Wear of polyethylene acetabular cups against alumina femoral heads

5 prostheses compared in a hip simulator for 35 million walking cycles

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5 ultra-high molecular weight polyethylene acetabular cups articulated against alumina femoral heads for 35 million walking cycles in a hip joint simulator designed for wear tests of total hip prostheses. The specimens were from Protek, Biomet, Link, Howmedica and Thackray. In the Howmedica specimen, the wear was zero, and in Link and Thackray practically negligible. In Biomet, however, the wear was noteworthy and in Protek disastrous after 20 million cycles because in these 2 prostheses the head was attached to a titanium-alloy stem by taper-fit: titanium-alloy particles that were removed from the taper because of micromotion between the head and stem were entrapped between the head and cup, adhering to the head and making it rough, which led to severe abrasive wear of the Protek cup. It was worn through at 26 million cycles, the total wear being 3,170 mg. In Biomet, Link and Thackray, the total wear was 124, 5.3 and 17.6 mg, respectively. Polyethylene wear particles may lead to adverse tissue reactions and eventual loosening of the implant. The results indicate that by the use of alumina heads, polyethylene wear can be eliminated, but this advantage may be lost if the head is attached to a titanium-alloy taper.

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Radiographic measurements of total hip prostheses have shown that the penetration of stainless steel femoral heads into ultra-high molecular weight polyethylene (UHMWPE) acetabular cups is 2.5 times higher than that of alumina ceramic (Al_2O_3) heads (Oonishi et al. 1992). The corresponding ratio between Co-Cr-Mo and alumina has been found to be 3.7 (Schüller and Marti 1990). Laboratory wear studies show consistently that alumina heads are superior to stainless steel and Co-Cr-Mo heads (Saikko et al. 1993), although the difference was more pronounced. The superiority of alumina as a head material is attributable to its higher abrasion and corrosion resistance, better smoothness and wettability, and lower frictional heat (Davidson and Schwartz 1987). A low wear rate of the cup is important because polyethylene wear particles are likely to be the main cause of adverse tissue reactions leading to the loosening of the implant (Amstutz et al. 1992). With hip joint simulators, the wear and friction behavior of new prostheses can and should be examined prior to implantation (Clarke 1981). In addition, there is reason to study established materials and designs by running very long tests corresponding to service lives that exceed those of contemporary clinical follow-up studies, to see if major changes in the behavior occur in the long-term. In the present study, the long-term wear behavior of the polyethylene/alumina articulation was investigated with a hip joint simulator, the test length corresponding to a service life of 35 years. So far as the author knows, no study of this kind has been previously reported. It was thought to be interesting to see if the wear rate in the long term will remain as low as during the first few million cycles (Saikko et al. 1993), or if some unexpected phenomena will cause a deleterious late change in the wear mechanisms.

Material and methods

The components included in the test are presented in Table 1. The measured head diameters and surface roughnesses are presented in Table 2, and the measured taper angles of the heads and the stems in Table 3. In the following, the specimens are called, after their manufacturers or vendors, Protek, Biomet, Link, Howmedica and Thackray. The "walking" simulator and the test procedure have been described in detail elsewhere (Saikko et al. 1992, 1993). The apparatus simulates the flexion-extension motion and the vertical

Table 1. Description on package, catalog number, material, and nominal diameter (mm) of specimens, and method of attachment of head to stem

Manufacturer or vendor	Acetabular cup	Femoral head	Femoral stem
Protek	Full profile cup 62.32.50 UHMWPE o.d. 50	Müller modular head 12.32.06 alumina Biolox dia. 32	Müller modular straight stem 22.00.29-200 Ti-6Al-7Nb taper-fit 1 : 10
Biomet	Modular acetabular liner 104308 UHMWPE (backed by Mallory-Head acetabular shell 11-104256 Ti-6Al-4V, o.d. 56)	Modular head component 131407 alumina Vitox dia. 32	Modular femoral component 163325 Ti-6Al-4V taper-fit 1 : 14.3
Link	Lubinus 102-110 UHMWPE o.d. 52	SP II - Prothesenkopf 128-709 alumina Biolox dia. 32	SP II Modell Lubinus 128-719 Co-Cr-Mo taper-fit 1 : 10
Howmedica	P.C.A. acetabular insert 6285-0-525 UHMWPE (backed by P.C.A. acetabular shell 6289-5-052 Co-Cr-Mo, o.d. 52)	Ceramic head 6290-1-032 alumina Biolox dia. 32 (Note: not a part of P.C.A. system)	P.C.A. primary femoral stem 6280-7-010 Co-Cr-Mo taper-fit 1 : 10 (between head and adapter sleeve of Vitallium, MU40-0-016, that makes use of alumina head possible)
Thackray	Charnley 62-3717 UHMWPE o.d. 43	Charnley ceramic head 62-6392 alumina Vitox dia. 22.25	Charnley femoral hip prosthesis 62-6392 stainless steel press-fit (polyethylene sleeve between head and trunnion)

Table 2. Measured diameter and surface roughness of alumina femoral heads

Specimen	Diameter (mm)	Original surface roughness R_a (μm)[a]	Surface roughness R_a of load-bearing area after the test (μm)
Protek	31.99	0.008	0.3 (0.1–0.8)
Biomet	31.86	0.011	0.03 (0.02–0.06)
Link	31.92	0.008	unchanged
Howmedica	31.97	0.009	unchanged
Thackray	22.23	0.011	unchanged

[a]Parameter R_a is the arithmetic mean of the deviations of the profile from the mean line.

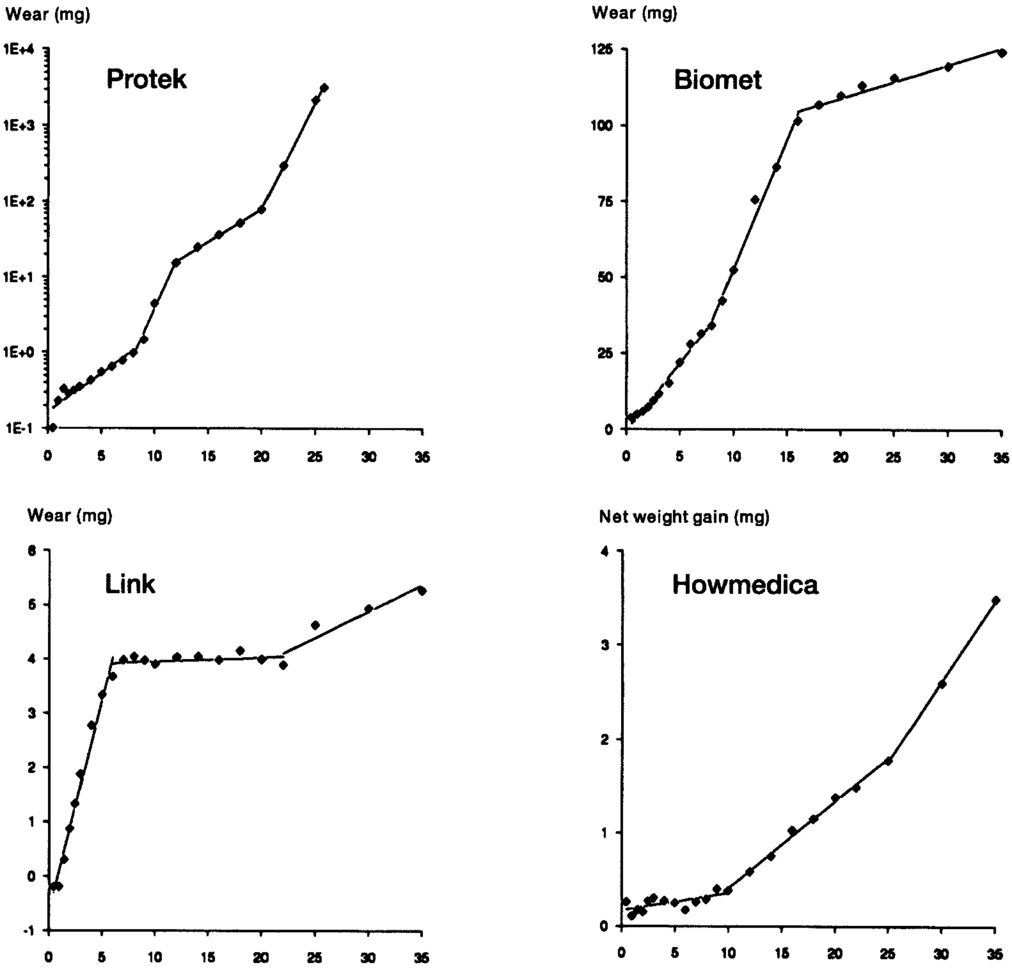
Table 3. Taper angles of heads and stems measured after simulator test with CNC coordinate-measuring apparatus

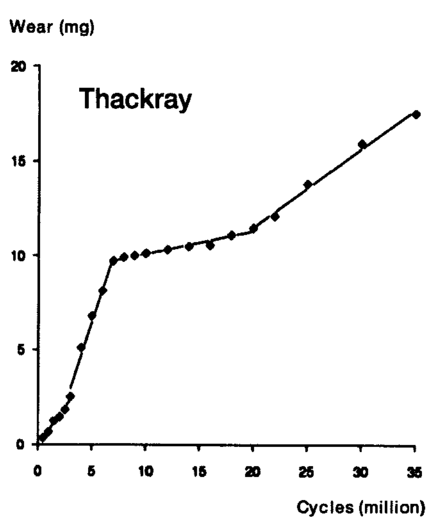
Specimen	Head	Stem
Protek	5°46'	5°38'
Biomet	4°2'	3°59'
Link	5°47'	5°40'
Howmedica	5°49'	5°40'[a]

[a]Outside taper of adapter MU40-0-016

component of the joint contact force in walking. The test frequency was 65 walking cycles per minute. The temperature of the test environment was 37 °C and the articulation was lubricated by distilled deionized water. At intervals, the test was halted for a gravimetric wear measurement that necessitated the removal of the cups. The wear was calculated from the weighings of the test and control cups so that the error caused by water absorption was minimized. The weighings were done after vacuum desiccation. The wear was calculated so that the mass of the test cup was subtracted from the initial mass, and the weight gain of the control cup was added to the remainder, because in the test cup the weight loss caused by the possible wear is partly or completely masked by the water absorption. The test length was 35 million cycles. Hence, the test was apparently the longest ever performed with a hip joint simulator; it took about 15 months to accomplish. The test length can be estimated to correspond to 35 years of in vivo activity. The first 3 million cycles of the test were included in the preceding article (Saikko et al. 1993) as the third alumina-head test.

Figure 1. Variation of wear or net weight gain of acetabular cups with number of simulated walking cycles. Note the exponential relationship in Protek. Note also that the y-axes have different scales. The net weight gain in Howmedica indicates directly how much more water the test cup absorbed compared with the control cup, since no wear occurred.





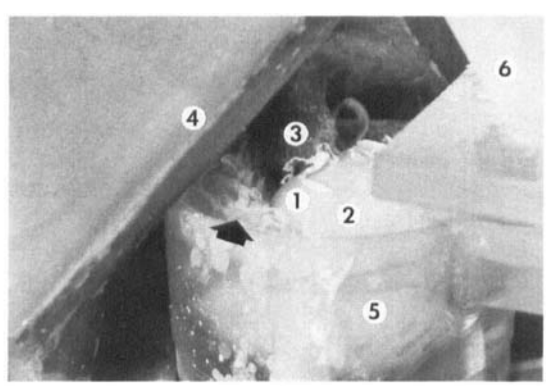


Figure 2. Protek in the hip joint simulator just prior to the wearing through of the cup. Note the bunch of streamer-like polyethylene debris (arrow). 1 femoral head, 2 acetabular cup, 3 neck of the femoral stem, 4 flexion-extension cradle, 5 cup holder, 6 water replenishment bottle.

Table 4. Regression equations, parameters *a* and *b* and correlation coefficient *r* for wear data shown in Figure 1; *w* wear of the cup (mg) (net weight gain in Howmedica) and *n* number of accumulated cycles (million)

Specimen	Form of equation	n	a	b	r
Protek	$w = a \times b^n$	0.5–8	0.16	1.26	0.93
		8–12	2.85×10^{-3}	2.05	0.99
		12–20	1.46	1.22	1.00
		20–25.8	1.96×10^{-4}	1.91	1.00
Biomet	$w = a + b \times n$	0.5–8	–0.33	4.40	0.99
		8–16	–33.2	8.57	0.99
		16–35	87.0	1.10	0.97
Link	$w = a + b \times n$	0.5–6	–0.70	0.79	0.99
		6–22	3.87	7.50×10^{-3}	0.34
		22–35	1.96	9.73×10^{-2}	0.94
Howmedica	$w = a + b \times n$	0.5–10	0.17	1.85×10^{-2}	0.69
		10–25	–0.53	9.34×10^{-2}	1.00
		25–35	–2.51	0.17	1.00
Thackray	$w = a + b \times n$	0.5–3	-8.60×10^{-2}	0.83	0.99
		3–7	–2.23	1.74	0.99
		7–20	8.90	0.12	0.98
		20–35	3.10	0.42	1.00

Results

The variation of wear of the cups with number of cycles is shown in Figure 1. In the case of Howmedica, however, the weight change was not caused by wear, but merely by water absorption that proved higher in the test cup than in the control cup, resulting in a net weight gain in the calculation. Therefore, the assumption that the amount of absorbed water in the test cup is equal to that of the control cup proved slightly inaccurate. The regression equations and calculated regression parameters and correlation coefficients are presented in Table 4. In all 5 specimens, the weight-change rate varied during the test, so that 3 or 4 different rates were observed. Hence, it is not possible to give a single value for the wear rate, as is often done in short-term tests of a few million cycles duration. In Protek, the relationship between weight change and accumulated number of walking cycles was exponential, whereas in the rest it was (recti)linear. Note that the y-axes in Figure 1 have different scales and that in Figure 1 (Protek) the scale of the y-axis is logarithmic, because the wear rate increased so steeply after 20 million cycles. The wear in Protek was moderate up to 20 million cycles, resembling that of Biomet, but after that it became disastrous. Large bunches of streamer-like polyethylene shreds were produced (Figure 2) and the head bored through the cup. The cup was worn through at 25.8 million cycles, the total wear being about 3,170 mg. The prosthesis had to be removed from the simulator and the test was continued with the remaining 4. In Biomet, the total wear was 124 mg. In Link and Thackray, the wear was practically negligible, although occasionally minor amounts of polyethylene particles were detected, the total wear being 5.3 and 17.6 mg, respectively. In Howmedica, the wear rate was zero. Macroscopic fractures could be seen in all cups.

The load-bearing area of the Protek head turned darker and duller during the test. When the head was inspected after the test, it was found that the surface roughness had increased considerably in the load-bearing area (Table 2). A scanning electron microscope analysis showed that this was caused by titanium particles adhering to the bearing surface (Figure 3). The particles clearly originated from the taper of the stem. The taper in the recess of the head was dark grey because of transferred titanium alloy (Figure 4), and loose titanium-alloy particles were found in the bottom of the recess, indicating micromotion and wear between the head and the stem during articulation. The other specimen with a titanium-alloy stem, Biomet, showed similar behavior, although to a much more moderate extent. In the specimens with Co-Cr-Mo stems, Link and Howmedica, considerable corrosion occurred in the taper-fit (Figure 4), but this did not have an adverse effect on the wear of the cup. In Thackray, a polyethylene sleeve completely prevented the contact between the head and the stem, and so there was no damage (Figure 4). Note that Thackray was the only specimen in which the attachment was based not on taper-fit, but on press-fit, the trunnion of the stem and the recess of the head being parallel-sided.

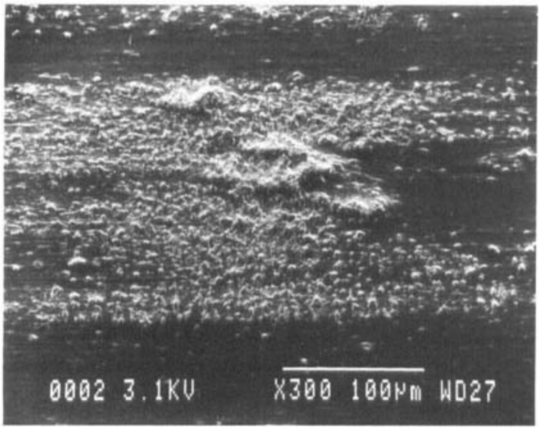


Figure 3. Scanning electron micrograph from the spherical load-bearing area of Protek head, after 25.8 million cycles in the simulator, showing titanium which forms protrusive streaks in the direction of sliding that is horizontal relative to this micrograph.

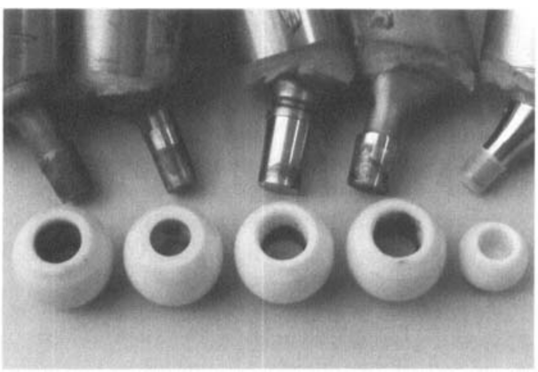


Figure 4. Femoral heads and proximal ends of stems that are in their holders, after the test. From left to right: Protek, Biomet, Link, Howmedica and Thackray. Note damage in all four taper-fit systems. Note also the adapter sleeve on the Howmedica spigot, and the polyethylene sleeve on the parallel-sided trunnion of the Thackray stem.

Discussion

It is generally known that a cautious attitude should be taken towards single-specimen wear test results, as considerable scatter is especially typical of wear (Clarke 1981). It is nevertheless clear that in some cases a wear test of proper duration may be so time-consuming and laborious that even a single result can be considered valuable. The duration of the present test was so long that it corresponds to a service life that no prosthesis has yet had in vivo, 35 years. The poor behavior of Protek could not have been predicted by a few million cycle test; the wear proved disastrous only after 20 million cycles. The wear rate determined from the first 3 million cycles was 0.2 mg per 1 million cycles (Saikko et al. 1993) but, just prior to the wearing through of the cup, the wear rate had increased to a value of about 1,300 mg per 1 million cycles! Note further that simulator results should not be regarded as precise predictions of in vivo behavior, but rather as indicative at best, because the conditions in the simulator are but a rough approximation of the very complex in vivo conditions. The preceding study (Saikko et al. 1993) showed that the repeatability of the simulator is satisfactory. Therefore, it is unlikely that the observations of the present study are merely meaningless anomalies. The author is, however, currently designing a new hip joint simulator to increase the testing capacity, so that even very long tests can eventually be repeated.

The behavior of titanium-alloy tapers is not really surprising if one bears in mind that titanium has a relatively poor resistance to rubbing because it is not hard enough. The taper-fits of Protek and Biomet differed in regard to the taper angle, diameter, contact length, surface roughness and angular mismatch (Table 3). The materials were not identical either (Table 1). It appears that the inferiority of Protek, compared with Biomet, is mostly attributable to the very coarse original surface of the Protek's titanium-alloy taper: the taper is manufactured by turning, so that the surface has distinct circumferential grooves. The ridges between the grooves are apparently worn flatter by the much harder alumina counterface on the areas of true contact and thus more titanium-alloy particles are generated than in Biomet's much smoother taper. Just after the installation, the true contact area at the head-stem interface is obviously very small, but it increases with the number of accumulated load cycles; in other words, subsidence takes place, which inevitably includes wear of the titanium-alloy taper. The Co-Cr-Mo tapers of Link and Howmedica did not pose any problems in regard to the wear of the cup, although they were badly corroded. The behavior of the press-fit of Thackray was excellent, but as the product was to be supplied with the head pre-mounted, it did not obtain approval among surgeons and it is not currently available. In all 4 taper-fit specimens, the taper angle of the head was slightly higher than that of the stem (Table 3), the mismatch varying between 3' and 9', which causes stress concentrations in the smaller-diameter end of the taper. The taper angles were measured only after the simulator test and so the wear and corrosion damage may have somewhat influenced the measurements. The traces of the angular mismatch are especially distinct in Link (Figure 4). The traces in Biomet and Howmedica tapers were less axisymmetric than those in Protek and Link, probably because of deviations from roundness. The reduction of the angular mismatch would obviously reduce the stresses in the head (Fessler and Fricker 1989, Andrisano et al. 1990), but would not, however, eliminate the propensity to corrosion of the head-stem interface.

It seems that the risks of the taper-fit have not yet been generally recognized. It may be very difficult to develop a method of attachment as convenient from the point of view of implantation as the taper-fit, but alternatives should be contemplated (Andrisano et al. 1990). Problems with the taper-fit have already become a clinical reality: corrosion has been detected in retrieved prostheses between Co-Cr-Mo heads and Co-Cr-Mo stems (Mathiesen et al. 1991), and between Co-Cr-Mo heads and titanium-alloy stems (Collier et al. 1991).

The results of the present study indicate that the polyethylene wear can be eliminated by using an alumina head, if no outside abrasive particles are involved. The behavior of Howmedica up to 35 million cycles was identical with the 2 preceding tests of 3 million cycles duration (Saikko et al. 1993): zero wear. In fact, the weight of the test cup increased faster than that of the control cup. This must be a consequence of the sliding motion, which is the only difference between the conditions of the test and control cups in the present simulator. The higher temperature of the test cup caused by frictional heat may explain the higher water absorption rate.

However, if an alumina head is attached to a titanium-alloy taper, severe wear of the cup can occur because of titanium-alloy particles removed from the taper, since the particles may be entrapped between the articulating surfaces, adhere to the head, render it rough, and thus severe abrasive wear of the soft polyethylene cup may follow as the asperity peaks of the head plow the very soft polyethylene surface. The height of the surface profile peaks of the Protek head could be several micrometers. This is sufficient to penetrate the lubricant squeeze film which, under ideal conditions, could separate the articulating surfaces during walking, at least for a short while after the heel strike, since the thickness of such a film has been calculated to be about one micrometer (Wang et al. 1990). Large amounts of polyethylene debris may be detrimental in vivo, leading to the loosening of the component fixation (Amstutz et al. 1992). It is true that the inverted position of the prostheses in the simulator, which is necessary to eliminate the risk of drying out of the articulation caused by air bubbles, facilitates the falling of the titanium-alloy particles from the taper between the articulating surfaces, but if such particles were generated also in vivo, their entrapment would certainly be possible.

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