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SYMPOSIUM: RETRIEVAL STUDIES

Constrained Cups Appear Incapable of Meeting the Demands of Revision THA

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Abstract

Background Failure rates of constrained cups for treating recurrent dislocation in revision THA range from 40% to 100%. Although constrained liners are intended to stabilize the hip by mechanically preventing dislocation, the resulting loss of range of motion may lead to impingement and, ultimately, implant failure.

Questions/purposes We therefore documented the mechanisms of failure of constrained acetabular cups in revision THA and determined the type and severity of damage (wear, fracture, and impingement) that occurs in situ.

Methods We retrieved 57 constrained components of four different designs at revision THA and examined for the presence of rim impingement, oxidation, cracks within the liner, backside wear, pitting, scratching, abrasion, burnishing, and the presence of embedded particles. Articular wear was calculated from the volume of the concave articular bearing surface, which was measured using the fluid displacement method.

Results Failure of the locking ring was responsible for 51% of failures, whereas 28% of revisions were the result of acetabular cup loosening, 6% backside wear, and 22% infection. Impingement damage of the rim of the polyethylene liner was seen in all retrievals with moderate or severe damage in 54%. The average volumetric wear rate of the articular surface was 95 mm^3/year.

Conclusions Failure of the locking liner ring and loosening of the acetabular cup are the primary causes of mechanical failure with constrained liners; polyethylene is an inadequate material for restricting motion of the hip to prevent instability. The durability of these devices is unlikely to improve unless the mechanical demands are modified through increased range of motion leading to less frequent rim impingement.

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Introduction

Dislocation after THA continues to challenge orthopaedic surgeons. Rates ranging from 0.3% to 10% have been reported after primary THA [12, 19, 48, 49] and 10% to 28% after revisions [7, 12, 24, 48, 49]. For the patient with persistent dislocation, recurrence rates after revision surgery range from 6% to 28% in most series [6, 17, 33–35, 45]. Within this context, a useful treatment option advocated by some authors has been implantation of an acetabular cup in which the femoral head is mechanically constrained within the polymeric liner [3, 29, 38, 46, 47]. Constructs of this type were originally restricted to one commercial design (the SROM cup; Joint Medical Products, Stamford, CT, USA) but have recently been produced by most implant manufacturers. The liner of the constrained cup extends beyond a hemisphere and has a mouth that is smaller than the mating head. Reduction of the head within the liner is achieved through mechanical expansion of the inner diameter of the rim; however, once the head is in place, an external metal ring is attached to the liner to prevent it from reexpanding, thereby maintaining the head within the articulation. Although this "overcoverage" of the head by the liner prevents dissociation of the joint in the majority of cases, an inevitable consequence is reduced ROM and a greater prevalence of impingement, especially with flexion and internal rotation [1–3, 8, 9, 51].

When impingement occurs through restriction of ROM, forces that would otherwise lead to dislocation are transferred to the rim and the shell of the constrained component. This generates impingement damage to the liner but may also cause overload of the prosthesis-bone interface, the liner-shell interface, or the locking mechanism of the femoral head [26]. Thus, the following modes of failure have been described by Yun et al. [51] in examining reports of cases involving revision of constrained acetabular components: (1) failure of fixation to the pelvis [2, 14, 20, 51]; (2) dissociation between the liner and the shell [1, 2, 14, 20, 51]; (3) biomaterial failure, defined as unintended wear or fracture of the liner or retaining ring [2, 8, 28, 39, 51]; and (4) separation of the femoral head from the constrained liner [2, 28, 51]. Beyond this classification of failure modes, specific information concerning the relationship between design features and the occurrence of failure has remained within the realm of speculation by clinical authors with no systematic analysis of retrieved components apart from the work of Shah and coworkers [43] who performed a retrieval analysis of tripolar acetabular cups of one design removed at revision surgery. Our systematic review of constrained components retrieved from revision surgery shows that attempts to limit joint motion using any of the designs evaluated often leads to damage to the polyethylene liner through impingement and wear. These damage mechanisms commonly occur in association with migration, abrasion, and even fracture of the constraining ring, typically leading to ultimate failure of the device through either recurrent dislocation or aseptic loosening.

In view of the observations reported at revision and the absence of more detailed information within the literature, we examined all constrained cups revised at our institution to: (1) document the mechanisms of failure of constrained acetabular cups; (2) report the type and severity of damage (wear, fracture, and impingement) observed in constrained components; and (3) evaluate the performance of the locking ring in maintaining constraint of the femoral head within the acetabular liner.

Materials and Methods

We examined a collection of 850 hip prostheses entered into an implant registry of the joint replacement service of a large teaching hospital. We identified 57 constrained liners from 51 patients (for each of six patients, two liners, used in sequential operations, were available) implanted between June 1993 and July 2010. Demographic details of the patients were obtained from the patients' office charts. Operative reports from the time of revision were available for 47 patients. Information retrieved included patient gender, age, affected side, body mass index (BMI), Charnley category [12], and time in situ of the implant (Table 1). There were 31 women and 25 men (one patient's gender was unknown) with a mean age of 61.0 years (range, 43–84 years). The median BMI was 27.0 kg/m^2 (range, 15.1–50.1 kg/m^2).

The revision procedures were performed by 14 different surgeons. The determination of the primary failure mechanism was based on the observations of the attending surgeon. Indications for use of the constrained liner included postoperative instability of primary or revision hip arthroplasty (18 liners, 13 patients), intraoperative instability during revision surgery (10 liners, 10 patients), absence or dysfunction of the abductor muscles of the hip (arising from nonunion of the greater trochanter or neurologic impairment, respectively; five liners, five patients), and patient noncompliance (one liner). Information on the indication for use of the constrained liner was unavailable for 23 liners.

The liners used for the present study were collected from the operating room through the pathology department as part of an ongoing retrieval analysis program. All implants were cleaned in a mild detergent bath, catalogued with an identification number (to ensure patient anonymity), and stored as part of a standard retrieval system. None of the components was autoclaved or otherwise sterilized between removal and examination. Most of the liners (n = 36 [63.2%]) were 32 mm in diameter.

Table 1. Demographic data of patients and details of implants

Component	Gender, age (years)	Side	BMI (kg/m^2)	Time in situ (months)	Rim (°)	Head diameter (mm)	Shell diameter (mm)	Charnley category
1	M, N/A	R	N/A	N/A	10	28	51	N/A
2	F, 82	L	Cachectic	12	10	28	50	C
3	F, 44	L	24.6	1	0	32	60	A
4	M, 43	L	34.1	6.5	0	32	N/A	A
5	F, 84	L	24.7	2.5	10	32	N/A	A
6	F, 84	L	24.7	60	10	32	N/A	A
7	M, 73	R	25.3	31	10	32	66	A
8	F, 80	L	20.3	30	10	32	N/A	C
9	F, 84	R	20.7	41	10	28	56	B
10	M, 65	R	27.3	45	10	32	60	B
11	M, 65	L	27.3	N/A	10	32	N/A	N/A
12	M, 69	R	27.3	31	10	32	63	B
13	M, N/A	L	N/A	N/A	10	32	N/A	N/A
14	F, 49	R	50.1	26	10	28	48	A
15	F, 50	R	50.1	17.5	10	28	48	A
16	F, 57	L	23.3	12.5	10	32	54	A
17	F, 57	R	23.3	N/A	10	32	N/A	A
18	F, 53	L	29.2	0.25	10	28	51	B
19	F, 50	R	29.1	91	0	28	48	B
20	F, 76	L	28.5	1	0	28	48	B
21	M, N/A	L	N/A	N/A	10	28	56	N/A
22	F, N/A	R	N/A	N/A	10	32	54	N/A
23	M, N/A	R	N/A	N/A	10	28	51	N/A
24	M, 49	L	22.0	48	10	32	54	A
25	M, 60	L	23.1	10	0	32	63	B
26	F, N/A	L	N/A	N/A	0	28	N/A	N/A
27	F, N/A	R	N/A	N/A	0	28	N/A	N/A
28	M, 54	L	N/A	38	10	32	60	A
29	N/A, N/A	L	N/A	N/A	10	32	N/A	N/A
30	F, N/A	R	N/A	N/A	0	28	51	N/A
31	F, 66	R	22.5	16	0	32	63	B
32	F, 66	L	22.5	25	0	32	N/A	B
33	F, N/A	L	N/A	N/A	10	32	N/A	N/A
34	F, N/A	L	N/A	N/A	10	28	60	N/A
35	F, 69	L	28.1	21	10	32	N/A	A
36	M, 80	R	35.6	19	10	32	60	A
37	M, N/A	R	N/A	N/A	0	32	58	N/A
38	M, 61	L	N/A	N/A	0	32	54	N/A
39	M,62	L	29.8	11	10	32	60	B
40	M, N/A	R	N/A	N/A	0	28	48	N/A
41	F, 68	R	21.6	25	0	32	N/A	B
42	F, 53	R	N/A	N/A	0	36	54	N/A
43	F, NA	R	N/A	N/A	10	32	50	N/A
44	M, 48	L	29.2	216	10	32	56	B
45	F, 77	R	15.1	N/A	0	32	N/A	C
46	F, 44	L	37.2	2	0	28	54	A
47	F, 44	R	37.2	2	0	32	N/A	N/A
48	M, 52	R	26.6	1	N/A	36	58	A

Table 1. continued

Component	Gender, age (years)	Side	BMI (kg/m^2)	Time in situ (months)	Rim (°)	Head diameter (mm)	Shell diameter (mm)	Charnley category
49	F, 64	L	22.2	1	10	32	N/A	C
50	M, 55	L	25.1	14	0	32	N/A	C
51	F, 47	L	30.1	84	0	28	52	N/A
52	F, 48	R	33.6	2	10	32	50	A
53	F, 75	L	22.3	1	0	32	N/A	B
54	F, 61	L	N/A	N/A	10	32	N/A	N/A
55	M, 55	L	27.0	N/A	0	38	65	A
56	F, 67	L	25	N/A	10	32	N/A	A
57	M, 75	R	26.8	156	0	32	65	B

BMI = body mass index; M = male; F = female; N/A = not available; R = right; L = left.

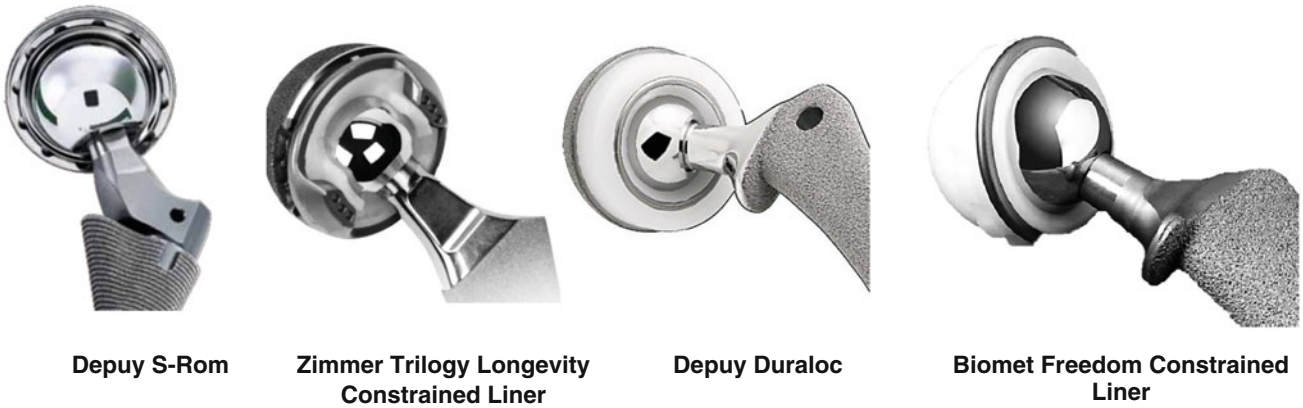

Fig. 1 The four different designs of constrained cups that were examined in this study are shown.

The 57 constrained components examined in this study were of four different designs (Fig. 1): (1) the S-ROM cup (n = 45; DePuy Orthopedics, Warsaw, IN, USA; Fig. 2); (2) the Trilogy Longevity Constrained Liner (Epsilon socket) (n = 8; Zimmer, Warsaw, IN, USA); (3) DePuy Duraloc (n = 3; DePuy Orthopedics); and (4) Biomet Freedom Constrained Liner (n = 1; Biomet Inc, Warsaw, IN, USA). Further details of each design are presented in Appendix 1. The retrieved components were examined by visual inspection before and after staining with India ink for the presence of damage resulting from wear, fracture, or delamination. Two of us (SD and NVB, neither a treating surgeon) examined the components by stereomicroscopy at ×5 to ×32 magnification using incident and transmitted light. All liners were examined for the presence of rim impingement, oxidation (as evidenced by the presence of discoloration and/or delamination), surface cracks, subsurface cracks (seen on transillumination), backside wear, pitting, coarse and fine scratching, abrasion, burnishing, and the presence of embedded particles using previously

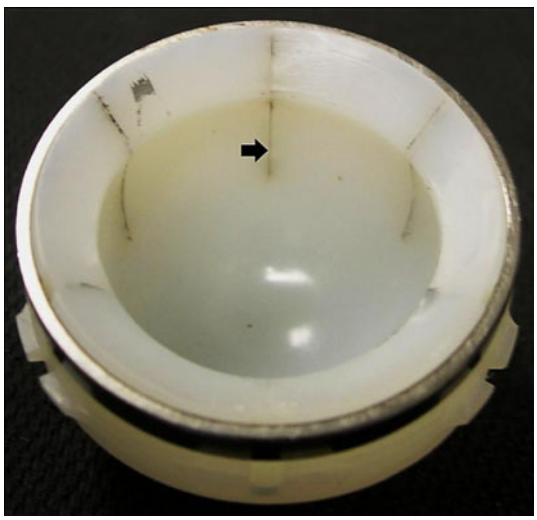

Fig. 2 A retrieved S-ROM constrained liner is shown with the titanium locking ring in place. The arrow points to one of the vertical slits, which, before snapping the metal ring, expand to allow reduction of the metal head. The liner shown has a flat rim.

published methods [5, 7, 22, 23, 39, 44]. Iatrogenic damage that appeared to have occurred during revision surgery was excluded from analysis.

Delamination was defined as subsurface cracking parallel to the surface. Rim impingement damage was defined by the formation of a blunted edge on the rim of the component as seen on visual and stereomicroscopic examination [21, 44]. The severity of impingement damage was classified as none, mild, moderate, or severe according to the 4-point grading system of Birman et al. [4].

All locking rings were carefully examined by stereomicroscopy to exclude damage resulting from intraoperative handling and disassembly, which, when present, was often restricted to discrete areas of the lower edge of the ring instead of the edge exposed to impingement. Damage to the metal locking ring was graded from 0 to 4 (0 = absent; 1 = mild, limited to light scratches; 2 = moderate, material loss involving mild blunting of the rim seen best at ×6 magnification; 3 = severe, gross material loss or ring deformation; 4 = ring breakage). Given the subjective nature of this rating scale, interobserver reliability for these measures was examined by four separate investigators with $\kappa = 0.73$ [23, 39].

Burnishing was defined as areas of the articular surface with a highly polished appearance [7] and was categorized as absent, present, or severe if it occupied more than 50% of the surface area. Abrasion was defined as the presence of areas with a roughened texture as a result of rubbing against a counterface [7] and was given a grade between 0 and 5 (0 = absent; 1 = subtle, seen only on magnification greater than ×10; 2 = confined to one to two spots; 3 = easily noted on magnification less than ×10; 4 = present in greater than 50% of the surface area; 5 = severe [11]). Given the subjective nature of this rating scale, interobserver reliability for these measures was examined by four separate investigators and rated as substantial for these measures ($\kappa = 0.67$) [25, 43]. Backside wear was graded on a similar 0 to 5 scale. The interobserver reliability of this method was reported as "substantial" (kappa coefficient = 0.66) [15] using the criteria of Landis and Koch [32]. Pitting was graded as absent, mild to moderate (less than 50% of surface area), and severe (greater than 50% of surface area) [7].

Articular wear was defined by the increase in volume of the concave articular bearing surface of the liner [51]. The volume of material lost was measured using the fluid displacement method [27]. The narrow expansion slits present in some of the designs were occluded to prevent leakage of the displaced fluid (Fig. 3). The fluid displacement method has been used in previous retrieval studies and has been shown to be accurate and reproducible in formal validation trials [13, 25, 37]. Annual volumetric wear rates were calculated from the measured volume between the head and the liner and the duration of implantation.

The type and severity of damage observed through inspection of the retrieved components were also reported using descriptive statistics. The presence of associations between qualitative variables was examined using the chi-square test for trend. The normality of the distributions of continuous variables (time in situ, annual volumetric wear rate) was assessed using normal plots. Both the time in situ and the volumetric wear rate of our components were not normally distributed and exhibited extreme leptokurtosis ($g > 11$) [20]. Therefore, associations involving these variables were assessed using the Mann-Whitney U test. Correlations between other variables (impingement severity, time in situ, annual volumetric wear rate) were examined using Spearman's rank order correlation (rho [ρ]) coefficient (two-tailed). Statistical analyses were performed with IBM SPSS Statistics Version 19.0 software (IBM Corporation, Armonk, NY, USA).

Results

The mean time in situ of the components examined was 36.0 months (range, 0.25–216 months). Operative reports from the time of revision were available for 47 patients. Of these, 11 [23%] revisions were performed for infection and

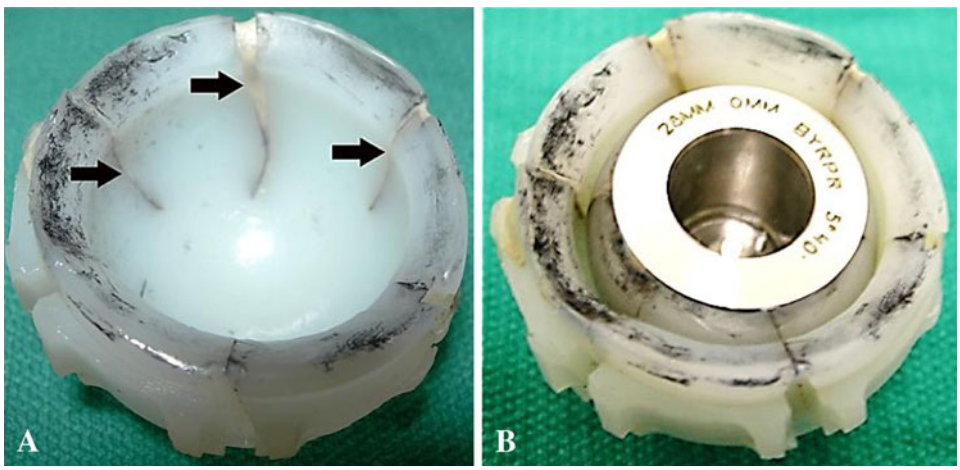

Fig. 3A–B Measurement of volumetric wear using the fluid displacement method is shown. The specimen with macroscopically open slits in the articular surface that have been occluded (arrows) to prevent fluid leakage (**A**). The same liner with the corresponding head (**B**).

two [4%] for metallosis. The most frequent cause of revision was redislocation, which occurred in 24 (51%) patients, four [9%] secondary to fracture of the retaining ring. In another 13 cases (28%), revision was performed to treat loosening of the acetabular shell. Disassociation of the liner from the shell occurred in three patients (6%). The average duration of implantation varied with the cause of failure, from 1.9 years for locking ring failure, 2.0 years for infection, 4.1 years for shell/liner disassociation, and 5.9 years for shell loosening.

The median wear rate of all components measured was 86.8 mm^3/year (mean, 314.0 mm^3/year) with large variations between cases. After exclusion of all liners that had been implanted for less than 6 months, the mean annual volumetric wear rate was 95.2 mm^3/year (95% CI, 66.4–124.0; range, 0–322). The volumetric wear rate decreased with longer periods of implantation ($\rho = -0.369$, $p = 0.003$). Backside wear was noted in 54 liners (95%) and was graded as medium or severe in 13 cases (23%). Surface cracks were observed in 26 (46%) liners. Transillumination revealed subsurface cracking in only four (7%) liners. We found embedded particles of metal and/or cement in 17 (33%) liners. A majority ($n = 44$ [77%]) had some degree of abrasion, which was a universal finding in all liners with embedded particles (Fig. 4). Overall, liners with embedded particles were more likely ($p < 0.001$) to have abrasion of increased severity. A majority of components (36 [63%]) had mild to severe burnishing, whereas 11 (19%) had some areas of pitting. Twenty (35%) liners demonstrated evidence of oxidation, which was moderate or severe in all but one case (33%). All polyethylene liners demonstrated some evidence of rim impingement, which was rated as mild, moderate, or severe in 26 (46%), 22 (39%), and nine (16%) liners, respectively. The severity of impingement damage increased with both the volumetric wear rate ($\rho = 0.356$, $p = 0.0325$) and the time in situ ($p = 0.044$). The prevalence of each mode of damage was similar in 28-mm liners compared with their 32-mm counterparts (Table 2).

All but three of 49 retrieved rings (94%) had some degree of damage (ie, damage score = 1 or greater) consistent with impingement between the ring and the neck of the femoral component. Half of the rings were classified as mildly damaged (25 of 49 [51%]) followed by lesser proportions of moderately (10 of 49 [20%]) or severely damaged components (six of 49 [12%]). Five rings (10%) had transverse fractures, all subsequent to material loss from impingement (Fig. 5).

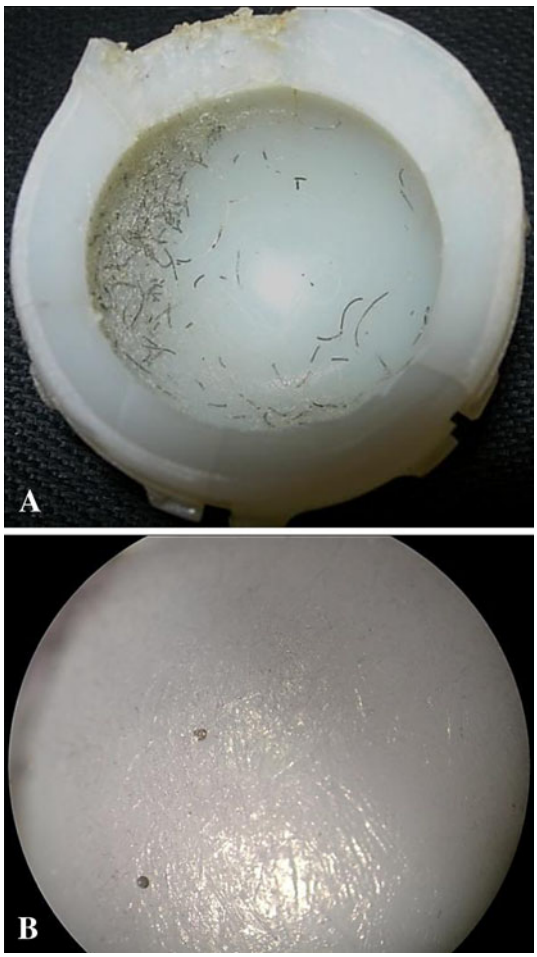

Fig. 4A–B Macroscopic view is shown of a 32-mm S-ROM constrained liner with numerous embedded metal particles (**A**). Appearance is shown under the microscope of isolated embedded particles and abrasion of the adjacent surface of a different S-ROM specimen (**B**).

Discussion

Prevention of instability of revision THA is a challenging problem, especially in cases of recurrent dislocation. One strategy has been to use a constrained acetabular cup, which physically closes the acetabular insert over the femoral head after intraoperative reduction. Although this may prevent instability of the articulation, the ROM of the implant is potentially compromised. This is expected to increase the risk of impingement and the loading of the prosthesis and may simply shift the mechanism of failure from dislocation to loosening or wear. In view of these concerns, we undertook a systematic retrieval study of all constrained cups revised at our institution to document the mechanisms of failure of constrained components and the type and severity of any damage (wear, fracture, and impingement) occurring secondary to function. In view of the unique design of constrained components, a secondary objective was to evaluate the performance of the locking mechanism in constraining the femoral head within the acetabular liner during the life of the component.

The authors acknowledge the limitations in the design of the present study. First, like with most retrieval studies, many of the conclusions reached were based on subjective observations. To make our conclusions more rigorous, indicators of damage to the retrieved components were explicitly defined and then graded according to an incremental scale based on objective criteria. To test the reliability of these methods, we also performed formal interobserver trials and calculated kappa values. In each case, the kappa value has been within the range of "substantial" reliability (0.61–0.80). Second, we had a limited number of retrievals for study. This means that the conclusions drawn from the finite population of retrievals examined in this study, although reliable in describing our sample, may not be broadly generalizable to all conditions and populations in the future. Third, and on a similar note, the majority of retrieved components in this study were of one specific design, the SROM. This means that the conclusions of the study may well be applicable to SROM cases but may be less generalizable to other designs given differences between components and the relatively small numbers of non-SROM components. Fourth, the displacement method of measuring changes in the internal volume of acetabular cups is based on the assumption that the initial clearance between the head and the liner is the same for each prosthesis. In reality, manufacturing tolerances introduce variability in the initial volume of the space between the articulating components. Individual values of this starting volume may be calculated if spatial coordinates defining both surfaces are measured rather than the total volume of the worn bearing component. Finally, it is preferable to assess wear rates using multiple measurements performed over a series of observations over the life of the implant [43]; however, because many revision patients are referred to our tertiary referral center from surgeons working elsewhere, this is often not possible.

Table 2. Comparison of the prevalence of wear features between 28- and 32-mm liners

Wear feature	28-mm liners (n = 17)	32-mm liners (n = 36)	p value
Impingement[‡]	10 (58.8%)	20 (55.6%)	0.82*
Oxidation	6 (35.3%)	14 (38.9%)	0.80*
Surface cracks	12 (70.6%)	17 (47.2%)	0.11*
Subsurface cracks	2 (11.8%)	2 (5.6%)	0.42[†]
Backside wear[§]	10 (62.5%)	29(80.6%)	0.09[†]
Pitting	2 (11.8%)	8 (22.2%)	0.36[†]
Coarse scratching	13 (76.5%)	28 (77.8%)	0.33[†]
Fine scratching	11 (58.8%)	26 (72.2%)	0.27[†]
Abrasion[‖]	14 (82.4%)	22 (61.1%)	0.88*
Burnishing	10 (58.8%)	26 (72.2%)	0.33[†]
Embedded particles	5 (29.4%)	12 (33.3%)	0.78[†]

Each wear feature was assigned as being absent or present, except when designated otherwise; * chi-square test with Yates; correction; [†]Fisher's exact test; [‡]assigned dichotomous variables (mild versus moderate/severe) for this statistical analysis; [§]assigned dichotomous variables (absent/light scratches only/barely discernible versus mild/moderate/severe) for this statistical analysis; [‖]assigned dichotomous variables (absent/subtle/spots versus present/widespread/severe) for this statistical analysis.

In undertaking this study, our first objective was to document the mechanisms of failure of constrained acetabular cups. The most common cause of revision was redislocation, which occurred in 24 (47%) patients, four [9%] secondary to fracture of the retaining ring. This incidence is higher than that reported by Berend et al. [2] (29%) and Yang and Goodman [50] (6%). Other common causes were loosening of the shell within the acetabulum (28%), which occurred less frequently than reported by Yang and Goodman [50] (45%) but was more common than observed by Berend et al. [2] (6%). These variations are no doubt the result of differences between each study, particularly the indications for use of a constrained component and the proportion of patients with a history of recurrent dislocation and multiaxial instability.

Despite relatively early failure of these components, we observed multiple modes of polyethylene damage, including moderate/severe creep and material loss secondary to

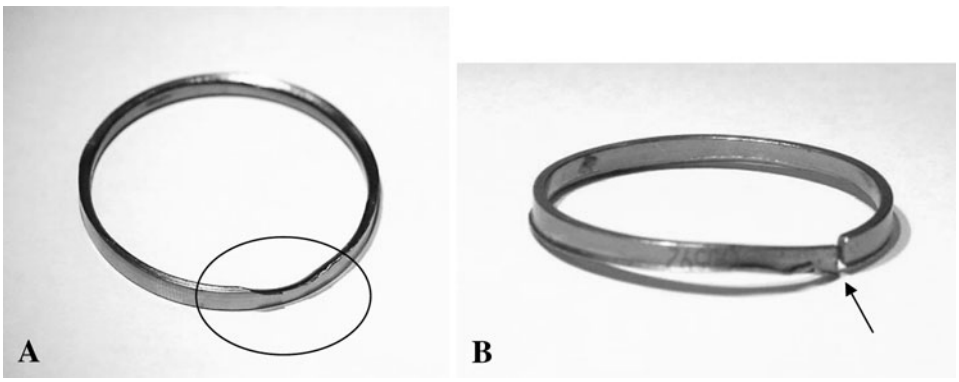

Fig. 5A–B Macroscopic view of damaged locking rings from S-ROM cups is shown. (**A**) Dramatic material loss (circle) caused by impingement against the femoral neck; damage was classified as severe. (**B**) Ring breakage (arrow) with material loss and gross deformation.

impingement (55%), surface cracking (46%), embedded particles (33%), abrasion (80%), oxidation (36%), burnishing (64%), pitting (20%), and backside wear (moderate or severe in 26%). These findings are similar to those of Shah et al. [43] who examined components of the Osteonics tripolar cup design after retrieval at revision surgery. However, our results highlight the fact that constrained cups with a single articulation, in contrast to the tripolar (ie, biarticular) design, are subject to large stresses primarily through prosthetic impingement. This leads to a remarkable incidence of deformation and cracking of the liner, even in components with highly crosslinked polyethylene bearings.

The average wear rate of our retrieved specimens was 95 mm^3/year, approximately one-third greater than more contemporary studies of wear in primary THA, which report values of 55 to 71 mm^3/year for conventional polyethylene and 14 to 17 mm^3/year for highly crosslinked polyethylene using 28-mm and 32-mm heads [4, 10, 11, 22, 27]. This would appear to support the claim of some investigators that the wear rate of constrained cups is higher than unconstrained devices of the same head diameter [1, 2, 18, 31]. However, wear rates of components retrieved at revision THA tend to be higher than those derived from primary cases, with average values of 94 and 99 mm^3/year reported in previous series [20, 48]. This increase may be the result of the occurrence of repetitive microseparation with rim impingement and the presence of third-body particles, either within the joint fluid or embedded within the articular surface [36, 41].

One of the novel features of our study is the detailed analysis of the performance of the locking ring in maintaining constraint of the femoral head within the acetabular liner. In this series, overt failure of the locking ring with dislodgement or breakage occurred in 51% of all revisions and 60% of mechanical failures. Although locking ring fractures have been the subject of isolated case reports [8, 28, 42] and infrequent complications of clinical series [2, 14, 40, 43], this study is the first to recognize that wear of metallic retaining rings is a frequent consequence of impingement, occasionally progressing to fracture. The abrasive damage of the retaining rings examined in this study occurred despite the presence of an interposed layer of polyethylene separating the inner surface of the ring from the neck of the femoral component. In service, the femoral neck of the hip prosthesis wore through the polyethylene layer through repeated impingement, or the ring migrated to a position in which direct contact with the neck became possible. In either case, direct contact between the ring and the neck ensued with metallic transfer and generating of wear debris.

In conclusion, our observations show constrained acetabular cups are subject to large contact forces in service, making them vulnerable to multiple wear mechanisms. This supports the general conclusion [1, 2, 40, 51] that the design strategy of preventing dislocation of an artificial articulation by limiting motion is likely to lead to a high incidence of failure, whether by redislocation, mechanical failure or aseptic loosening, or some combination of all three. The devices examined in this study demonstrate that prosthetic impingement is inevitable in many constrained devices and that the polyethylene is mechanically inadequate to absorb the repetitive impact that this generates without ultimate failure. In view of these findings, we recommend other implant options, including a tripolar cup [16], be explored for treatment of recurrent dislocation.

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Appendix 1: Devices examined in this study

The constrained cups examined in this study were of four different designs (Fig. 1):

(1) The S-ROM cup (DePuy Orthopedics, a Johnson & Johnson Company, Warsaw, IN, USA; formerly Joint Medical Products, Stamford, CT, USA). In this design, the additional constraint is afforded by the presence of an extended polyethylene rim, which, through the presence of six slits, deforms to accept the femoral head. A titanium alloy locking ring is assembled over the rim once the head is reduced to prevent reexpansion and dislocation. The liner is manufactured from conventional UHMWPE with a minimum thickness of 5 mm and fits shells of 48 to 66 mm (28-mm liner) and 54 to 75 mm (32-mm liner) outer diameter [23]. The constraining force of the ring is greater than 600 lbs for the 32-mm liner and 325 lbs for the 28-mm liner [50]. The leveraged torque to disengage the femoral head was 150 inch-lbs [30].

(2) The Trilogy Longevity Constrained Liner (Epsilon socket) (Zimmer, Warsaw, IN, USA) was introduced in 2003 and can be locked into any Trilogy cup or cemented into the acetabulum directly. The face of the liner consists of two elevated segments with contiguous cutouts located anterosuperiorly and posteroinferiorly. The cutouts are designed to reduce impingement in full flexion with internal rotation and in extension with external rotation as well as to maximize ROM. All Longevity liners use the highly crosslinked polyethylene (Durasul). Twenty-eight-millimeter, 32-mm, and 36-mm liners are available for implantation with a ROM of 115°, 120°, and 125°, respectively, according to the product data sheet. Proprietary testing

determined the lever-out strength is 235 inch-lbs. A titanium alloy constraining ring is mounted on the face of the liner to hold its elevated fingers in a closed position. Pegs on the constraining ring lock into slots on the periphery of the liner with impaction.

(3) DePuy Duraloc (DePuy Orthopedics). The UHMWPE (Enduron) liner has a minimum thickness of 6 mm and is compatible with either the Duraloc or Solution acetabular shells [47]. Reducing the femoral head requires 75 lbs of pressure, after which the titanium alloy ring is secured into a circumferential groove on the liner face. The 28-mm diameter liner has a pullout strength of 416 lbs and a lever-out strength of 170 inch-lbs [50].

(4) Biomet Freedom Constrained Liner (Biomet Inc, Warsaw, IN, USA), a recent design using compression-molded ArCom polyethylene. The 36-mm femoral head has circumferential flats, which can counteract distractive forces leading to dislocation. Compared with older designs, there is an increased ROM before impingement to 110° and lever-out strength of 198 inch-lbs [50].

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