Wear of the polyethylene acetabular cup

Metallic and ceramic heads compared in a hip simulator

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Ultra-high molecular weight polyethylene acetabular cups of 5 different total hip systems (Müller, Mallory-Head, Lubinus, P.C.A. and Charnley-Elite) were worn on a new 5-station hip joint simulator. The cups articulated against modular metallic (stainless steel in Müller and Charnley-Elite, ion-implanted Ti-6Al-4V in Mallory-Head, and Co-Cr-Mo in Lubinus and P.C.A.) and modular alumina ceramic femoral heads for 3 million walking cycles. The mean wear rate of cups against alumina heads (range 0–5.7 mg/10^6 cycles, corresponding to 0–0.008 mm/year) was usually lower than against metallic heads (range 3.9–178 mg/10^6 cycles, corresponding to 0.005–0.24 mm/year). In the metal-head prostheses, the mean wear rate was highest against stainless steel heads, and lowest against ion-implanted Ti-6Al-4V heads.

As the wear rates are compared with published clinical observations, it can be concluded that the hip joint simulator is capable of producing realistic wear rates; it is a useful instrument in the study of the wear behavior of new designs, materials, surface treatments and coatings prior to clinical trials. However, the taper-fit attachment of modular heads proved problematical, showing corrosion and wear at the conical head-spigot interface.

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Osteolysis and loosening of a total hip endoprosthesis are biologic responses to wear and corrosion products of the implant, mainly ultra-high molecular weight polyethylene (UHMWPE) wear debris. Thus friction and wear at the femoral head—acetabular cup articulation may be major factors determining the overall performance of the prosthesis (Willert and Semlitsch 1977, Revell et al. 1978, Santavirta et al. 1990, Amstutz et al. 1992). New designs, materials, surface treatments and coatings are frequently introduced. A hip joint simulator is an instrument that can be used in the evaluation of the friction and wear behavior in vitro, preferably prior to clinical trials. A review of laboratory wear studies of biomaterials was made by Clarke (1982). During the past decade hip joint simulator studies seem to have been published less and less frequently, although the complex of wear problems is still far from solved and more efforts are definitely needed. We started our cooperation in the study of friction and wear of total hip joints in 1987. The new 5-station hip joint simulator became operational in April 1991 and has been running almost continuously ever since, effectively producing wear data. It has 5 identical test stations, and the conditions of the prostheses simulate walking (Saikko et al. 1992). The main features are in accord with ISO TR 9325 (1989). Wear of polyethylene cups articulating against metallic vs. ceramic femoral heads was considered to need more elucidation, and the study of the wear behavior of different joints in identical conditions was expected to give useful information. In addition, reference data are necessary for future work with experimental materials.

Abbreviations used in text

δM_t	Wear of an acetabular cup after a given number of simulated walking cycles, expressed as loss of mass $= (M_{t0} - M_t - M_{s0} + M_s)$ (mg)
M_{t0}	Initial mass of the test acetabular cup (mg)
M_t	Mass of the test acetabular cup (mg)
M_{s0}	Initial mass of the soak control cup (mg)
M_s	Mass of the soak control cup (mg)
k	Wear factor $= V / (N \int L dx)$ ($mm^3 N^{-1} m^{-1}$)
V	Volume of material removed (mm^3)
N	Number of walking cycles
L	Load applied to the hip joint (N)
x	Relative movement between the head and the cup (m)
r_H	Radius of the femoral head (mm)
r_C	Internal radius of the acetabular cup (mm)
R_a	Surface roughness; the arithmetic mean of the departures of the profile from the mean line (µm)

Table 1. Femoral heads

Manufacturer or vendor	Description on package	Cat. No.	Material	Material standard	Nominal diameter (mm)
Protek	Müller modular head	35.32.01	Stainless steel 316L Protema-42	ISO 5832/1, ASTM F 138	32
	Müller modular head	12.32.06	Alumina ceramic BIOLOX	ISO 6474, ASTM F 603	32
Biomet	Modular head component	163185	Ti-6Al-4V, ion-implanted	ISO 5832/3, ASTM F 136	32
	Modular head component	131407	Alumina ceramic	ISO 6474	32
Waldemar Link	SP II - Prothesenkopf	128-705	Co-Cr-Mo	ISO 5832/4	32
	SP II - Prothesenkopf	128-709	Alumina ceramic BIOLOX	ISO 6474, DIN 58 835	32
Howmedica	Femoral head component	6284-0-132	Co-Cr-Mo Vitallium	ISO 5832/4, ASTM F 75	32
	Ceramic head	6290-1-032	Alumina ceramic	ISO 6474	32
Thackray	Elite modular head	62-5671	Stainless steel ORTRON 90	ISO 5832/9, BS 7252/9	22.25
	Charnley ceramic head	62-6392	Alumina ceramic	ISO 6474, BS 7253/2	22.25

Table 2. Acetabular cups

Manufacturer or vendor	Description on package	Cat. No.	UHMWPE material standard	Nominal OD (mm) (MB metal-backed)
Protek	Full profile cup	62.32.50	ISO 5834/1+2, ASTM F 648	50
Biomet	Modular acetabular liner	104308	ISO 5834/1, ASTM F 648	56 (MB)[1]
Waldemar Link	Lubinus	102-110	ISO 5834/2, DIN 58 834	52
Howmedica	P.C.A. acetabular insert	6285-0-525	ISO 5834/2, BS 7253/4	52 (MB)[2]
Thackray	Charnley	62-3717	ISO 5834/1+2, BS 7253/4+5	43

[1] Biomet's Mallory-Head Ti-6Al-4V acetabular shell, Cat. No. 11-104256
[2] Howmedica's P.C.A. Co-Cr-Mo Vitallium acetabular shell, Cat. No. 6289-5-052

Table 3. Femoral stems

Manufacturer or vendor	Description on package	Cat. No.	Material	Corresponding heads	
				metal	ceramic
Protek	Müller modular straight stem	16.00.20-200	Stainless steel 316L PROTEMA-42	+	
	Müller modular straight stem	22.00.29-200	Ti6Al7Nb PROTASUL 100		+
Biomet	Modular femoral component	163325	Ti-6Al-4V	+	+
Waldemar Link	SP II Modell Lubinus	128-719	Co-Cr-Mo	+	+
Howmedica	P.C.A. primary femoral stem	6280-7-010	Co-Cr-Mo Vitallium	+	+
Thackray	Elite femoral prosthesis	62-5507	Stainless steel ORTRON 90	+	
	Charnley femoral hip prosthesis	62-6392	Stainless steel ORTRON 90		+

Material and methods

Various commercially available modular prostheses, representing cemented and cementless designs common in Finland (Paavolainen et al. 1991), were selected for the study (Tables 1 and 2). Some femoral heads, however, were included because they were considered otherwise interesting. The heads were attached to the cradle of the simulator by means of femoral stems (Table 3). The information was obtained from the packages, manufacturers and vendors. All heads and cups were new; they had been manufactured for implantation and were received in their original packages. All polyethylene cups had been sterilized by gamma irradiation. The heads and the load-bearing areas of the cups were not modified in any way prior to the tests. The prostheses were manufactured or supplied by Protek AG (Switzerland), Biomet, Inc. (U.S.A.), Waldemar Link GmbH & Co (Germany), Howmedica, Inc. (U.S.A.), and Chas. F. Thackray Ltd (England). Müller, Lubinus and Charnley-Elite are used with cement and Mallory-Head and P.C.A. without. The attachment of the heads was taper-fit, with exception of the Charnley alumina ceramic head. Its attachment was based on a parallel-sided recess in the head, a parallel-sided trunnion of the stem, and a thin intermediate press-fit sleeve, made of polyethylene. Howmedica's alumina ceramic head was not a part of the P.C.A. system, but the accommodation was possible by means of an adapter sleeve made of Vitallium alloy, Cat. No. MU40-0-016.

The structure and function of our 5-station simulator, and the attachment of femoral stems and acetabular cups to the simulator have been described in detail by Saikko et al. (1992). The motion simulated flexion-extension; range was 60 degrees and the duration of the extension phase of the gait cycle was 0.62 s and that of the flexion phase 0.31 s. The load simulated the superior-inferior component of the joint contact force; the load was constant, 3.5 kN, during the extension phase and the flexion phase was not loaded. The heel strike occurred at maximum flexion and the toe-off at maximum extension: at maximum flexion the load rose rapidly to 3.5 kN and at maximum extension it dropped rapidly to zero. The trough between the heel-strike and toe-off load peaks, often seen in mechanical studies, was omitted for simplicity. The articulation was lubricated: the prosthetic joints were located upside-down in acrylic lubricant receptacles that functioned also as acetabular cup-holders. The lubricant used in the tests was distilled, deionized water. The temperature of the test environment was maintained at 37 ± 1 °C. The gravimetric method of wear measurement used stationary, loaded soak-control prostheses. The cups could readily be removed for periodic weighings and reassembled in exactly the original position. In addition, frictional torque was measured (Saikko 1992) for 22 different head-cup combinations, including the 10, the wear behavior of which was the subject of the present paper.

The wear of the acetabular cups of each system articulating against metallic or alumina ceramic femoral heads was measured. 5 different head-cup combinations were tested simultaneously in identical conditions. The 3 metal-head tests were performed first, and then the 3 alumina-head tests.

The test frequency was 65 walking cycles per minute, and the duration of one test was 3 million cycles. The cups were weighed at 500,000 cycle intervals. The running between the weighings was continuous, and thus one test took about 5 weeks, including the weighing stops. Wear δM_t was calculated from the initial and latest test specimen masses and from those of the soak control so that the effect of water absorption was minimized, as recommended in ISO TR 9326 (1989). The test and soak control specimens were identical and had been kept in identical conditions during the test runs in regard to exposure to lubricant, environment temperature and loading, and during the weighing stops. Each test cup had a soak control cup of its own and thus a total of 30 soak control cups was needed. Pre-soaking of the acetabular cups was not considered necessary, because the deviations of weight gain rates are apparently small compared with the substantial weight-loss rates due to wear: in absorption tests (Clarke et al. 1985) the water absorption rate of 11 Müller cups was 0.148 ± 0.060 mg/d during the first 28 days, and 0.054 ± 0.012 mg/d during the period of 28–225 days.

The main activities during the periodic weighing stops were:

1. The cups were removed from the simulator together with their holders. It was allowed to touch the cups only wearing disposable polyethylene gloves.
2. The edges of the load-bearing area of the cups were wiped gently with a fingertip to detach the wear particles that were about to come off.
3. The wear particles were rinsed off with a jet of distilled water and stored together with the lubricant from the cup holder for later analysis. The wear particles on the femoral head and on the neck of the stem were also rinsed off.
4. The cups were removed from their holders.
5. The cups were kept in a warm atmosphere (37 °C) until the water on the surfaces had evaporated.
6. The cups were cleaned with ethanol.
7. The cups were vacuum-desiccated for 30 minutes, after which the weight of the residual water in the test specimen was assumed to be equal to that in the soak control specimen. The desiccation unit consists of an Edwards E2M2 high-vacuum pump fitted with a vacuum chamber.
8. The cups were weighed to the nearest 0.01 mg, using a Mettler AT261 Delta-Range analytic balance.
9. The cups were rinsed in distilled water to remove possible dust particles and other airborne contaminants. The cup-holders were also rinsed.
10. The cups were reassembled in the simulator in exactly the original position and the test could be continued.

The wear rates were determined from the weighing data using the linear regression method. The wear-axis intercept was also taken into account, so that wear rate was taken to be the intercept/10^6 cycles + the slope. Positive intercept represents the amount of wearing-in. Negative intercept indicates that the wear rate has increased during the test. In the conversion of the mean wear rate into the clinical wear rate unit, mm/year, it was assumed that (a) 1 million cycles in the simulator represents a 1-year average in vivo activity, (b) the density of the polyethylene is 0.94 mg/mm^3, and (c) the femoral head bored into the cup so that the diameter of the resulting cylindric wear tun-

Table 4. Size, form and surface finish measurements of specimens

Specimen	Diam. (mm)	Departure from roundness (μm)	Surface roughness R_a of intact area (μm)	Surface roughness R_a of load-bearing area, if increased (μm)	Radius of corresponding cup (mm)
Protek					
metal	31.93	1.10	0.021	0.093	16.02
	31.91	1.10	0.024	0.040	16.00
	31.94	1.19	0.009	0.061	16.01
ceramic	32.00	0.16	0.031		15.98
	31.99	0.52	0.021		15.99
Biomet					
metal	31.94	8.51	0.023	0.673	16.06
	31.93	10.39	0.016	0.106	15.98
	31.93	5.61	0.022	0.087	16.00
ceramic	31.90	0.18	0.020		15.82
	32.00	0.32	0.006	0.060	15.94
Link					
metal	32.03	2.51	0.022	0.039	16.38
	32.04	3.08	0.016	0.072	16.32
	32.04	1.64	0.029		16.29
ceramic	31.91	0.68	0.010	0.062	16.49
	31.92	0.34	0.010	0.056	16.38
Howmedica					
metal	31.96	0.77	0.004	0.073	16.23
	31.96	1.08	0.005	0.094	16.26
	31.97	0.45	0.005	0.087	16.32
ceramic	32.00	0.86	0.010		16.41
	32.00	0.28	0.030		16.34
Thackray					
metal	22.20	1.19	0.013	0.033	11.21
	22.20	1.03	0.016	0.080	11.13
	22.14	0.80	0.015	0.036	11.23
ceramic	22.23	1.25	0.020		11.21
	22.24	0.39	0.022		11.26

nel was equal to that of the femoral head, which is a common feature in retrieved cups (Atkinson et al. 1985). Thus the mm/year value was obtained by dividing the mg/10^6 cycles value by ($0.94 \times \pi \times r_H^2$). The wear factor k was calculated according to ISO TR 9326 (1989). The value of the integral $\int Ldx$ of one gait cycle of the HUT simulator is ($3.49 \times r_H$) N m (r_H in millimeters). The value was obtained by load and motion measurements, and numeric integration.

The 2-tailed t-test was used to analyze differences in means. The Smith-Satterthwaite procedure (Milton and Arnold 1990) was used to calculate the number of degrees of freedom, γ, because the variances could not be assumed to be equal, and so γ was not simply $n_1 + n_2 - 2$. In addition, the 95 percent confidence intervals for the differences in means were calculated.

The measurements of diameter, roundness, and surface roughness of the femoral heads (Table 4) were performed after the simulator tests in the same outside laboratory and in the same way as those of the heads used in the frictional studies (Saikko 1992). The "original" R_a value was measured on the intact non-load-bearing area, and the R_a value of the load-bearing area was measured perpendicular to the direction of sliding. The specimens were not measured prior to the wear tests in order to ensure the preservation of the original surface finish. The heads met the requirements of ISO 7206-2 (1987) with a few exceptions: the departure from roundness of 2 Biomet's heads exceeded the maximum limit of 8 μm for titanium alloy, and the diameter of all Link's metal heads were larger than their nominal diameters. The internal radius r_C of acetabular cups was measured in-house after the simulator tests using the 3-ball method (Saikko 1992). It was measured in the mediolateral direction, 45 degrees from the apex, because it was assumed to represent the original value of interest. In some joints, r_C was smaller than the radius of the head r_H, but that does not necessarily mean that the clearance was negative: the internal geometry of these cups proved slightly toroidal, the apex area being flatter, as described by Saikko (1992). The equatorial diameter was larger than that of the head, but the equatorial diameter does not contain all the essential information on the geometry when distinct non-spherical features are present. The extent of contact is likely to change considerably

Table 5. Linear regression parameters for variation of wear of acetabular cups with number of cycles, wear rate (mg/10⁶ cycles), mean wear rate (mg/10⁶ cycles), mean yearly wear (mm/year) and mean wear factor (10^{-6} mm^3 N^{-1} m^{-1}). 95% confidence limits

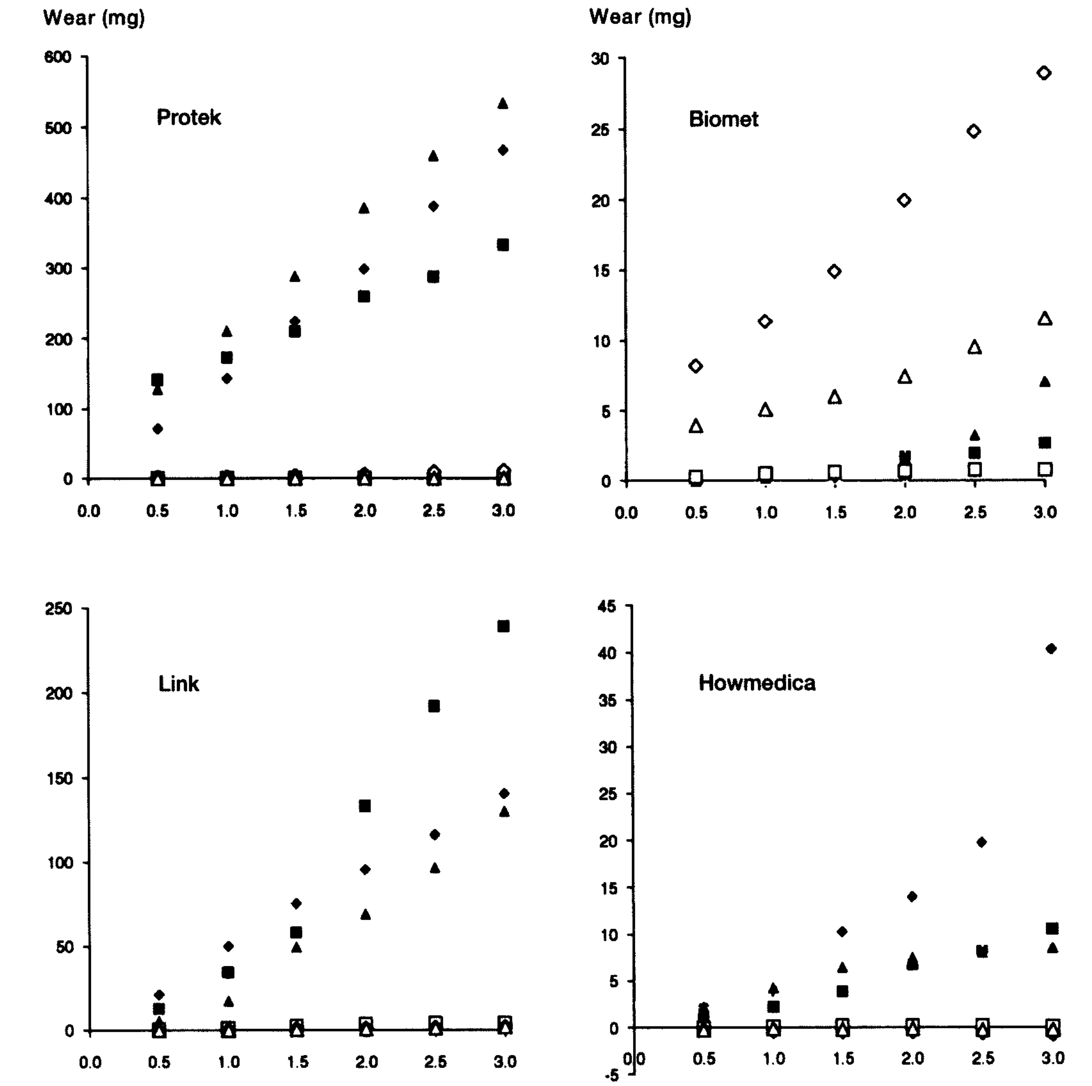
Specimen	Intercept (mg)		Slope (mg/10^6 cycles)		Correlation coefficient r	Wear rate	Mean			
							Wear rate		Yearly wear	Wear factor
Protek										
metal	99.0	*14.1*	77.0	*7.25*	1.00	176	178	*82.5*	0.24	3.39
	–13.4	*12.9*	159	*6.63*	1.00	146				
	48.1	*16.5*	164	*8.47*	1.00	212				
ceramic	0.11	*0.09*	0.21	*0.04*	0.99	0.32	0.88	*2.67*	0.00	0.02
	–2.30	*1.72*	4.42	*0.88*	0.99	2.12				
	0.12	*0.14*	0.08	*0.07*	0.85	0.21				
Biomet										
metal	–0.90	*0.81*	1.16	*0.41*	0.97	0.26	0.20	*0.18*	0.00	0.00
	–0.21	*0.25*	0.33	*0.13*	0.96	0.12				
	–2.38	*3.38*	2.61	*1.74*	0.90	0.22				
ceramic	0.25	*0.19*	0.19	*0.10*	0.94	0.44	5.69	*14.0*	0.01	0.11
	3.09	*1.84*	8.54	*0.94*	1.00	11.6				
	1.95	*1.33*	3.03	*0.68*	0.99	4.99				
Link										
metal	–56.1	*49.6*	95.9	*25.5*	0.98	39.7	37.3	*30.5*	0.05	0.71
	1.70	*7.59*	46.5	*3.90*	1.00	48.2				
	–26.3	*15.7*	50.3	*8.05*	0.99	24.0				
ceramic	0.12	*0.99*	1.40	*0.51*	0.97	1.52	0.67	*1.92*	0.00	0.01
	–0.23	*0.27*	0.74	*0.14*	0.99	0.51				
	–0.89	*0.47*	0.89	*0.24*	0.98	0.00				
Howmedica										
metal	–1.18	*1.50*	3.79	*0.77*	0.99	2.61	3.89	*2.79*	0.01	0.07
	–9.08	*15.3*	13.8	*7.84*	0.93	4.72				
	1.80	*2.09*	2.53	*1.07*	0.96	4.33				
ceramic	0.00	*0.21*	0.06	*0.11*	0.59	0.05	–0.17	*0.55*	0.00	0.00
	–0.21	*0.11*	–0.17	*0.06*	–0.97	–0.38				
	–0.14	*0.20*	–0.04	*0.10*	–0.46	–0.18				
Thackray										
metal	25.4	*3.63*	25.5	*1.86*	1.00	50.9	53.1	*21.0*	0.15	1.46
	–21.6	*14.4*	84.0	*7.38*	1.00	62.4				
	7.24	*15.6*	38.8	*8.03*	0.99	46.0				
ceramic	–2.36	*3.79*	3.16	*1.95*	0.91	0.80	0.83	*0.27*	0.00	0.02
	0.32	*0.35*	0.63	*0.18*	0.98	0.95				
	–0.09	*0.34*	0.83	*0.18*	0.99	0.74				

during articulation due to the low creep resistance of polyethylene, wear and elevated temperature, and so the concept of clearance is not readily applicable. The specimens of the sixth test (alumina heads) are not included in Table 4 because the test was considerably extended and it was still running when the present paper was prepared. The test started in November 1991 and reached 35 million cycles at completion in February 1993. Hence it is probably the longest test ever performed on a hip joint simulator.

Results

The Protek, Link and Thackray metal cups clearly produced more polyethylene wear debris than the rest (Figure 1). The cups that articulated against alumina ceramic heads usually produced minor amounts of polyethylene wear debris, but so did also the Biomet metal cups. The Howmedica ceramic cups were the only ones that did not produce any polyethylene particles at all, and the wear rate was, indeed, zero. δM_t was often slightly negative, indicating that the water absorption rate of the test cups was higher than that of soak controls, probably because of the sliding motion.

The results of least squares fits of straight lines to the data are presented in Table 5 for each cup. In most cases the value of correlation coefficient r was high, indicating a clear linear relationship. In some cases the intercept was negative, but the same method of determining the wear rate was still applied to every cup for uniformity. The Protek metal heads caused the highest mean wear rate (0.24 mm/year), and Howmedica ceramic heads the lowest. Some 95 percent confidence limits of mean wear rates (Table 5) included even negative wear rates, which are naturally impossible in reality, and are due to the small sample size and



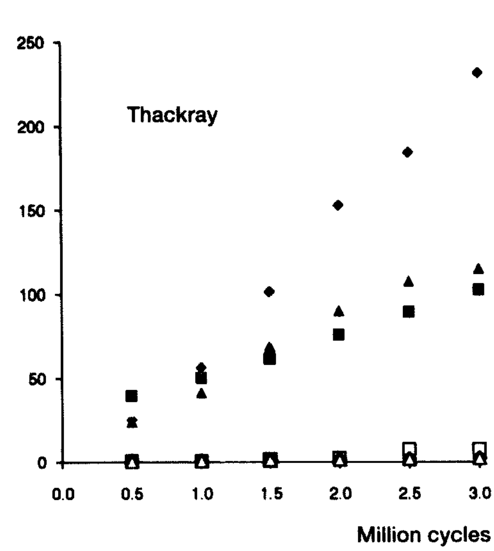


Figure 1. Variation of wear of acetabular cups with number of simulated level walking cycles. Solid squares, diamonds and triangles refer to the 1st, 2nd and 3rd tests (metal heads), and open squares, diamonds and triangles refer to the 4th, 5th and 6th tests (alumina heads), respectively. Note differences in scale for wear rate.

Table 6. Comparison of mean wear rates for metal vs. ceramic heads

Samples compared	Difference in means (mg/10^6 cycles)	γ	t	p	At 95% confidence level, the difference in means is between (mg/10^6 cycles)
Protek	177	2	9.24	0.006	95 and 260
Biomet	–5.5	2	–1.69	0.117	–19 and 8.5
Link	37	2	5.15	0.018	6.0 and 67
Howmedia	4.1	2	6.13	0.013	1.2 and 6.9
Thackray	52	2	10.72	0.004	31 and 73

assumption of normal distribution. Considerable scatter is, however, typical of wear rate data (Atkinson et al. 1985, Wallbridge and Dowson 1987).

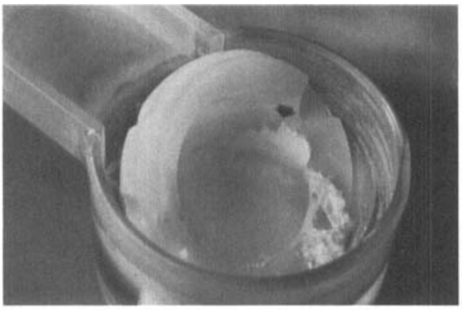
Polyethylene wear debris accumulated mainly on the anterior side of the load-bearing area, because the articulation was loaded only during extension (Figure 2). Usually, the higher the wear rate, the larger were the polyethylene particles. In the cups with the highest wear rates (metal heads Protek, Link, and Thackray), most of the polyethylene debris consisted of large, elongated, thin, chip-like shreds, or streamers, even as long as a few centimeters, and a distinct ridge could be seen between the loaded and non-loaded areas after the test. On the loaded area, grooves parallel to the direction of sliding were typical. At the anterior edge of the loaded area, plate-like formations could be detected. They still adhered to the surface and apparently represented a stage of the chip formation process. In cups, articulating with Biomet and Howmedica metal heads, the streamers were clearly finer.

Taper-fit of the femoral heads proved problematic. When the heads were removed after the completion of a test, the conical contact surfaces always appeared different from the initial condition to a greater or lesser extent. The head-spigot interface of Pm and Tm was badly corroded (Figure 3). Corrosion was also evident on the spherical articulating surfaces of these 6 stainless steel heads, to which the high wear rates of the corresponding cups are apparently attributable. All Co-Cr-Mo-recess—Co-Cr-Mo-spigot interfaces (Link metal and Howmedica both metal and ceramic with the Vitallium adapter sleeve) were somewhat corroded, and at the Ti-6Al-4V-head—Ti-6Al-4V-spigot interface (Biomet metal), discoloration was observed. In all alumina heads there was metal on the conical contact surface of the recess, most pronounced in Protek and Biomet heads, that had been attached to titanium alloy stems (Figure 4). The metal was transferred from the spigot because alumina is much harder than the implant metals. The transfer was manifested as darkening of the light-colored alumina surface. The metal layer indicates that there was micromotion between the head and the spigot during articulation. Titanium is known to be especially susceptible to rubbing. Prior to the following test, rust or any protrusive damage on the spigots had to be removed, using fine abrasive paper, in order to ensure the correct taper-fits for the new heads. The spherical articulating surface of one of the Bc head darkened also during the test at the load-bearing area. A scanning electron microscope analysis revealed streaks of titanium adhering to the surface at the load-bearing area of the spherical surface (Figure 5), which explains the darkening. The titanium probably originated from the head-spigot interface. The 2 other interfaces in which titanium was involved and which were exposed to the lubricant, viz., the interface between the acetabular shell and the acetabular liner, and the interface between the acetabular shell and the acetabular shell holder of PMMA, are very unlikely sites of origin. It is not surprising that the wear rate was the highest of all the cups that articulated against an alumina head, because the streaks obviously function as abrasive protrusions that plough the soft polyethylene counterface causing considerable wear. The articulating surfaces of ion-implanted Ti-6Al-4V heads (Biomet metal) turned bluish, and the surface roughness increased notably (Table 4), but in spite of this the wear rates of the corresponding cups were strikingly low; the total wear of Biomet metal-articulating cups were 2.7, 0.9, and 7.1 mg.

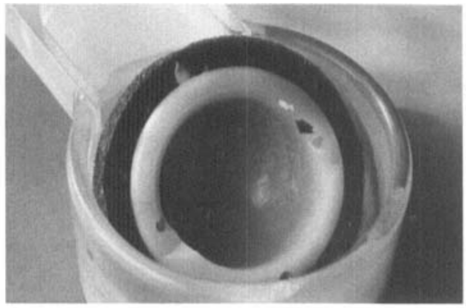
For Protek, Link, Howmedica, and Thackray the mean wear rate of the cups was higher for metal than for ceramic heads (Table 6). These differences in means were significant, but the difference between the Bm and Bc cups was not. In these 5 comparisons the main affecting variable is the head material, and so in 4 of the 5 systems studied, the alumina ceramic head was superior to the metallic. The behavior of Bc was erratic and more tests might be worthwhile.

There was no correlation between wear rate and coefficient of friction. New joints, however, were used in friction measurements (Saikko 1992), and it is likely that the coefficient of friction varies considerably during a long-term wear test, because of wear-related changes at the articulating surfaces.

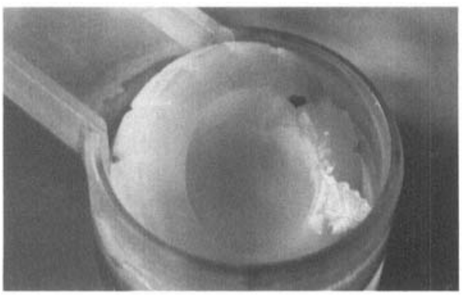
Figure 2. Acetabular cups in their acrylic holders after the completion of the third metal-head test. Note polyethylene wear particle formations on the anterior side of the load-bearing area (arrow).



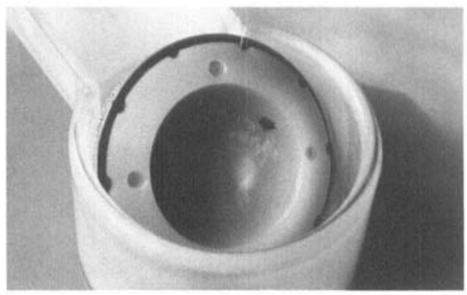
Protek cup 3



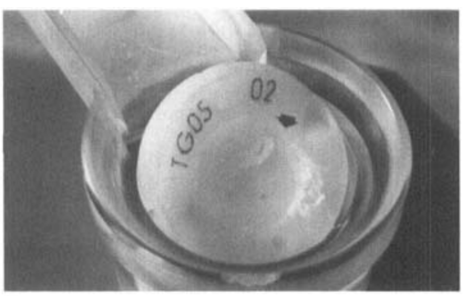
Biomet cup 3 in Mallory-Head acetabular shell.



Link cup 3.



Howmedica cup 3 in P.C.A. acetabular shell.



Thackray cup 3.

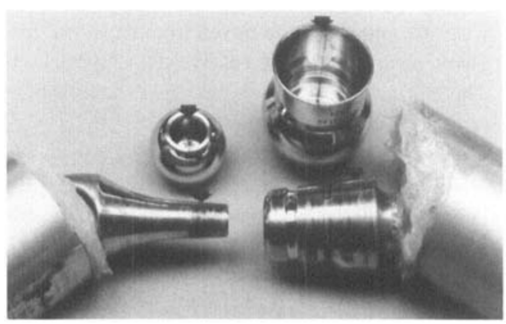


Figure 3. Conical contact surfaces of modular stainless steel femoral components corroded considerably during the tests (arrows). Thackray's head and Elite stem on the left, and Protek's head and Müller stem on the right. Stems are in their holders.

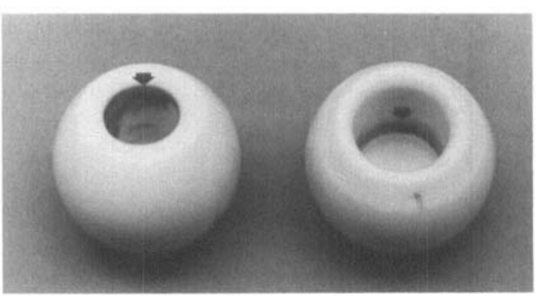


Figure 4. Traces of titanium on the conical contact surface of Biomet's alumina femoral head 1 (left), and traces of Co-Cr-Mo on Link's alumina femoral head 1, respectively (arrows). The shape of the trace on the Link head discloses that the cone angle of the recess is distinctly higher than that of the spigot of the Link's SP II Modell Lubinus stem.

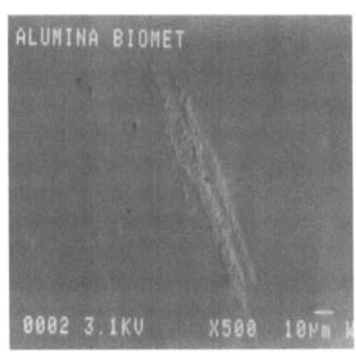


Figure 5. Scanning electron micrograph from the spherical load-bearing area of Biomet's alumina femoral head 2 showing a streak of titanium, which is likely to originate from the spigot of the stem.

Discussion

Simulator vs. in vivo results

The evaluation of wear results obtained from a hip joint simulator is difficult because there is no accurate method of quantifying wear of acetabular cups in vivo or ex vivo. The radiographic in vivo method and the ex vivo method of examining internal volume changes of acetabular cups removed from patients cannot distinguish between wear and creep deformation. The creep resistance of polyethylene is relatively poor. Thus these 2 methods are actually methods of measuring penetration, not wear, because the proportion of creep remains unknown. Our mm/year values represent true wear, being free from the creep error, since the method of wear measurement is gravimetric. However, in vivo and ex vivo studies are the best references available for comparison with simulator results.

Schüller and Marti (1990) performed radiographic penetration measurements on polyethylene cups for 66 Weber prostheses at 9–11 years; half represented Co-Cr-Mo and half alumina. The mean wear (penetration) rates were 0.10 mm/year and 0.03 mm/year, respectively, but the proportion of creep was not estimated. The difference was consistent with the present study, although less pronounced.

Internal volume changes of 25 acetabular cups of Charnley prostheses, removed from patients after 2–16 years, have been examined by Atkinson et al. (1985). The mean wear (penetration) rate was 0.19 (< 0.005–0.623) mm/year (!), and the mean wear factor k was 2.9 (< 0.09–7.2) $\times 10^{-6}$ mm^3 N^{-1} m^{-1}; it was assumed that the proportion of creep was less than 10 percent. In the calculation of k, the activity histories of the prostheses naturally had to be estimated, whereas, in a simulator study, the activity history is known precisely. In the present study, the mean wear rate of sample Tm, in essence modular Charnley prostheses, was 0.15 mm/year and k was 1.5×10^{-6} mm^3 N^{-1} m^{-1}. The value of the integral in the denominator of the expression of k (see Abbreviations) used in the internal volume change study (Atkinson et al. 1985) was only about a half of that of our simulator. The number of load cycles per annum was estimated on the basis of the age of the patient. The means of these 2 studies was rather similar, so our HUT simulator may be regarded as capable of producing realistic wear rates.

The mean R_a value of the load-bearing area of sample Tm femoral heads was 0.05 μm, which is close to the 0.054 μm reported in the internal volume change study as a mean for 12 femoral components of the sample under consideration. It was suggested that bone cement particles frequently detected in retrieved cups cause roughening of stainless steel femoral heads, and thus increase the wear on the cup. In the present study, however, roughening was attributable to corrosion. The mean penetration rate in 87 retrieved Charnley cups examined later by the same team was 0.21 mm/year (Isaac et al. 1992). In 21 Charnley cups removed from patients, a ridge between the high-wear and low-wear areas has been observed (Dowling et al. 1978). This is also consistent with our observations.

Abrasion of metal heads

Scratches parallel to the direction of sliding usually appeared on the load-bearing area of metallic heads during the tests. On ceramic heads, however, it was usually difficult to discern any changes, and the surface roughness was seldom increased (Table 4). This absence of abrasive damage on ceramic heads may partially explain the difference between wear rates against metallic vs. alumina heads. The abrasion of metallic heads may be attributable to airborne contaminants, such as dust particles, although care is naturally taken that the risk of contaminant entrapment is minimized. Particles from the interface between the head and the stem may also abrade a metallic head. The superior abrasion resistance of alumina ceramic is an indisputable asset, and the better wettability of the alumina surface apparently results in more efficient boundary lubrication and consequently lower wear rate of the polyethylene counterface. In addition, alumina has a negligible propensity to corrosion and to metal ion release during articulation compared with metallic implant materials (Davidson and Kovacs 1989).

The excellent behavior of sample Hc shows that in vitro wear of polyethylene cups can really be zero, as they articulate against alumina heads if no outside particles enter between the articulating surfaces. Slight flattening of the machining grooves at the load-bearing area of Hc cups was practically the only sign of the 3 million simulated gait cycles. The Co-Cr-Mo backing of the inserts may have been a contributing factor, since non-metal-backed sample Link ceramic, which also used a Co-Cr-Mo alloy stem, produced polyethylene debris, albeit minute amounts. It is, indeed, difficult to eliminate the risk of particles getting between the articulating surfaces in vivo. Metal particles from surgical tools and from the implant, and cement particles, as well as wear debris have been shown to be present in synovial fluid aspirated from patients with artificial joints (Mears et al. 1978).

Taper-fit attachment

Crevice corrosion at the head-spigot interface of retrieved modular femoral components with a taper-fit method of head attachment has recently been detected (Collier et al. 1991, Mathiesen et al. 1991). In our study also, corrosion was detected at the interface. The taper-fit attachment may obviously cause a considerable impairment of the otherwise excellent articulation. It seems that the taper-fit cannot be the ultimate method of attaching the modular head, but better methods should be developed. Since the taper-fit induces considerable tensile stresses to the head (Seidelmann et al. 1982), the method may even be considered unsound in the attachment of brittle ceramic heads. Fractures of ceramic heads are known to have occurred in vivo. The behavior of the attachment of the Tc heads, based on a parallel-sided press-fit and an intermediate polyethylene sleeve, was quite flawless in the present study. Note that the prostheses are located upside-down in the HUT simulator in order to eliminate the risk of the articulation drying out due to air bubbles. The shortcoming of the upside-down position is that the particles generated at and leaving the head-spigot interface, due to gravity, probably get more easily between the articulating surfaces than they would if the prostheses were not inverted. The loose polyethylene particles, however, are not easily entrapped, because they are lighter than water and rise to the surface. The facts that the polyethylene particles float, and that the simulator lacks activities other than level walking, explain why the large polyethylene streamers are not easily entrapped between the articulating surfaces and ground finer.

Titanium femoral heads

Titanium alloy femoral heads are not very popular at present. The literature contains contradictory viewpoints on the suitability of the material (Clarke et al. 1983, Nasser et al. 1990, Davidson 1992). In our simulator the performance of ion-implanted sample Biomet heads was, perhaps somewhat surprisingly, quite satisfactory considering the very low wear rate of the cups, despite the considerable roughening of the titanium articular surfaces during the tests. The Biomet system was indeed the only one, of the 5 systems studied, in which the ceramic head was not better than the metallic. Ion implantation of titanium seems advantageous, since non-ion-implanted titanium heads were severely damaged in relatively short friction tests, showing extremely poor wear resistance (Saikko 1992). Sioshansi et al. (1985) also found the ion implantation effective in pin-on-disk tests. The improvement obtained by ion implantation has been related to the increased surface hardness. The hard surface layer, however, is very thin, about 0.1 µm, and it has been shown by Davidson and Kovacs (1989) that the breakdown of the surface layer can occur in the presence of bone cement debris.

Lubricants in wear tests

The use of protein-containing lubricants in laboratory wear tests has been emphasized by McKellop and Clarke (1985), in order to prevent the formation of a polyethylene transfer layer on the counterface, since a transfer layer was not observed on retrieved femoral components, but a heavy transfer occurred in water or saline lubricated, statically loaded, pin-on-disk tests. The fact that such a heavy transfer was not observed in the present study indicates that the presence of proteins is not the only factor that determines polyethylene transfer. In our simulator, water is likely to be drawn between the articulating surfaces during the non-loaded flexion phase of the gait cycle. That water is then apparently capable of lubricating the articulation during the loaded extension phase and of retarding the formation of the transfer layer. The lubrication mechanisms are difficult to specify, but squeeze-film and mixed lubrication seem the most likely ones (Saikko 1992). As synovial fluid is not available in such amounts as required by an extensive laboratory wear test scheme, some researchers prefer to use blood serum as lubricant, although its composition is different from that of synovial fluid in a replacement hip joint. A literature survey indicated that when blood serum was used, only relatively short tests or a single test of long duration were performed. This was probably due to problems posed by the degradation of the fluid, which, in addition, involves another source of error, since the lubrication properties of degradable fluids are likely to change with time in long-term wear tests. This issue is generally, however, not discussed in the studies in which degradable fluids have been used as lubricants. The most essential differences in wear factors are observed between wet and dry conditions (Atkinson et al. 1985). Hence, distilled water was considered the obvious choice.

Comparison of simulators

The 10-station hip joint simulator of the University of Southern California (USC) (McKellop and Clarke 1984, 1985) is perhaps the most widely known. Its structure and function differ from those of our simulator (Saikko et al. 1992) in many respects. Perhaps the most notable difference is that in the HUT simulator the direction of loading is virtually stationary relative to the cup, representing the superior-inferior component of the joint contact force. In the USC simulator the direction of loading is stationary relative to the femoral head, and the cup is moved relative to the head so that the contact area rotates about the apex of the cup, the circumference of the resulting circular wear track on the cup being, in theory, $2.46 \times r_C$. Hence, the motion of the USC simulator is unidirectional, of constant velocity, not reciprocating as in vivo and in the HUT simulator. Wroblewski (1985) examined 22 Charnley cups removed from patients and observed that the head had penetrated into the cup in a certain direction. This indicates that an arrangement with the direction of loading stationary relative to the cup is more realistic. In addition, the HUT simulator has the self-centering of the prostheses, loaded soak control, and a controlled temperature. The USC simulator does not have these features. The lack of self-centering means that the true load acting between the head and the cup is unknown, due to lateral force components that are inevitably caused by the malalignments, however small, of the prosthetic components relative to the motion and load axes of the apparatus. The load of the USC simulator is not measured, but the servo-control of the load uses the signal of a pressure transducer that measures the pressure acting in the hydraulic loading cylinder as an indirect feed-back. This may increase the inaccuracy of the load, because the piston of the cylinder is not free to move with the malalignment and deformation of the cup: the cylinder is rigidly attached to the frame of the apparatus and so is the femoral head to the piston rod. The loading of the soak control cups is important because the loading has been found to considerably increase the water absorption rate. The tests should naturally be performed at body temperature. Only 1 design, Link's 32-mm Co-Cr-Mo on UHMWPE, was included both in the present study and in the study based on the USC simulator (McKellop and Clarke 1985). In the latter report the components were not specified by catalogue numbers, though, and the duration of a test was only 1 million cycles and the peak value of the load curve was only 2 kN. The wear rates obtained were 39.7, 48.2, and 24.0 mg/10^6 cycles in the HUT simulator, and 36.4, 32.4, and 35.9 mg/10^6 cycles in the USC simulator. The difference in the mean wear rates was small, despite the many differences between the simulators and the test conditions.

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