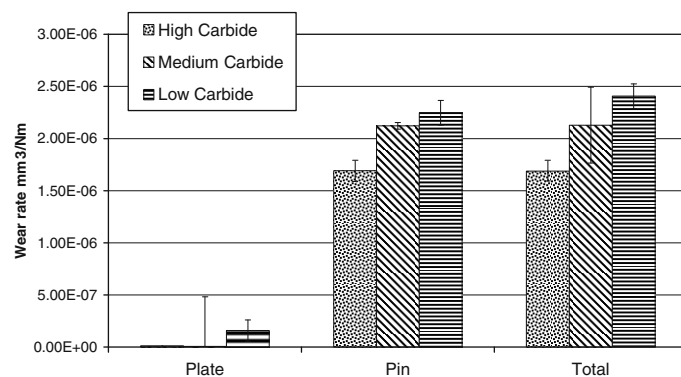
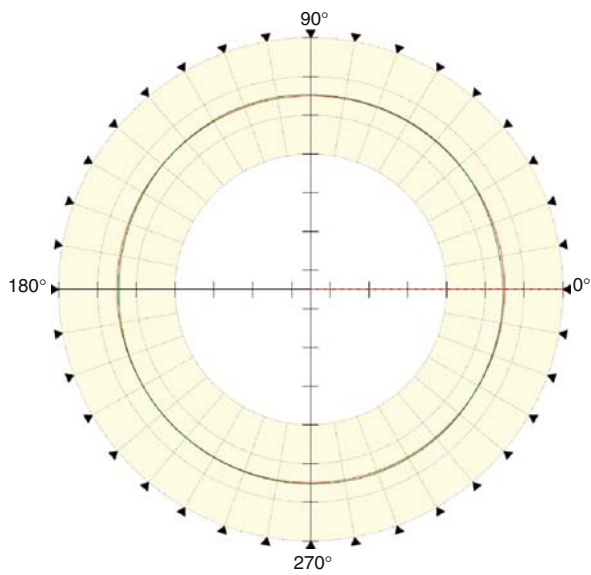
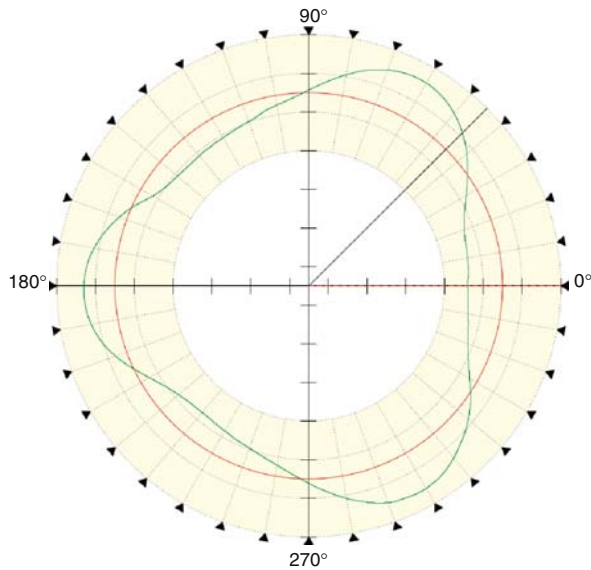


FIG. 4.4. Comparison of the CoCr alloy wear with varying carbide volume fraction.



A



B

FIG. 4.5. Equatorial roundness measurements shown (A) before and (B) after the removal of fixation surfaces.

Birmingham Hip Resurfacing (BHR) devices with approximately 250- μm clearances without observing any significant discoloration of the fluid lubricant. It should be pointed out that irrespective of the clearance, minor discoloration will occur in lubricant during the early stages of wear tests due to the generation of more wear debris (running-in phase) compared with the latter stages of the test (steady-state phase) and also due to the degradation of the fluid lubricant (newborn calf serum).

Further studies were carried out by the IDC to find out how the researchers managed to get such profound metal staining from the BHR. In close examination of the study by Dowson

et al. [3], the authors stated that “the acetabular components were modified to remove fixation surfaces that would have hindered component location and measurements.” Although this practice may have made it easier for the researchers to mount the components in their hip simulators, unfortunately, as any modification to the fixation surfaces, it would have also deformed the cups during the removal of the fixation surfaces. Also, this process could release the residual stresses within the material, which in turn would deform the cups even further. From a manufacturing point of view, this type of material removal should never be carried out without consideration of the final bearing geometry. In order to demonstrate this phe-



A



B

FIG 4.6. Photographs of a cup (A) before and (B) after the removal of its fixation surfaces.

nomenon, the IDC repeated this type of maneuver and then carried out a similar hip simulator study. In this study, five 50-mm BHR devices with 240- to 250- μm clearances were tested. The fixation surfaces of three cups were removed, and the other two cups were tested without the removal of their fixation surfaces. A series of roundness measurements was carried out before and after the removal of the fixation surfaces of the cups to determine the amount of cup deformation (Fig. 4.5).

It is clear from the above measurements that the cups are significantly deformed (compressed by approximately 180 μm from their original shapes) due to the fixation removal process. This amount of artificial deformation would have a detrimental

effect on the implant wear. Although cup deformation can clearly be observed using a roundness machine, it is extremely difficult to detect this with the naked eye, as shown in Fig. 4.6.

After the removal of the fixation surfaces, the implants were tested in a ProSim multi-axis hip wear simulator. The simulator was stopped every 10,000 cycles and photographs taken of each station. This procedure was repeated, and pictures of the stations were taken until 150,000 cycles, as shown in Fig. 4.7. The lubricant was then changed at 150,000 cycles, and the test was continued until 300,000 cycles. The machine was stopped every 50,000 cycles for photographs to be taken of each station, as shown in Fig. 4.8.

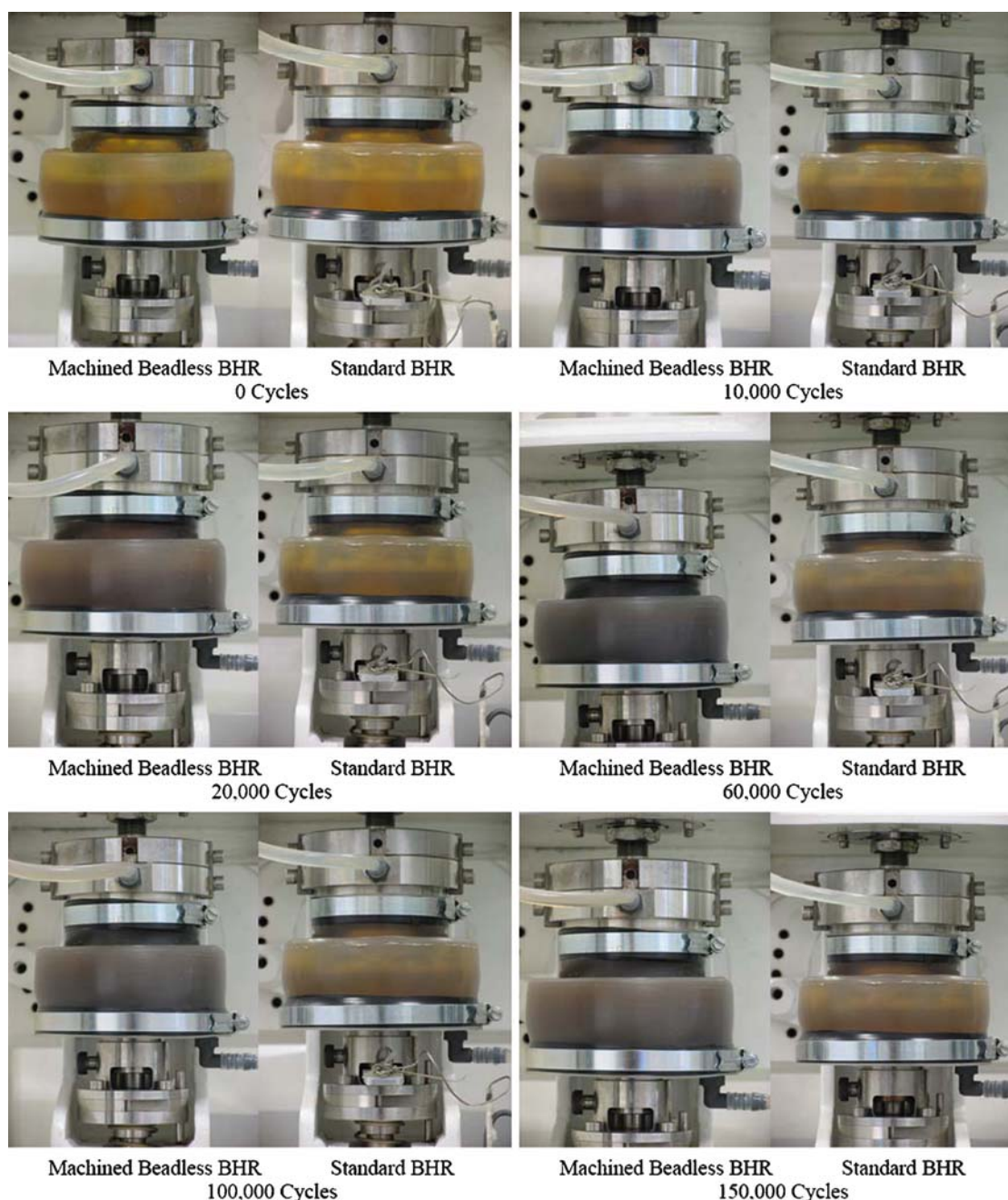


FIG. 4.7. Photographs of representative stations up to 150,000 cycles of wear testing before serum change (left, BHR cup with removed fixation surfaces; right, standard BHR cup).

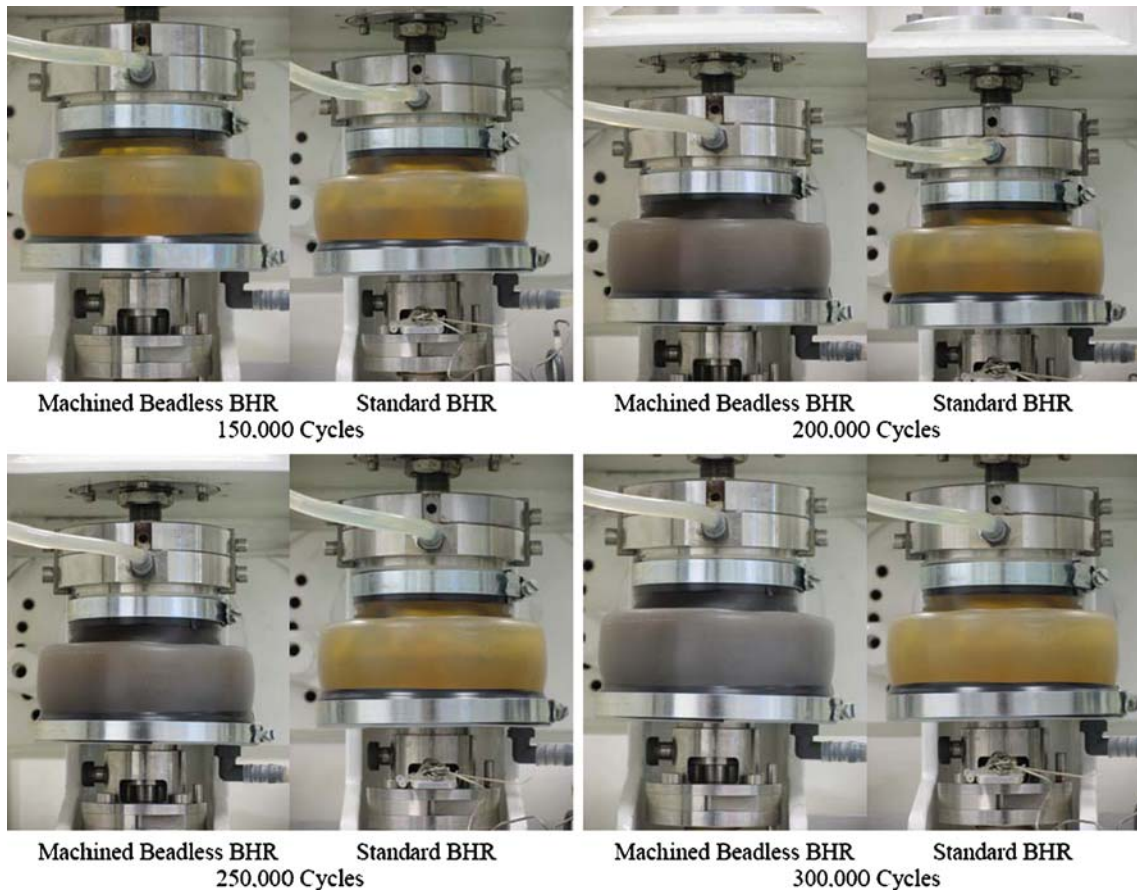


FIG. 4.8. Photographs of representative stations up to 300,000 cycles of wear testing before serum change (left, BHR cup with removed fixation surfaces; right, standard BHR cup).

The color of the serum in the stations with the BHR that had the fixation surfaces removed from the backs of the cups showed similar discolorations to the ones published by Dowson et al. [3] and Isaac et al. [4]. This study has clearly demonstrated the importance of test setup in hip simulator studies and its effect on implant wear. It should be pointed out that test protocols should always consider the final representative features of a product.

It is clear from Fig. 4.8 photographs that the fluid lubricant in the station with the standard BHR device has not visibly changed color during the second 150,000 cycles of the test. However, the station with the BHR with removed fixation surfaces continues to show darkening of the fluid lubricant.

The Effect of Diameter on the Tribology of a Metal on Metal Implant

The effect of metal on metal hip implant diameter has been examined, and the results are as follows. The wear equation states that:

$$V = K \times L \times x, \quad (4.1)$$

where V is the volume of wear (mm^3), K is the wear factor ($\text{mm}^3/\text{N}\cdot\text{m}$), L is load (N), and x is the sliding distance (m) covered during the test. Thus wear increases as any of these parameters increases and vice versa. K , the wear factor, is related to the probability of producing a wear particle, so under different conditions of surface cleanliness or the chemical nature of the surfaces, K will vary and so will the wear rate. However, for a given load and a given surface condition, the wear volume is directly proportional to the sliding distance, which in turn is directly proportional to the radius of the joint. Thus greater wear volumes arise from larger heads. Hence, a large head (e.g., 50 mm diameter) would wear more than a small head (e.g., 38 mm diameter) provided that K and L did not vary.

However, if we use larger head diameters in the presence of a lubricant, then there is an increased chance of fluid-film lubrication and thus a reduced probability of producing a wear particle (as K reduces) because the surfaces do not make contact except occasionally. Thus from a fluid-film lubrication point of view, the smaller head represents the worst-case scenario in terms of wear as K would be largest for the smaller head.

The equation governing the film thickness is as follows:

$$\frac{h_{\min}}{R_x} = 2.798 \left[\frac{\eta u}{E' R_x} \right]^{0.65} \left[\frac{L}{E' R_x^2} \right]^{-0.21} \quad (4.2)$$

E' and R_x are defined as below:

$$R_x = \frac{R_1 R_2}{R_1 - R_2} \quad (4.3)$$

In equation (4.3), R_1 and R_2 is the radius of the cup and the head respectively.

$$\frac{1}{E'} = \frac{1}{2} \left[\frac{1 - \sigma_1^2}{E_1} + \frac{1 - \sigma_2^2}{E_2} \right] \quad (4.4)$$

In equation (4.4), E_1 and E_2 are the Young's modulus of the cup and the head, respectively, and σ_1 and σ_2 are the Poisson's ratio of the cup and the head, respectively.

It is generally considered that fluid film lubrication occurs when the fluid-film thickness is three times larger than the combined surface roughness of the articulating surfaces. Theoretical calculations using equation (4.5) are used to determine the lubrication regimes generated between the bearing surfaces:

$$\lambda (\text{Lambda Value}) = \frac{h_{\min}}{\text{Surface roughness of the articulating surfaces}} \quad (4.5)$$

$1 > \lambda$	Boundary lubrication
$1 < \lambda < 3$	Mixed lubrication
$3 < \lambda$	Fluid film lubrication

However, it is not as simple as this. Surfaces that have the same surface roughness can have peaks and valleys that are very differently distributed. Surfaces with a positive skewness in the distribution of surface asperities are less easy to lubricate with fluid film lubrication as the peaks of the surface roughness can penetrate the fluid film more readily. Surfaces with a negative skewness have more valleys than peaks and are easier to lubricate with fluid film mechanisms. This is explained in more detail in the "Superfinishing" section in Chapter 3.

Another important factor in producing fluid-film lubrication is the clearance between the ball and socket of the resurfacing device. The effects of this can be seen in equation (4.3) where R_x is dependent on the radial clearance, $R_1 - R_2$. If $R_1 - R_2$ is very small, then the fluid film thickness becomes greater and K should reduce. However if the clearance ($R_1 - R_2$) becomes too small, then the risk of the two surfaces "clamping" through cup deformation increases.

Theoretical Calculations

In order to assess what might be happening in the different sizes and clearance of components, calculations have been performed on a range of products.

38-mm-Diameter BHR Head

If we calculate the film thickness using a typical synovial fluid viscosity of 0.01 Pa·s, an entraining velocity of 0.02 m/s, $E\phi = 2.3 \times 10^{11}$, and relative radius of curvatures based on

implanted clearances of 130 μm and 260 μm , then the range of film thicknesses is from 0.085 μm to 0.05 μm giving λ values of between 2.4 and 1.4 (for a combined surface roughness of 0.035 μm).

Thus, the theoretical predictions are that for a 38-mm femoral head with the smallest specified diametral clearance, the joint will be close to fluid film lubrication ($\lambda = 2.4$ rather than 3 for full fluid film), but at the higher clearance, more asperity contact would be expected as the λ value is calculated at only 1.4.

50-mm-Diameter BHR Head

Again the viscosity was chosen as 0.01 Pa·s $E\phi = 2.3 \times 10^{11}$, the entraining velocity in this case is 0.026 m/s, and hence for an implanted clearance of 190 μm , the film thickness is 0.115 μm and for an implanted clearance of 320 μm the film thickness is 0.077 μm . Hence the λ values again vary from 3.3 to 2.2 indicating that the joints operate about the fluid level, but with some asperity contact depending on the clearance.

Small-diameter heads (in this case 38 mm) do not produce a sufficiently thick film of lubricant to separate the surfaces. Thus this would be the worst-case scenario for reducing K in the wear equation. However, as we know that some asperities penetrate fluid film and cause metal on metal wear, then for smaller heads, the sliding distance (x) is the shortest, consequently the wear volume will also be small. Thus the small head is the best-case situation for metal on metal direct wear. In the large-diameter heads (in this case 50 mm), the opposite of this would be true. The film thicknesses would be greater, thus reducing K , as x would be greater.

On balance, it would be expected that the wear rates would be similar at all diametral sizes because of these two competing factors (K and x).

Experimental Work

Hip simulator studies have been carried out at Durham University investigating the effects of head sizes on the wear of the implants. BHR 38-mm and 50-mm devices were tested in the Durham hip function wear simulator I for 5 million cycles each. These studies showed no statistically significant differences between the wear rates generated by the 38- and 50-mm BHR devices (1.32 and 1.08 mm³/million cycles, respectively). Bowsher et al. [5] have also demonstrated in a hip simulator study that running-in wear did not correlate with joint diameter.

The experimental results are in agreement with the theoretical calculations. When a larger-diameter (50 mm) joint has been used, the surface will have traveled faster than that of a smaller-diameter (38 mm) joint causing a thicker fluid film to develop. Fluid film lubrication will prevent asperities touching and therefore little wear will be accumulated.

With the smaller-diameter joint (38 mm), a thinner fluid film will be generated, as the surface will move more slowly, however there will be less wear due to the shorter sliding distance of the smaller-diameter head. The above theoretical calculations and the experimental studies have shown that the effect of head sizes in wear of implants is insignificant.

The Effect of Clearance on the Tribology of a Metal on Metal Implant

In vitro friction and hip simulator studies continue to be conducted to determine the optimum clearance for a given bearing diameter. There has been a consistent trend in these studies showing that low clearance in a bearing improves the lubrication between the articulating surfaces and consequently reduces the friction and wear generated between the bearings. However, if we consider friction hip simulator studies, most of them have employed bovine serum (BS) or bovine serum with added carboxy methyl cellulose (CMC) as the lubricant, as this combination is believed to simulate the viscosity of synovial fluid. In real life, as soon as the joint is implanted, the joint is actually bathed in blood and not even synovial fluid. Blood contains macromolecules and cells that measure 5 to 20 μm or more. The effect of these on friction is not fully understood.

Also, none of the previous friction studies have taken cup deformation, which occurs during cup implantation and may also occur during physiologic loading *in vivo*, into consideration. It should be pointed out that intraoperative measurements of the BHR devices have shown up to more than 100 μm of cup deflection immediately after implantation. Cementless cup designs in metal on metal hip resurfacing devices generally depend on a good primary press-fit fixation, which stabilizes the components in the early postoperative period. This allows bony ingrowth or ongrowth to occur, which in turn provides durable long-term fixation. However, press-fitting the cup into the acetabulum generates nonuniform compressive stresses on the cup and consequently causes nonuniform cup deformation. That in turn may result in equatorial contact, high frictional torque, and femoral head seizure. Increased bearing friction in the early weeks and months after implantation can lead to micromotion and has the potential to prevent effective bony ingrowth from occurring. Therefore, friction in the early postoperative period can be critical to the long-term success of joint fixation.

Concerns were raised by McMinn et al. [6] in a clinicoradiologic study of metal on metal bearings with closely controlled 100- μm clearance. A progressive radiolucent line indicated by the arrows in Fig. 4.9 around the acetabular component, seen in some of these cases at follow-up, raises the possibility that increased friction is affecting component fixation. It should

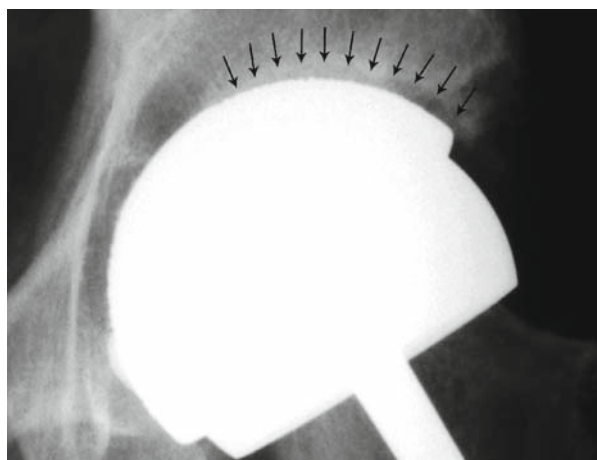


FIG 4.9. A 2-year postoperative radiograph of a patient with a low-clearance BHR, showing a progressive radiolucent line around zones 1 and 2 of the acetabular component (arrows), suggesting increased friction and micromotion resulting in poor fixation.

be pointed out that this phenomenon has not been observed in devices with regular (higher) clearances.

In order to identify the optimum clearance for a given bearing diameter and to understand the above-mentioned phenomenon, a series of friction simulator tests under physiologically relevant conditions was carried out by the author.

Initially, six BHR devices with various diametral clearances (80 to 306 μm) were tested in a hip friction simulator to determine the friction between the bearing surfaces. The components were tested in whole and clotted blood (viscosity 0.0083 and 0.0108 Pa·s, respectively), which are the primary lubricants during the early postoperative period and also in BS + CMC and BS + CMC + hyaluronic acid (HA) with viscosity of 0.01 Pa·s. The results are presented in Fig. 4.10.

When serum-based lubricants are used, it can be observed that as the clearance increases, the friction between the articulating surfaces also increases. A slight increase in friction was noted when HA was added to the serum, as shown in Fig. 4.10, which may have been generated due to the shearing of the HA molecules. Statistical analysis showed that this difference was not significant ($p > 0.05$).

However, when physiologically relevant lubricants such as whole and clotted blood are used, the friction between the bearings with low clearances (80 and 135 μm) is significantly increased ($p < 0.05$) in comparison with those generated in serum-based lubricants. It can also be observed that as the clearance is increased, the friction is reduced, following the opposite trend to that of the serum-based lubricants.

The components were then deflected by 25 to 35 μm using a two-points pinching action and tested in clotted blood, which is the primary lubricant during the early postoperative period. The results for this test are presented in Fig. 4.11.

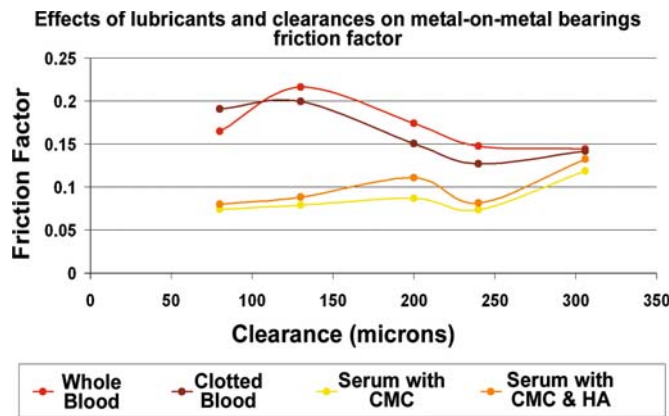


FIG. 4.10. Effect of clearance and lubricant on friction factor.

The results presented in Fig. 4.11 show that for the reduced-clearance components, friction was significantly increased ($p < 0.05$) when the cups were deflected by only $30\mu\text{m}$. However, for the components with higher clearances, the friction did not change before and after deflection. It is postulated that the larger clearances can accommodate the amount of distortion introduced to the cups in this study.

It has been reported that reduced clearance results in reduced friction. However, factors such as cellular and macromolecular shear and cup deflection that can affect friction in these bearings *in vivo* have not been specifically investigated *in vitro* before. Progressive radiolucent lines that appeared in a few patients with low-clearance bearings alerted us to the need to study this issue of increased friction in these bearings.

The results of this study suggest that reduced clearance bearings have the potential to generate high friction especially in the early weeks after implantation when cup deflection has occurred due to press-fitting of the cup into the acetabulum and the presence of whole and clotted blood as lubricant. Friction factors in higher clearance bearings

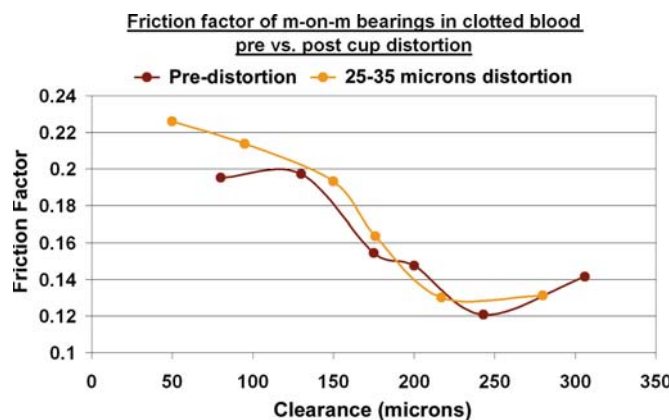


FIG. 4.11. Effect of cup deflection on friction in clotted blood.

are much reduced in comparison. The increase in friction results in increased frictional torque at the implant-bone interface, which could in turn cause cup loosening and increased wear.

The Effect of Cup Orientation on the Tribology of a Metal on Metal Implant

There is increasing evidence from retrieval studies that cup positioning, particularly inclination angle, has a significant impact on the wear of metal on metal bearings [7,8]. The effect is identified by *edge wear* on the peripheral edge of the cup component. Also, correct orientation of the implant is essential for maximizing its range of motion as well as preventing impingement and dislocation [9,10].

A hip simulator study was carried out by the IDC to compare the effect of cup orientation on the wear performance of BHR devices.

The wear test was performed in a 10 station ProSim hip joint simulator. A series of 50 mm BHR devices was tested in this study. The bearings were divided into three groups ($n = 3/\text{group}$) with various cup orientations and one control sample. Thereby, the distance between the wear patch of the bearings and the superior edge of the cup was different for each group. This would then allow investigation of changes to the contact and lubrication between the articulating surfaces and consequently wear of the implants. All the implants were mounted in an anatomical position. The cup orientations are as follows (Fig. 4.12):

Group A: The edge of the wear patch at maximum flexion is 19 degrees away from the edge of the cup ($n = 3$).

Group B: The edge of the wear patch at maximum flexion is 8 degrees away from the edge of the cup ($n = 3$).

Group C: The edge of the wear patch at maximum flexion is 5 degrees away from the edge of the cup ($n = 3$).

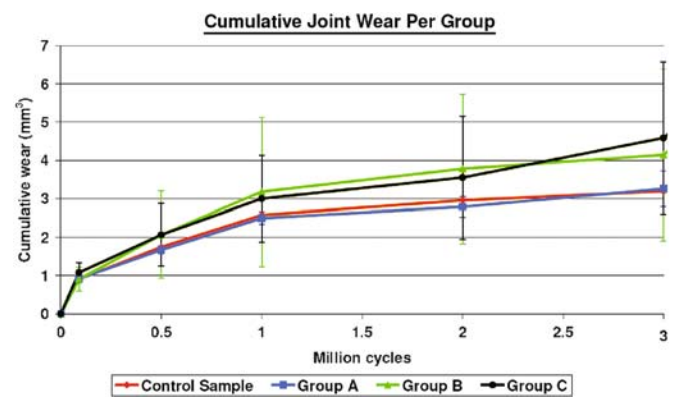


FIG. 4.12. Average cumulative wear volume loss of BHR bearings with varying cup orientations and slow walking speeds.

Control sample: The edge of the wear patch at maximum flexion is 38 degrees away from the edge of the cup ($n = 1$).

The lubricant in this study was newborn calf serum with 0.2% sodium azide concentration, which was diluted with de-ionized water to achieve average protein concentration of 20 g/L. The flexion/extension and internal/external rotation of the implants were +30 degrees/-15 degrees and ± 10 degrees, respectively. Paul-type stance phase loading with a maximum load of 3000 N and a standard International Standards Organization (ISO) swing phase load of 300 N were applied to the implants. The frequency was 1 Hz.

The average cumulative wear volumes for the three BHR groups at various orientations plus a control sample are presented in Fig. 4.12.

As it can be observed, the devices showed the typical characteristics of wear for metal on metal joints, with a high wear rate during the initial running in period (0 to 1 Mc) followed by a lower steady-state wear rate between 1 and 3 Mc.

At 2 million cycles, one implant in group C (implant no. 9) exhibited extremely high wear. This was caused by neck (head holder) impingement, resulting in head articulation against the edge of the cup. It is interesting to note that edge articulation caused a 60-fold increase in wear generated between 2 and 3 Mc (8.74 mm^3) in comparison with that generated between 1 and 2 Mc (0.15 mm^3) for implant no. 9, as shown in Fig. 4.13. The wear result of implant no. 9 was excluded from this study after 2 million cycles of testing.

The joints showed no significant difference between the groups during their running-in period, nor was there any significant difference during the steady-state period of the test between the groups ($p > 0.05$). Hence, when articulation occurred within the bearing surfaces of the implants, no significant differences in wear rates were observed between the groups implanted at various alignments.

The surface measurements in this study showed an increase in average surface roughness of the heads and cups combined

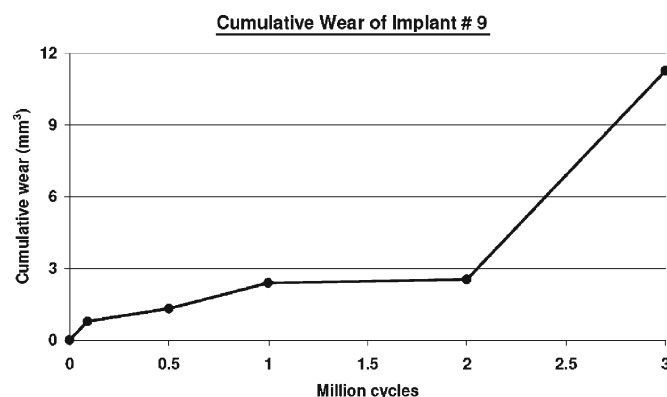


FIG 4.13. Cumulative wear volume loss of implant no. 9 in group C.

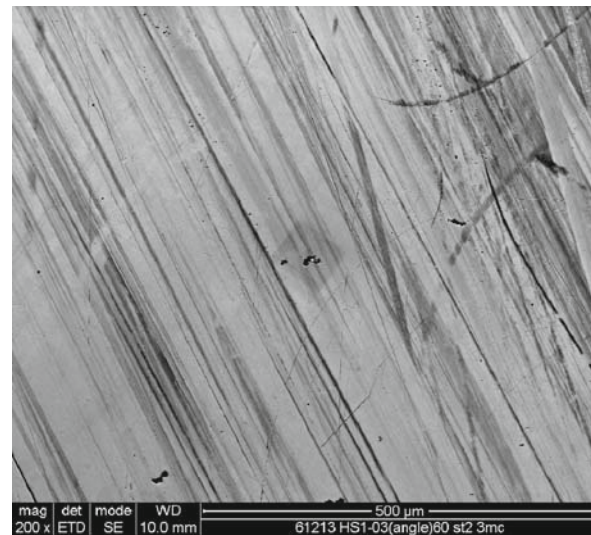


FIG 4.14. SEM image of typical abrasive wear characteristics on the articulating surfaces.

with the significant reduction in the average skewness of the test samples, clearly indicating the diminishing of peaks and increasing valleys on the surface of both cups and heads during the initial 90,000 cycles of the test (approximately 1 month *in vivo*). This is due to the abrasive wear mechanism that occurs between the articulating surfaces, as shown in Fig. 4.14.

In conclusion, no significant differences were observed between the wear rates of the groups when articulation occurred within the bearing surfaces of the implants. However, a significant increase in wear rate was measured in one of the joints when articulation occurred on the edge of the cup. These findings suggest that the wear rate will not be affected by cup orientation as long as the articulation occurs within the bearing surfaces of the implant. However, edge loading/articulation will increase the wear of the implant significantly, which in turn may result in osteolysis and/or aseptic loosening of metal-on-metal bearings.

Summary

Tribological testing using hip wear and friction simulators continues to be carried out in order to investigate the performance and to predict the longevity of different designs of hip prosthesis. Despite their limitations, these machines are efficient tools for basic research and are essential in improving our understanding of the wear, friction, and lubrication generated by the metal on metal bearings. However, it should be pointed out that, in order to generate meaningful data, it is crucial that the test setup in hip simulator studies is correct and that the final representative features of the products have been considered in the test protocol.

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