Ceramic on ceramic bearings in hip arthroplasty: State of the art and the future

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Abstract

This systematic review of the literature summarises the clinical experience with ceramic on ceramic hip bearings over the past 40 years and discusses the concerns that exist in relation to the bearing combination. Loosening, fracture, liner chipping on insertion, liner canting and dissociation, edge loading and squeaking have all been reported in the literature, and the relationship between these issues and implant design and surgical technique is investigated. New design concepts are introduced and analysed with respect to previous clinical experience.

Introduction

Ceramic on ceramic (CoC) bearings in hip replacement were developed by Boutin, Griss and Mittelmeier in the early 1970's in France and Germany and the earliest designs consisted of a bulk alumina acetabular cup that was fixed in the acetabulum either by cement or with a press-fit.[1-3] Fixation ultimately proved to be insufficient and the predominant cause of failure was aseptic loosening. High fracture rates were also reported. Throughout the 1980's, the Autophor (or Mittelmeier) ceramic prosthesis was widely used with follow-up studies published in Europe, North America and Asia.[4-9] As with previous designs, the Autophor design was a bulk alumina acetabular cup, but had a threaded external surface to fix in the acetabulum without cement. This gave primary stability but there was no porous surface to facilitate secondary bony ingrowth. The design did not improve the rate of aseptic loosening, but the fracture rate was notably reduced due to improvements with material manufacture. An important finding upon revision of these devices was that although ceramic debris was observed within macrophages in the local tissue, evidence of periprosthetic osteolysis was markedly reduced compared to metal on polyethylene bearings of the same era. Since the early 1990's the dominant design has been a rough or porous coated titanium shell with an alumina liner. This design has generated excellent implant survival and patient satisfaction results, but difficulties in assembling the acetabular cup intraoperatively have been reported in a small number of cases. Edge loading is the dominant wear mechanism observed but has little clinical significance apart from occasional noise generation. Squeaking of the bearing has been reported: benign occasional squeaking is not normally a clinical problem and severe squeaking with every step of walking is rare.

The great success of contemporary CoC bearings is leading manufacturers to continue pushing the limits of the technology, particularly in the wake of the recent MHRA guidance on metal on metal (MoM) bearings.[10] Thin walled (≤5 mm) ceramic acetabular cups are already on the market, and several manufacturers are currently developing CoC resurfacing devices with large diameters (38 mm – 58 mm). The aim of this article is to identify the clinical and engineering issues reported in four decades of experience with CoC hips, and relate them to these emerging technologies. This will enable surgeons to challenge the industry to explain how the risks have been mitigated during product development when presented with these new ceramic devices.

Methods

The Pubmed database (www.ncbi.nlm.nih.gov/pubmed) was used on 10 May 2011 to identify relevant literature by searching for the following terms in the title, abstract or keyword fields: “ceramic on ceramic” AND “hip” (n=193), “ceramic ceramic” AND “hip” (n=62), “alumina on alumina” AND “hip” (n=83), “alumina alumina” AND “hip” (n=34), “zirconia on zirconia” AND “hip” (n=0), “zirconia zirconia” AND “hip” (n=0), “zirconia on alumina” AND “hip” (n=0), “zirconia alumina” AND “hip” (n=4), “alumina on zirconia” AND “hip” (n=0), “alumina zirconia” AND “hip” (n=11). Articles were discarded if they were not clinical studies, reported on less than 10 patients or were not written in the English language, leaving a total of 74 studies. The references of these articles were reviewed and identified a further 14 articles not found in the initial search, giving a total of 82 articles. Full manuscripts were obtained and reviewed. Articles were discarded if they were summaries of national registries or other national database, or if they includes several implant designs and it was not clear which design any complications referred to. In cases where the same authors presented the results of the same patients at different follow up periods, only the most recent article was included. This reduced the number of articles included to 54.

A subsequent search was performed to identify literature relating to the use of ceramics in hip resurfacing by searching for the following terms in the title, abstract or keyword fields: “surface replacement” AND “ceramic” (n=6), “ceramic” AND “hip” AND “resurfacing” (n=23), “double cup” and “ceramic” (n=5). Neglecting duplicates and non-clinical studies left 3 articles for review. Searching the references of these manuscripts identified a further 2 articles, thus increasing the total to 5.

Observations during the past four decades

The studies included in the review are summarised in Tables 1-5. The following were reported: loosening, fracture, liner chipping on insertion, liner canting and dissociation, edge loading and squeaking.

Aseptic loosening

Although high rates of aseptic loosening were reported for CoC bearings in the 1970’s and 1980’s (Tables 1 & 2), these were due the fixation method of the components and not a biological reaction to the ceramic wear debris.[1,3,8,11] Indeed, the clinical data indicates the CoC bearing couple has no adverse reaction to wear debris at 10-20 years (Table 1 & 2). Histological examination of long term retrievals has identified ceramic wear debris within individual macrophages in periprosthetic tissue, but the limited quantity or relative inertness of the debris does not generate the foreign body granuloma necessary to trigger the osteolytic reaction. One study reports a granulomatous reaction and presence of giant cells in the periprosthetic tissue of CoC bearings at revision,[12] but these findings have not been repeated elsewhere. Considering contemporary designs and materials, radiographic analysis has shown a significantly lower rate of osteolysis with CoC bearings compared to MoP bearings (1.4% vs. 30.5% respectively) at 8 years.[13]

CoC bearings are the only combination that have long term (10-20 years) survival without adverse reaction to wear debris. The use of MoM bearings has declined rapidly in the past few years due to concerns about metal debris, particularly in women, and metal or ceramic on polyethylene bearings are associated with osteolysis and loosening in the long term. Cross linked polyethylene is showing encouraging mid-term results,[14] but whether these will continue in the long term is unknown. Vitamin E polyethylene has demonstrated excellent wear results in the laboratory,[15] but must be considered experimental until clinical data appears in the literature to validate the preclinical testing.

Ceramic fracture

Contemporary alumina material is very different to that associated with the high fracture rates in the 1970's (Table 6). The introduction of improved raw materials and hot isostatic pressing during manufacture served to reduce the grain size and increase the density of the material, which improved its mechanical properties.[16] Further improvement in mechanical properties was achieved in the 2000's with the introduction of alumina/zirconia composite materials. Zirconia occupies a monoclinic crystalline structure at room temperature, but changes to a smaller volume tetragonal structure at temperatures greater than about 1100°C. It is possible to maintain the tetragonal structure at room temperature within the alumina matrix by stabilising with yttria. In the presence of a crack, the restraint on the zirconia crystalline structure is removed and the zirconia transforms back to the larger volume monoclinic structure which generates compressive stress and retards the crack growth. Another mechanism to dissipate the energy of a crack in contemporary material is the addition of strontium oxide which forms long crystals (platelets) of strontium aluminate in the alumina matrix. These deflect the crack and increase the distance it must travel to progress through the material, thus increasing the energy required for it to propagate. Figure 1 shows a micrograph of a contemporary transformation toughened ceramic (Biolox Delta, Ceramtec AG, Plochingen, Germany), identifying the alumina matrix, zirconia crystals and strontium aluminate platelets. The rate of fracture of femoral heads with this material is reported by the manufacturer to be 1 in 50,000.[17] Only one study was found that reported clinical data of transformation toughened ceramic in a CoC bearing, and the results were similar to alumina on alumina bearings (Table 4). There may be some concerns that the zirconia phase in the material may become unstable in an aqueous environment at body temperature as was observed for zirconia-yttria femoral heads in the 1980's and 1990's. However, over 600,000 CoC bearings have been implanted using alumina/zirconia composite ceramic with no adverse reports to date regarding the stability of the zirconia phase. Indeed, laboratory testing under accelerated ageing conditions has demonstrated no measurable difference in terms of surface roughness or strength of the alumina/zirconia composite material at a simulated 40 years *in vivo*. Under the same conditions, the zirconia-yttria ceramic had a 25x fold increase in surface roughness.[18]

When a ceramic part is manufactured, there is a chance that a flaw exists within the part that can lead to fracture – even with contemporary ceramics. To identify components with flaws in critical areas, proof testing is performed as a destructive inspection test. A proof test uses mechanical methods to generate a stress state in the part that represents a more severe scenario than could reasonably be expected *in vivo*, and, if any critical flaws are present in regions of high stress, the part is automatically destroyed. During surgery, the component can be placed in any orientation about its axis and the proof test must therefore generate the stress state axisymmetrically. Ceramic parts in hip replacement and hip resurfacing are usually dome shaped, and therefore very efficient at transferring axisymmetric loads. Mechanical proof loads greater than 50 kN (60x bodyweight) are commonly required to generate an axisymmetric stress state in the component equivalent to a non axisymmetric *in vivo* stumbling load. Caution must therefore be exercised when interpreting the impressively high loads ceramic parts can withstand, as they may represent axisymmetric loads that are not encountered *in vivo*.

Ceramics are vulnerable to point loading which can occur if there is debris at the taper mating surfaces. This is equally applicable at the femoral head/stem interface and the acetabular shell/ liner interface, and can lead to fractures in both scenarios.[19] Clean assembly of the components is therefore important, but sometimes difficult to achieve during surgery. Failure to engage the tapers of the titanium shell and ceramic liner properly may also be responsible for fracture of the liner. A liner seated in a canted position will generate a two point support for the liner with an adverse stress distribution in the ceramic, while a liner that is not impacted properly may dissociate from the shell due to the suction force that can be generated by the lubricated bearing[20] and re-seat in a canted position. Both scenarios also provide a pathway for debris to the taper interface that may cause point loading of the ceramic. It is important to note that proof testing assumes a perfect assembly and does not consider these scenarios of point loading.

Liner chipping on insertion

Failure to seat the ceramic liner fully in the shell prior to impaction may increase the risk of rim chipping. As has been noted for liner canting, the titanium shell may deform by 0.6 mm during impaction such that when the ceramic liner is placed it gets supported at two diametrically opposing points only. Careful trialling of the prepared acetabular socket is therefore important, and some groups recommend reaming line to line in cases of sclerotic or dense acetabular bone, but this is highly dependent on implant design. No cases of ceramic chipping on insertion were noted for monoblock devices implanted in the 1970's and 1980's.

Liner chipping during insertion was a particular complication noted by D'Antonio, Capello and colleagues in their landmark prospective randomised study of ceramic and polyethylene bearings.[21] This may have been related to the design of the taper which, rather than continuing to the cup edge, transformed to a cylinder about 6mm from the rim of the cup and may have increased the risk of impacting a malseated liner. To protect the ceramic during assembly, the manufacturer encased the liner in a titanium sleeve that sat proud of the ceramic rim (Figure 2). This eliminated liner chipping as a complication, but may have introduced other complications associated with squeaking and liner malseating.

Liner canting or dissociation

Liner canting (malseating) or dissociation has been reported for surgery performed in the 1980's,[22] 1990's[23,24] and 2000's,[25-27]. No adverse events have been directly related to liner canting or dissociation, but it has been suggested that an uncoupled liner could increase the risk of squeaking[28], fretting corrosion of the titanium surface[26] and fracture of the liner.[29]

A normal force across the interface between the titanium shell and ceramic liner is required to maintain the static friction force that keeps the components assembled. The titanium shell has a thinner wall thickness than the ceramic, and material stiffness approximately 30% that of ceramic. Impacting the ceramic liner into the shell therefore expands the titanium shell and generates circumferential tensile stress in the titanium. The titanium shell then acts like an elastic band on the liner and generates circumferential compressive stress in the ceramic and a uniform normal force across the interface. Appropriate surgical technique is essential to properly assemble the cup, but this is not always achieved and a number of factors may contribute to this.

Intra-operative measurements indicate the titanium shell can deform by 0.6mm diametrically,[30] which is sufficient to limit ceramic liner contact to two diametrically opposing areas. Reduced shell thickness, increased bone stiffness and increased interference fit all act to increase the shell deformation. Soft tissue entrapment, bone or HA fragments are other possible mechanisms that may prevent uniform seating and generate non-uniform loading of the ceramic liner.

Failure to impact the liner with sufficient force during assembly may also contribute to canting and dissociation. A suction force of up to 30 N (3 kg) can be generated by a lubricated bearing when the head separates from the cup.[20] This force acts against the static friction that keeps the cup assembled.

Implant design also is a factor in liner canting, with a notably higher rate reported for the Trident device (Stryker, Mahwah, USA).[25-27] A particular feature of this device is that the ceramic liner is pre-assembled in a titanium sleeve which is then fitted to the titanium shell intraoperatively. The rim of the titanium sleeve sits proud of the outer shell and full seating of the device can therefore be difficult to check in the inferio-medial aspect. A further design aspect that may be relevant in liner canting is the mating taper angle. The most common liner taper angle is 19°, but the Trident device has a 10° taper angle. The smaller taper angle generates a smaller window of insertion for which the taper will engage, and may be related to the increased incidence of canting. Increasing the taper angle may ease the liner insertion, but generating the required normal force at the interface for static friction to keep the assembly together may be difficult to achieve.

Edge loading

The finished bearing surface of a ceramic liner is ground and polished to specification after sintering (firing). To account for manufacturing tolerances during sintering (where parts shrink by approximately 10-20%), a small lead-in surface is designed into the pre-machined component and a hard edge is created where the lead-in surface intersects the ground and polished bearing surface (see Figure 3). The edge of the bearing surface therefore sits a couple of millimetres in from the face of the cup, effectively reducing its included angle, and this can be seen on ceramic components as a discontinuity in reflected light (Figure 3). When the hip contact force vector moves over this hard edge (edge loading), the contact stress increases and both surfaces are damaged. The long narrow pattern of roughened damaged surface on the head is referred to as stripe wear.

Stripe wear provides the necessary evidence to understand the mechanisms of edge loading. In the 1970's, Boutin proposed edge loading was due to steeply positioned acetabular cups reducing the contact area such that a greater proportion of load is transmitted through the edge of the cup.[1] Separation during the swing phase of gait followed by the head crashing into the rim at heel strike has also been proposed as a mechanism in cases where laxity exists at the joint.[31] However, measurement of the location and orientation of stripe wear has indicated the most common edge loading mechanism is subluxation of the head over the hard posterior edge of the bearing surface during deep flexion.[32] In severe cases there is evidence of impingement roughly diametrically opposite the location of the wear scar on the acetabular cup,[33] and severe edge loading is certainly a function of poor acetabular cup position. More often there is no evidence of prosthetic neck to prosthetic rim impingement and indeed posterior edge loading is seen commonly on well-positioned and well-functioning retrieved components indicating that, for the implant designs reported in Tables 1-4, edge loading is not a clinical problem and probably part of normal hip function.

When a normal hip edge loads the native head does not see a dramatic increase in local stress due to the gradual change in hardness/stiffness from the articular cartilage to labrum to capsule under tension, however, when a CoC bearing edge loads there is a dramatic increase in local stress producing the stripe. In total hip replacement edge loading cannot be avoided unless there is hip stiffness, in particular a loss of hip flexion. Even if edge loading does occur, the wear volumes generated under CoC edge loading are so small that wear debris rarely causes an osteolytic reaction. This is a major advantage compared to MoM bearings, where edge loading is associated with run-away wear of the bearing surfaces.

Squeaking

Ceramic squeaking has received much attention over the past few years, but the literature indicates it is not a major clinical problem and can often be avoided by activity modification alone.[34] However, in the rare cases that squeaking is persistent, for example occurring with every step of gait, revisions have been performed and it makes sense to understand the phenomenon.

Audible squeaking requires an impulse and amplification. The impulse at the bearing surface is caused by frictional conditions and the amplification is vibration of the components.

Impulse generation

The vibrations that initiate squeaking are local intermittent motions caused by stick-slip friction. A static friction force exists at the bearing surface that opposes rotation. The rotational force overcomes static friction, and because the kinetic friction force is less than the static friction force, the local surface accelerates to a speed greater than overall rotation. This causes the local rotational force to reduce below the static frictional force and the local surface slows to a halt. This localised acceleration/deceleration is manifest as vibration within the material, and laboratory data indicate the acceleration can be up to 450 m/s^2 (46 times the acceleration due to gravity).[35]

The friction force of a virgin CoC bearing under optimal lubrication conditions is too low to initiate stick-slip. Several mechanisms have been identified that can increase the friction at the bearing interface, most of which are related to edge loading. The lack of lubrication during edge loading, combined with the damaged bearing surface and higher contact stress may be sufficient to increase friction such that the stick-slip conditions are generated, but other mechanisms have been proposed. Laboratory data indicates that ceramic debris between the bearing surfaces,[36] metal transfer to the head[37] and starvation of lubricant without edge loading[35,37] can all increase friction at the bearing.

Clinical observations have noted relationships between squeaking and patient characteristics, surgical technique and implant design. Heavier and more active patients have an increased risk of squeaking due to increased mechanical loading.[38] Component orientations associated with impingement or edge loading have been associated with squeaking, thus highlighting again the importance of cup positioning.[38] It follows that an implant design that increases the risk of impingement or edge loading should be associated with higher than normal rates of squeaking, and indeed this has been reported for the Trident device.[39] The elevated titanium rim of this device (see Figure 2) reduces the available range of motion, thus increasing the risk of neck to rim impingement and subsequent edge loading.

Vibration of components

Clinical studies have found significantly increased rates of squeaking when particular components are used together. For example, the Trident acetabular cup has a significantly higher rate of squeaking when paired with an Accolade stem compared to an Omnifit stem (both Stryker, Mahwah, USA), possibly due to the thinner neck section and lower natural frequency of the former.[40] The natural resonant frequency of the CoC hip components have been analysed, and only the titanium shell and titanium stem have a natural frequency in the audible range.[28] The natural frequency of the ceramic components is above the human threshold, but it may be possible for them to vibrate at lower than natural frequencies in the audible range because of stick-slip frictional mechanism described above. In this case, the frequency of the vibration depends on the frictional characteristics of the bearing system, including the friction, clearance and rotational speed, rather than the natural frequency of the parts. Laboratory data has demonstrated audible squeaking for an unlubricated CoC bearing without metallic parts.[35] If this lower frequency ceramic vibration coincided with the natural frequency of a particular stem, as observed for the Trident cup with Accolade stem, resonance may occur thus amplifying the sound.

The future of CoC bearings & potential risks for new devices

Considering THA, the DeltaMotion device (developed by Finsbury Orthopaedics, Leatherhead, UK, now manufactured by DePuy, Leeds, UK) is preassembled by the manufacturer and seems to overcome several of the concerns relating to the intra-operative assembly including chipping on insertion and liner canting (Figure 4). This also allows for a 5 mm wall thickness such that a 46 mm cup can take a 36mm bearing. A drawback of the preassembled design is the inability to use screws to achieve stability and the increased difficulty to achieve primary stability compared to a thin titanium shell. The device is more like a metal resurfacing cup in this regard, but these have achieved satisfactory long term fixation. The development of thin walled acetabular cups have also allowed for larger femoral heads and sizes up to 48 mm are now available from at least two different manufacturers. The theoretical benefits of large diameters in THA (>36 mm) are improved range of motion, reduced risk of dislocation and perhaps improved patient proprioception, but it may be difficult or even impossible to prove these benefits in a clinical setting. The issue of bearing diameter is likely to be debated in the orthopaedic community for some time.

CoC hip resurfacings are currently being developed by several manufacturers and necessarily have large diameter bearings, but the use of ceramic in hip resurfacing is not new (Table 5). CoC resurfacing was tried in the 1970's and relied on a press fit with the acetabular and femoral bone for fixation. As with similar total hip replacement designs of the era, this proved inadequate and high rates of aseptic loosening were reported. In the 1970's and 1980's, the ceramic resurfacing head was paired with a polyethylene acetabular cup. The large articulating surface area generated high levels of polyethylene wear debris which led to osteolysis and high rates of aseptic loosening. No fractures of the ceramic resurfacing head are reported in the literature, but the numbers reported are too small to make any conclusions.

Contemporary composite alumina/zirconia ceramics have superior strength to the material used in the 1970's and 1980's, and preliminary testing of a contemporary ceramic resurfacing head indicates the material is strong enough for a resurfacing application with a generous factor of safety.[41] Considering ceramic acetabular cups with 5 mm wall thickness are already available and that Harlan Amstutz was performing hip resurfacings with a 5 mm thick acetabular cup until 2003[42] indicates CoC resurfacing is certainly possible with current technology.

Although contemporary materials may be strong enough for thin walled CoC acetabular cups and resurfacing designs, the clinical data identifies risks other than component fracture. We know from the literature that edge loading causes stripe wear, and this is related to increased friction at the bearing surface. Edge loading is observed on even well positioned and well functioning components and probably a part of normal hip function that cannot be avoided. New implant designs must therefore consider edge loading, particularly larger diameter devices where the lever arm available between the friction force and tapers or fixation interfaces is increased. Implant designs that reduce range of motion or move the edge of the bearing surface further into the acetabular cup must also be considered carefully, as these can increase the likelihood of edge loading. This is particularly relevant for current ceramic bearings where the edge of the bearing surface can be several millimetres below the cup face (Figure 2).

Risk mitigation forms the core of new device design, and standard testing protocols (ISO or ASTM standards) are widely used by manufacturers to mitigate identified risks. However, while there are extensive guidelines for femoral stem testing, there are a paucity of standard tests for acetabular cups, particularly related to edge loading and intraoperative assembly as described in this review. The responsibility is therefore on the individual manufacturers to mitigate these risks through their own test protocols. The risk management process is usually kept confidential and only reviewed externally by a notified body prior to awarding a CE mark. However, failings in this process have been demonstrated by recent high profile product recalls. Greater transparency of the identified risks and how they have been mitigated during product development could, through peer review, provide an additional mechanism to help prevent faulty designs reaching the market.

This article presents the clinical observations for CoC devices over the past 40 years and therefore provides the data necessary to challenge manufacturers as to how certain risks have been mitigated. The surgeon is ultimately responsible for the device implanted in the patient, and must be satisfied that clinical risks have been mitigated to his or her satisfaction, and not rely solely on the award of a CE mark or other regulatory approval.

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Study	Implant	No. hips	Mean f/u years (range)	Ceramic fracture (%)	Liner chipping on insertion (%)	Dislocation (%)	Aseptic loosening (%)	Squeaking (%)	Liner dissociation (%)	Survival %
Boehler 1994[43]	Alumina cup with pegs (Rosenthal Technik)	67	12	13.4	n/a	-	12.4 *	-	n/a	88% @ 11yrs
Boutin 1988[1]	Alumina cups with peg(s) (Ceraver)	>422	Max 15	-	n/a	-	0.5 - 26	-	n/a	-
	Cemented alumina cup (Ceraver)	560	5(1-9)	0.5	n/a	-	3.2	-	n/a	88% @ 7yrs
Fritsch 1996[44]	Alumina threaded cup (Autophor) and alumina cemented cup (Xenophor)	2832	11 (0-20)	0.2	n/a	-	Not clear	-	n/a	-
Garcia Cimbrelo 1996[11]	Alumina threaded cup (Autophor)	83	12 (5-16)	3.5	n/a	-	53% *	-	n/a	84% @ 16yrs
Griss 1981[2]	Cemented (Friedrichsfeld) and cementless (Lindenhof) alumina cup	130	3 (1-6)	6.9	n/a	0.8	2	-	n/a	-
Hamadouche 2002[45]	Cemented alumina cup (Ceraver-Osteal)	33	20 (19-21)	0	n/a	0	38.8	-	n/a	86% @ 20yrs
	Cementless alumina cup (Ceraver-Osteal)	85			2.3**		49.3			61% @ 20yrs
Mittlemeier 1992[3]	Alumina threaded cup (Autophor) and cemented alumina cup (Xenophor)	3079	Max 16	0.3	n/a	-	3-22	-	n/a	-
Nizard 1992[46]	Cemented alumina cup (Ceraver-Osteal)	187	Max 11	2.7	n/a	0.5	8	-	n/a	83% @ 10yrs
Riska 1993[47]	Cemented alumina cup (Ceraver Osteal)	143	7 (1-12)	0	n/a	-	9.8	-	n/a	-
	Ti threaded shell with alumina liner (Ceraver Osteal)	112	4 (1-7)				1.8			
Winter 1992[48]	Bulk alumina cup (Lindenhof)	100	12 (10-14)	8	n/a	-	14	-	n/a	-

Table 1: Ceramic hip replacements with series beginning between 1970 and 1979. All implants are cementless monoblock designs unless stated otherwise. *radiographically loose, ** peg fracture on insertion

Study	Implant	No. hips	Mean f/u years (range)	Ceramic fracture (%)	Liner chipping on insertion (%)	Dislocation (%)	Aseptic loosening (%)	Squeaking (%)	Liner dissociation (%)	Survival %
Hamadouche 1999[49]	Bulk alumina cup (Cerapress, Ceraver-Osteal)	62	6	1.6	n/a	-	4.8	-	n/a	91% @ 6yrs
Huo 1996[50]	Alumina threaded cup (Autophor)	93	9(5-11)	0	n/a	6.5	5.4	-	n/a	88% @ 10yrs
Jazrawi 1999[8]	Alumina threaded cup (Autophor)	60	13 (10-14)	0	n/a	-	33	-	n/a	-
Mahoney 1990[9]	Alumina threaded cup (Autophor)	42	4(2-6)	0	n/a	4.8	35.7 *	-	n/a	-
O'Leary 1988[6]	Alumina threaded cup (Autophor)	69	3(2-4)	0	n/a	-	27	-	n/a	64% @ 4yrs
Petsatodis 2010[4]	Alumina threaded cup (Autophor)	109	21	0	n/a	0	5.5	0	n/a	84% @ 21yrs
Yoon 2008[5]	Alumina threaded cup (Autophor)	84	17	0	n/a	-	15.5	-	n/a	81% @ 17yrs

Table 2: Ceramic hip replacements with series beginning between 1980 and 1989. All implants are cementless monoblock designs. *radiographically loose

Study	Implant	No. hips	Mean f/u years (range)	Ceramic fracture (%)	Liner chipping on insertion (%)	Dislocation (%)	Aseptic loosening (%)	Squeaking (%)	Liner dissociation (%)	K-M Survival
Baek 2008[51]	Porous Ti coated Ti shell with alumina liner (Aesculap Plasmacup)	100	7 (6-9)	0	0	0	0	1.0	-	96% @ 5yrs
Bizot 2001[52]	HA coated Ti shell with alumina liner	96	1(0-3)	0	-	-	1.0	-	-	-
Bizot 2004[53]	Ti mesh backed Ti shell with alumina liner (Ceraver Osteal)	71	8 (6-12)	1.4	-	2.8	1.4	-	-	94% @ 9yrs
Boyer 2010[54]	Ti mesh backed or HA coated Ti shell with alumina liner (Ceraver Osteal)	83	(7-15)	1.2	-	3.9	4.8	1.2	-	92% @ 10yrs
Garino 2000[23]	Porous coated Ti shell with alumina liner (Wright Medical Transcend)	333	2(2-3)	0	0.9	0.9	0	-	0.6	-
Iwakiri 2008[55]	Cemented UHMWPE encased ceramic liner (Kyocera ABS)	72	6(4-7)	4.2	-	1.4	-	-	1.4	91% @ 8yrs
Kim 2010[56]	Ti bead coated Ti shell with alumina liner (DePuy Duraloc)	102	11 (10-13)	0	-	2.0	0	1.0	-	99% @ 11yrs
Kress 2011[57]	HA coated Ti shell with alumina liner (Ceraver Osteal Cerafit)	75	11(10-11)	0	-	-	1.3	-	-	-
Lee 2010[58]	Porous Ti coated Ti shell with alumina liner (Aesculap Plasmacup)	100	>10	2.0	-	1.0	0	1.0	-	97% @ 10yrs
Mai 2010[59]	Porous or HA coated Ti shell with alumina liner (Stryker PSL/ABC) & HA coated shell with Ti encased alumina liner (Stryker Trident).	336	4(2-10)	-	-	0.3	-	10	-	-
Mesko 2010[60]	Porous or HA coated Ti shell with alumina liner (Stryker ABC)	325	8	0.3	2.6*	2.3	0.2	1.5	-	97% @ 10yrs
	HA coated Ti shell with Ti encased alumina liner (Stryker Trident)	605	5		0			0.7		
Murphy 2006[24]	Porous coated Ti shell with alumina liner (Wright Medical Transcend)	174	4(2-9)	0.6	-	0	0.6	-	0.6	96% @ 9yrs
Park 2006[61]	Porous coated Ti shell with UHMWPE encased alumina liner (Lima SPH Contact)	357	4(3-8)	1.7	-	0.3	-	-	-	-
Pattyn 2008[62]	Ti shell with alumina liner (Wright Medical)	190	3(3-6)	0	-	4	-	-	-	-
Sugano 2007[63]	Porous coated Ti shell with alumina liner (Anca Cermescoli)	180	6 (5-8)	0.6	-	3.9	0.6	-	-	-
Sexton 2011[64]	Alumina and alumina/zirconia composite liner (brand not provided)	2406	-	-	-	-	-	3.1	-	-
Vendittoli 2007[65]	Ti mesh backed Ti shell with alumina liner (Ceraver Osteal)	71	7(4-9)	0	0	0	0	-	0	-
Yoo 2005[66]	Porous Ti coated Ti shell with alumina liner (Aesculap Plasmacup)	93	6(5-7)	1.1	-	0	0	-	-	-

Table 3: Ceramic hip replacements with series beginning between 1990 and 1998. *reported by the same authors for the same patients in a previous article.

Study	Implant	No. hips	Mean f/u years (range)	Ceramic fracture (%)	Liner chipping on insertion (%)	Dislocation (%)	Aseptic loosening (%)	Squeaking (%)	Liner dissociation (%)	K-M Survival
Bascarevic 2010[67]	Ti fibre coated Ti shell with alumina insert (Zimmer Trilogy)	82	4 (3-5)	0	3.7	1.2	0	-	-	-
Chang 2009[68]	Porous coated Ti shell with alumina liner (Zimmer Trilogy, Plus EP-FIT, DePuy Duraloc)	42	5(3-8)	0	-	4.8	0	0	-	-
Chevillotte 2011[69]	HA coated Ti shell with alumina liner (Amplitude Horizon)	89	9 (6-10)	0	1.1	6.7	1.1	5.6	-	96% @ 9yrs
Cogan 2011[70]	HA coated Ti shell with alumina liner (Ceraver Osteal Cerafit)	265	(3-5)	-	-	-	-	2.6	-	-
Greene 2009[71]	Stryker devices	103	4(4-5)	0	-	0	0	4.9	-	-
Hamilton 2010[29]	Porous Ti coated Ti shell with alumina/zirconia composite liner and alumina/zirconia composite head (DePuy Pinnacle)	177	3(2-4)	1.1	1.1	2.8	1.1	0	-	98% @ 3yrs
Jarrett 2009[72]	HA coated Ti shell with Ti encased alumina liner (Stryker Trident)	131	2(1-3)	-	-	1.9	-	10.7	-	-
Keurentjes 2008[73]	Porous Ti coated Ti shell with alumina liner (Stryker Trident)	43	4(3-5)	0	-	-	-	21	0	-
Ki 2011[74]	Porous or HA coated Ti shell with alumina liner (Stryker Osteonics, Aesculap Bicontact, Howmedica ABG)	61	6(2-9)	1.6	-	-	-	22.9	-	-
Kim 2009[75]	Porus coated Ti shell with alumina liner (DePuy Duraloc)	100	6(5-7)	-	-	1.0	0	-	-	-
Koo 2008[76]	Porous Ti coated Ti shell with alumina liner (Aesculap Plasmacup)	359	4(3-5)	1.4	-	-	-	-	-	Not clear
Lombardi 2010[77]	Porous Ti coated Ti shell with UHMWPE encased alumina shell articulating with alumina/zirconia composite head (Biomet Exceed)	65	6 (2-9)	1.5	-	1.5	1.5	0	0	95 @ 6yrs
Millar 2010[78]	HA coated Ti shell with Ti encased alumina liner (Stryker Trident)	48	3(2-4)	0	-	2.1	0	0	-	-
Parvizi 2011[39]	HA coated Ti shell with Ti encased alumina liner (Stryker Trident)	1508	4(0-8)	0	-	-	-	7	-	-
	Porous Ti coated Ti shell with alumina liner (Biomet C2a, DePuy Duraloc, Stelkast Surpass)	248			-	-	-	0	-	-
Schroder 2010[79]	HA coated Ti shell with Ti encased alumina liner (Stryker Trident)	431	4(2-6)	0	-	1.1	0	1.9	-	-
Garcia Cimbrelo 2008[80]	HA coated Ti shell with alumina liner (Cerafit Triradius, Ceraver Osteal)	314	5 (3-8)	0.3	0	0.6	2.5*	0	0	97% @ 8yrs
Hasegawa 2006[81]	HA coated Ti shell with UHMWPE encased ceramic liner (Kyocera)	33	6(5-7)	6.1	-	9.1	6.1	-	3.0	83% @ 6yrs
Poggie 2007[82]	UHMWPE encased alumina liner compression moulded into porous tantalum shell (Implex Hedrocel)	315	Min 2	3.8	-	0.3	-	0.3	4.4	-
Swanson 2010[83]	Porous coated Ti shell with alumina liner (Encore Keramos, Wright Lineage, Plus Intraplant) with 12/14 taper stem	225	4(2-9)	-	-	-	-	3.6	-	-
	HA coated Ti shell with Ti encased alumina liner (Stryker Trident) with V40-taper neck stem	45		-	-	-	-	35.6	-	-

Table 4: Ceramic hip replacements with series beginning between 1999 and 2011. *radiographically loose

Study	Implant	Date implanted	Hips	Fixation	Follow-up	Aseptic loosening
Cotella 1990[84]	ICLH alumina on polyethylene (Morgan)	1980-4	6	Cemented. Four circular cement key holes at pole and circumferential cement grooves on wall.	7 yrs	17%
Furuya 1978[85]	Alumina on polyethylene	1971-4	13*	Circumferential grooves on wall, fixation method unspecified	3 yrs*	58%
Knahr 1981[86] and Salzer 1978[87]	Alumina on alumina (Rosenthal-Technik)	1974-80	19	Press-fit. Circumferential grooves on wall for initial stability and bone ingrowth.	2 yrs	68%
Wagner 1978[88]	Alumina on polyethylene (Rosenthal-Biokeramik)	1974-8	40	Cemented. Circular cement key holes at pole and on internal walls.	0.5-4yrs*	1.4%*

Table 5: Summary of the clinical literature on ceramic components in hip resurfacing. *Includes metal components that were part of the same study.

Property	Alumina 1970's	Alumina 1990's	Alumina/Zirconia composite 2000's
Bending strength (MPa)	400	580	1150
Fracture toughness (MPa $m^{1/2}$)	4	4.3	8.5
Vickers Harndess HV1 (GPa)	20	20	19
Average grain size (µm)	7.2	1.8	0.6
Young's Modulus (GPa)	380	380	350
Hot isostatic pressing	no	yes	yes
Proof testing	no	yes	Yes

Table 6: Summary of the mechanical properties of the material used in CoC bearings over the past 40 years.[16,89]

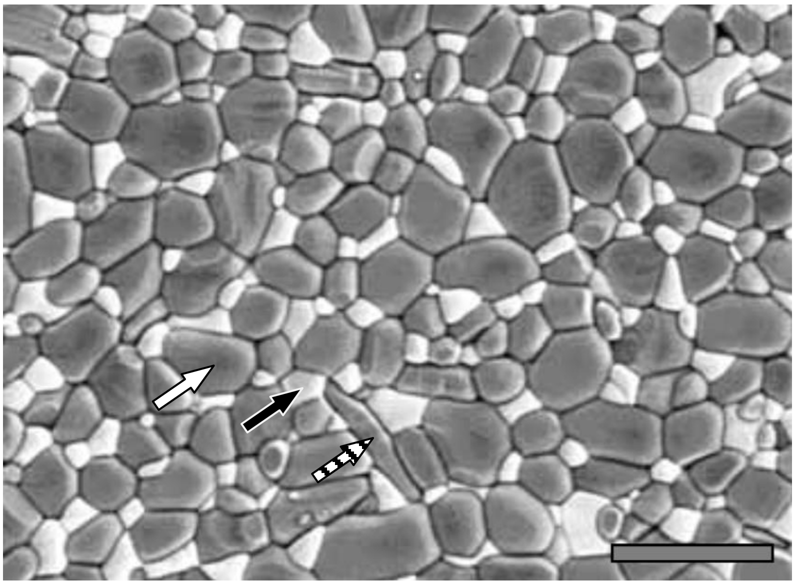

Figure 1: Micrograph of a contemporary ceramic material showing the alumina matrix (white arrow), zirconia crystals (black arrow) and strontium aluminate platelets (striped arrow). Scale bar represents 2 µm. (courtesy of Ceramtec AG, Plochingen, Germany)

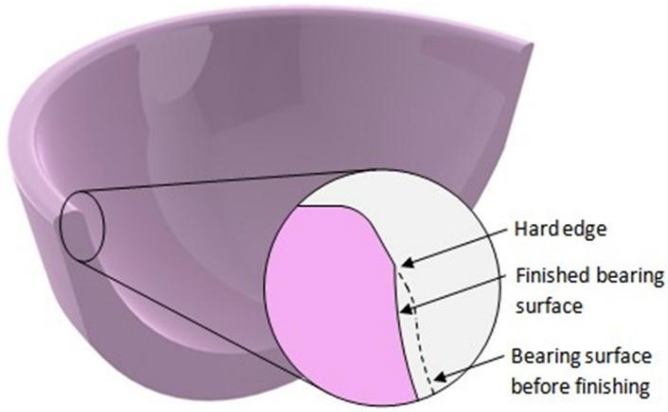

Figure 2: Cross-section image of a ceramic acetabular cup highlighting the sharp edge generated when the bearing surface is ground and polished. This can be seen a couple of millimetres in from the rim of the cup as a discontinuity in reflected light.

Figure 3: Photograph of the Stryker Trident acetabular cup

Figure 4: Photograph of the DeltaMotion acetabular cup

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