Chapter 11
Options for Primary Hip Arthroplasty

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Introduction

Total hip arthroplasty (THA) is widely regarded as one of the most successful procedures in orthopedic surgery. It significantly reduces pain, increases mobility, and restores function to patients who are otherwise incapacitated by degenerative joint disease. In addition, THA has a cost/utility ratio that rivals treatments for hypertension and coronary artery disease making it one of the most cost-effective medical interventions known [1, 2]. Despite this, an ever-increasing life expectancy and greater patient expectations for post-surgical activity have spurred advances in design and surgical technique which seek to increase the longevity of the prosthesis while minimizing morbidity. Such developments are crucial to reducing revision rates in THA patients who are younger and may require multiple revisions in their lifetime.

This chapter will seek to provide an introduction to the rationale behind the design of the acetabular and femoral stem components as well as the articulating surfaces. We will also examine three surgical approaches commonly used in THA implantation and discuss the advantages and hazards of each.

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Implant Design

In its simplest form, the design of hip arthroplasty consists of two components: one acetabular and one femoral separated by a bearing surface. Ideally, these components are rigidly attached to the surrounding bone while maintaining a nearly frictionless articulation between them. Thus, the design choices available to orthopedic surgeons relate to one of three elements: the acetabular component, the femoral component, or the bearing surface. The femoral component is perhaps the most complex and has a multitude of different options that relate to its shape, fixation method, and modularity. The main options for the design of the acetabular component relate to its fixation method. Lastly, the bearing surface options relate to the materials used in the articulating interface.

Bearing Surfaces

The bearing surface is the articulation between the femoral and acetabular components of the prosthesis. The ideal bearing surface materials are those that exhibit low friction, minimize wear, and have sufficient toughness to resist fracture. Additionally, any debris produced by the bearing surfaces should be biocompatible, i.e. not evoke an immune response. To date, no perfect bearing surface exists and arguments can be made for each in certain circumstances. In general, the materials used as bearing surfaces can be broken down into metals, ceramics, and plastics. These materials are coupled with either a similar (e.g., metal-on-metal) or different material (e.g., metal-on-plastic).

Perhaps the most difficult problem to solve has been the excessive wear of the articulating surfaces. This phenomenon is an obstacle both because it serves to disrupt the shape of the articulation surface and because it produces a significant amount of particles over time. This wear debris can in turn cause catastrophic implant failure or a localized resorptive response at the bone–implant interface that leads to implant loosening. The osteolytic response to wear debris remains the most frequent cause of failure and subsequent revision in total hip arthroplasty [3, 4]. In addition to the material used in the bearing surfaces, the femoral head size can have a significant effect on the wear generated. As such, a balance must be reached between the increased stability that larger heads provide with the increased wear particles that they produce. As patients are both living longer and the incidence of early arthritis is increasing, the need to improve upon the wear properties of bearing surface materials is self-evident.

Ultra-High Molecular Weight Polyethylene (UHMWPE) and Highly Cross-Linked Modifications

Although the original articulation surfaces in THR devices were metal-on-metal (MoM), the true success of hip arthroplasty began with the adoption of polyethylene as part of the bearing couple [5]. Early prostheses made by one of the

originators of THR, John Charnley, incorporated a plastic bearing on the acetabular component coupled with a metallic femoral head. Eventually, Charnley settled on high molecular weight polyethylene as the bearing of choice and thus created the first metal-on-polyethylene (MoP) device. Since then, polyethylene has been coupled with numerous metal alloys (e.g., stainless steel, cobalt-chromium, and titanium alloys) as well as ceramics (e.g., aluminum and zirconium oxides). Metal-on-polyethylene (MoP) couples remain the bearing of choice in the majority of total hip replacements today.

Despite this success, UHMWPE was shown early on to result in high amounts of wear in both laboratory and clinical studies. As mentioned earlier, the debris produced by this wear can result in implant failure through the gradual process of osteolysis and implant loosening. There have been numerous studies which have examined the wear of the Charnley hip prosthesis and although the literature differs in many details, it can be combined to form a cohesive picture of the wear process. In general, the wear rate of the polyethylene is highest at the beginning of the lifetime of the prosthesis and subsequently decreases to a relative steady state in the long term (16–18 months) [6]. The precise reason for this is unknown but it is hypothesized that creep, bedding-in, and decreased patient activity over time play a role. This means that examinations of wear rate immediately after implantation should be taken with caution as long-term studies of the same subjects will show a decrease in this property. Data gathered from hip simulators have shown that volumetric wear rates typically range from 23.2 mm^3/million cycles to 32.8 mm^3/million cycles [7]. When examining the clinical ramifications of this process, other studies have noted that wear rates of 38.8 mm^3/million cycles have resulted in a high risk for revision [8, 9]. In practice, this means that there is a relatively small difference between the typical wear rate seen in a successful versus unsuccessful arthroplasty. Thus, poor surgical technique, non-ideal implant placement, and increased load on the prosthesis due to a variety of patient factors can all significantly increase the risk of revision.

The size of the femoral head is also an important factor when discussing the wear of a bearing surface. Increasing femoral head size is an enticing design decision because it is one way of reducing hip prosthetic dislocation rates. Unfortunately, increasing the diameter of the femoral head also increases the sliding distance of the bearing and therefore increases the volumetric wear in all types of bearing surfaces [10]. A hip simulator study performed by Clarke et al. found a proportional increase in volumetric wear of approximately 7.8 % for every millimeter that the head diameter increased in MoP implants [7]. An in vivo radiographic study found that there was a 74 % increase in volumetric wear when 28 and 32 mm MoP bearings were compared [11]. A revision retrieval study of loose MoP acetabular components found an increase of 5.1 mm^3/year for each millimeter increase in the radius of the head [12]. Thus a tradeoff exists between the reduced dislocations that larger femoral heads provide and the increased bearing surface wear that they produce.

In an attempt to improve the properties of UHMWPE, namely the wear rate, technologies were developed to cross link adjacent polyethylene molecules. The product, Highly Cross-Linked Polyethylene (HXLP), was designed for its resistance to wear, reduction in wear particle volume, and subsequently its theoretically reduced rate of implant loosening. It is produced by manipulation of ultra-high

molecular weight polyethylene with either an electron beam or gamma irradiation. This treatment leads to the creation of free radicals along the backbone of the polyethylene molecule which combines to form a cross link between two separate molecules and results in the production of HXLP. This new material has been shown to reduce volumetric wear between 70 and 90 % in vitro [13, 14]. Muratoglu et al. used a hip simulator to show that, compared to UHMWPE, HXLP had similar mechanical properties with greatly improved wear resistance [15]. In fact, initial results obtained from hip simulators were very encouraging and showed no increase in the rate of wear with increasing head size even when using 46 mm heads. In vivo studies however have shown mixed results in the short term (3 year follow-up). In one study, linear wear was not found to increase with increasing head size, however other studies with a medium length follow-up showed an increased volumetric wear with increasing head sizes [16–18].

Simulators are often imperfect models for the conditions seen in patients. Thus, despite the lower volume of wear debris produced from HXLP when tested in a laboratory, it is plausible that the true amount of wear produced by a bearing surface could be greater than predicted. Conditions in the human body are known to be more damaging to arthroplasty components and therefore could result in significantly more wear. Relevant factors that increase debris generation include third-bodies, microseparation, edge loading, and damaged femoral heads. More complex in vitro studies have been carried out which have attempted to mimic these effects and compare the rate of wear between UHMWPE and HXLP. A study examining the performance of HXLP in contact with scratched surfaces noted a tenfold increased rate of wear in HXLP as compared with UHMWPE (i.e., 30× versus 3× that of a surface contacting an undamaged femoral component) [19]. Polymethyl methacrylate debris was found to increase wear 80-fold in HXLP as opposed to sixfold in UHMWPE when compared to an articulation that was not subjected to this type of third body [20]. Other studies have contradicted this data. McKellop et al. found that HXLP interfacing with roughened surfaces had better wear resistance than UHMWPE but the clinical relevance of the study was limited by the fact that the hip simulator used higher than physiological concentrations of protein in the lubricant [21]. Another property that is often lacking in hip simulators is microseparation due to joint laxity. This phenomenon is described as the separation of the bearing surfaces with concentric relocation during normal gait and is also associated with significant edge loading [22]. Interestingly, microseparation and edge-loading are not known to have a detrimental effect on hard-on-soft bearing surfaces such as MoP [23–25]. Conversely, metal-on-metal and ceramic-on-ceramic bearings are negatively affected by these phenomena and the ramifications of this effect will be discussed later.

Other concerns include the differences in the sizes of debris particles produced by wear in HXLP versus UHMWPE. This property is a known determinant of the ability of the material to produce an immune response and therefore cause implant loosening. It has been noted that particles less than 0.5 μm in diameter have the greatest effect on response by macrophages and the subsequent release of inflammatory cytokines [26–28]. Endo et al. showed that while non-cross-linked UHMWPE produces larger debris volumes, the particles produced by HXLP were

smaller and in a more biologically active (smaller) size range [29]. A mouse model with identically sized particles of HXLP and UHMWPE implanted under the periosteum of the calvaria showed a greater osteolytic response in the HXLP group (35 % versus 9 %). The overall message of this data suggests that although there is great potential in the decreased volume of wear produced by HXLP, the beneficial clinical outcome of decreased wear debris may be offset by the increased tendency of HXLP particles to induce an immune response that leads to osteolysis and eventual loosening.

Current clinical outcomes data has demonstrated the superiority of HXLP over UHMWPE but suffers from a lack of studies looking at outcomes beyond 10 years after implantation, primarily because HXLP was adopted relatively recently. Studies have shown a risk ratio of 0.4 for radiological evidence of osteolysis when comparing HXLP with UHMWPE [30]. A systematic review of studies that looked at greater than 5-year follow-up also supported this trend and has encouraged the continued use of the cross-linked polymer in bearing surface designs [31]. More recent randomized controlled trials have also been favorable towards HXLP. A 7-year, double blind, randomized controlled trial by Thomas et al. compared femoral head penetration between HXLP and UHMWPE acetabular liners. It was demonstrated that HXLP has a significantly lower steady state wear rate compared to UHMWPE, with a mean of 0.33 mm compared to 0.55 mm [32]. Shaun et al. reviewed 46 primary THAs that used first generation HXLP liners with a mean follow-up of 9 years. It was found that the linear penetration rate was 0.037 mm/year, demonstrating a 74 % reduction in total penetration when compared to conventional polyethylene [33].

Although the vast majority of polyethylene bearing surfaces articulate with metal femoral components, ceramic-on-polyethylene (CoP) bearing couples also exist and are an enticing option due to the lower surface roughness of ceramic femoral heads when compared with metallic alloys. Once again, initial simulator data showed a 20-fold reduction in the volumetric wear of ceramic versus metallic heads [34]. Subsequent in vitro studies using joint lubricant with a more physiologic composition were less favorable but still showed a 50 % reduction in polyethylene wear when using ceramic versus metallic heads [35]. Clinical studies found that CoP had a linear wear rate that was two to four times less than a MoP bearing [36, 37]. A study of 31 matched pairs of a CoP and MoP with a follow-up of 15–20 years found a 37 % decrease in the mean wear rate, but found no statistically significant difference in patient functional scores, radiographic evidence of osteolysis, or revision [37]. Ceramic-metal composites have also been developed to combine the surface hardness and scratch resistance of ceramics with the fracture resistance of metals (e.g., oxidized zirconium, OxZr). Surface hardness studies have found that OxZr heads have more than twice the hardness of CoCr heads while retaining the same wear effects on polyethylene as ceramic heads in vitro. In vivo, a randomized study comparing wear and migration of CoCr and OxZr heads articulating with HXLP found no difference after 2 years [38]. Early retrieval case reports in patients found reduced resistance to surface damage of the OxZr [39, 40]. Hip simulator studies using damaged OxZr heads retrieved from patients found a 50-fold increase in polyethylene wear as compared with pristine implants of the same material [41].

Metal on Metal (MoM) Articulations

In 1938 Philip Wiles used a MoM articulation in what is thought to be the first THA [42]. Later, in 1953, George McKee of Norwich, England, adopted the MoM articulation in combination with a modified stem originally used for hemiarthroplasty. Whilst this device showed good functional outcomes, its use was gradually phased out due to the success of the Charnley MoP arthroplasty, which demonstrated reduced short-term loosening rates. Recently there has been a resurgence in the use of MoM articulations due to the increased resistance to wear they offer over conventional MoP bearings [43]. Today, MoM bearings are used in traditional total hip arthroplasties as well as hip resurfacings.

One of the main factors responsible for the loosening of prostheses is wear debris. Volumetric wear is inversely proportional to the hardness of the softest surface of a given bearing couple [10]. In a MoP articulation, this is clearly the polyethylene and so that natural design progression is to replace this surface with another metal and thus form a MoM bearing. Indeed, MoM bearings have shown significant in vitro reductions in volumetric wear. Studies in simulators have shown wear rates between 0.2 and 2.5 mm^3/million cycles for MoM bearings as compared with 32.8 mm^3/million cycles and 9 mm^3/million cycles for similarly sized heads in couples incorporating HMWPE and HXLP bearings, respectively [7, 44, 45]. Despite the lower volumetric wear rate of MoM bearings compared to metal-on-polyethylene bearings, the size of MoM wear particles has been shown to be around 50 nm in size (cobalt chromium alloy particles) compared to 500 nm for polyethylene [46]. As a result of this the actual number of particles and the surface area of debris generated by MoM wear is greater and may raise concerns regarding a greater tissue response per unit volume (3, 4) [46, 47].

The actual effects of metallic particles surrounding tissues is an area of intense research. Lohann et al. have shown that phagocytosis of metal particles leads to a decrease in cell osteoblastic activity, which may contribute to the cellular events that lead to aseptic loosening of the implant [48]. Additionally, adverse local tissue responses (ALTR) that are distinct from those seen in patients with MoP prostheses have been observed in patients with MoM implants [49–52]. One subtype of these adverse responses is termed aseptic lymphocyte-dominated vasculitis-associated lesions (ALVAL). In contrast to the primarily macrophage and giant cell response seen in MoP implants, this lymphocyte-dominated reaction is much more severe and can result in not only implant loosening but severe soft tissue necrosis and pseudotumor formation [53]. Mahendra et al. describe a spectrum of necrotic and inflammatory changes in response to the deposition of cobalt-chrome (Co-Cr) wear particles in periprosthetic tissues [54]. It appears that the incidence of pseudotumor formation in patients with MoM resurfacings is somewhat based on patient demographics [55]. Glyn-Jones et al. examined a cohort of 1419 patients who received hip resurfacings and found that age at implantation and the sex of the patient significantly affected the need for revision due to pseudotumor formation. The overall revision rate for women was 3.8 % compared to 0.5 % for men. Younger patients required revision for pseudotumor formation more often: 6 % for individuals less than 40 years old and 1.4 %

for patients older than 40. In women under 40 years of age, 13.1 % required revision for pseudotumor formation at 6 years of follow-up [55].

Other studies have noted systemic distribution of metal ions. Urban et al. demonstrated that metal particle migration can lead to metal deposition in the liver, spleen, and para-aortic lymph nodes [56]. Increased chromosomal aberrations in peripheral blood lymphocytes have also been noted [57]. In an effort to monitor the reaction of individual patients to MoM implants, blood monitoring analyses have been developed. At baseline, patients receiving MoM arthroplasties show increases in metal ion concentrations. Daniel et al. showed that the levels of cobalt and chromium significantly increased at 1 year, followed by a decreasing trend until the 6th year [58]. A 30-year follow-up of patients with MoM or MoP by Dunstan et al. showed that while the levels of all metals used in the bearing surface remained elevated during the duration of implantation, Co levels in the blood increased by up to 50-fold in patients with loose MoM implants compared to the stable group. These results suggest a role for Co blood monitoring in patients with MoM implants as a means of screening for loose prostheses [59].

Despite this, some long-term follow-up studies have been favorable towards MoM systems. As expected, the rate of osteolysis in patients with these implants has been lower than those with MoP, on the order of 0–3 % at 10 years [60–64]. The Metasul metal-on-metal hip system (Zimmer, Warsaw, Indiana) was studied by Saito et al. and showed excellent long-term results. 90 patients were monitored with a mean follow-up of 12.3 years. The survival rate with an endpoint defined as revision surgery and radiologic loosening was 94.4 %. In this study no adverse reactions due to excess metal debris were observed [64]. Other studies have noted a 0–5 % rate of ALTR in this implant type [65]. One theory for this discrepancy is that correct implant positioning is paramount in achieving optimal implant survival and lower levels of wear debris in implants with MoM bearings. In vitro analysis of acetabular component orientations supports this hypothesis. Angadji et al. used a hip simulator to demonstrate increased wear rates and total wear volume with cup angle orientations of over 50° [66]. Campbell et al. found that misalignment of the acetabular component led to mechanical problems including increased edge loading and failure rates in constructs that utilized MoM articulations, specifically Birmingham hip resurfacings [67]. This edge loading leads to increased wear and as a result, particle release. Hart et al. demonstrated that cup inclination of over 50° in Birmingham hip resurfacings leads to an increased whole blood level of cobalt and chromium, further suggesting that metal levels can be minimized by correct alignment of the acetabular component [68].

Higher rates of soft tissue reactions have been described in constructs with large-diameter heads [69]. At the 2011 British Hip Society Annual Conference, large diameter MoM bearings were discussed and it was concluded that their use should be avoided. It also recommended that patients with MoM bearings be followed up for the life of the implant, especially in the first 5 years after implantation. Any patients with MoM bearing presenting with pain should be investigated appropriately with proper three-dimensional imaging in order to detect the presence of ALTRs and respond accordingly [70].

More research is needed to elucidate the conditions which induce ALTR to occur. It is possible that certain factors leading to ALTRs are not inherent to MoM designs and are thus under the control of the designer, manufacturer, or surgeon. For instance, while correct implant alignment and surgical technique are crucial for the longevity and function of all THAs, it appears that the tolerances allowable when using MoM are more stringent. Additionally, certain implant designs have been found to be more prone to failure as evidenced by the recall by DePuy of its ASR and ASR XL arthroplasty systems which will be discussed later in the chapter. Ultimately, MoM replacements are promising in their theoretical ability to decrease the rate of loosening and expand the indications of arthroplasty to younger patients. Unfortunately, these replacements are currently plagued by setbacks that are unique to the environment of two articulating metals.

Ceramic on Ceramic Articulations

Ceramic-on-ceramic (CoC) bearings represent another approach taken to avoid the frequency of debris-induced osteolysis in MoP arthroplasty. Since the 1970s, CoC bearings have been used due to their very hard nature, scratch resistance, and improved sliding properties. In fact, CoC bearings display even lower volumetric wear rates than MoM bearing couples. Some studies have observed wear rates as low as 0.004 mm^3/million cycles, a rate approximately 6000 times less than similarly sized MoP bearings [35]. Others have noted wear rates between 0.05 and 0.1 mm^3/million cycles which is still considerably less than both MoP and MoM bearing couples [71–73]. The frictional properties of ceramics also appear to be superior to the other bearing couples. Excess friction has been hypothesized to contribute to sudden loosening of acetabular components and thus should be minimized if at all possible [74]. In a hip simulator study using 25 % calf serum as lubricant, Brockett et al. found CoC to exhibit the lowest friction factor of any combination tested (e.g., CoM, CoP, MoP, and MoM) [75]. Additionally, in vitro studies have shown that ceramic wear debris is significantly less inflammatory than UHMWPE debris with a minimum volume of 100 μm^3 needed to induce the production of TNF-alpha, an inflammatory marker. Thus, given the biocompatibility and low rate of wear production, the volume needed to induce osteolysis is unlikely to ever be reached in vivo and is therefore not clinically relevant.

The analysis of actual clinical outcomes of CoC bearings presents a complex picture. Early CoC bearings utilized alumina as the bearing surface and showed decreased osteolysis, loosening and inflammation in comparison with polyethylene [76]. The main drawback of the early generation of alumina-on-alumina (AoA) bearings was high levels of ceramic fracture. Long-term survival of the early generation of AoA bearings was between 45 and 68.3 % at 18 years [77]. Recent advances in ceramic design and production have led to significant reductions in ceramic fracture rates. Hannouche et al. reported 13 fractures (8 in the femoral head component and 5 in the acetabular component), in a cohort of 5500 alumina components (3300 in AoA and 1200 in alumina-on-polyethylene) [78]. Long-term data has favored the use of CoC implants. For instance, a minimum 20-year follow-up for 85

hips with CoC bearings showed a 1.2 and 7.1 % incidence of radiolucencies measuring >2 cm around the femoral and acetabular components, respectively [79]. In this series, 6/85 (7.1 %) required revision, all due to aseptic loosening of the acetabular component. A different comparative study of CoC versus MoP bearings with a mean 8-year follow-up demonstrated osteolysis in 1.4 % of CoC implanted hips and 30.5 % of MoP implanted hips [80]. This combination of low fracture rate, reduced wear volume, with reduced rates of osteolysis has led to a resurgence in implants with a CoC articulation especially in young patients.

Despite these advantages, CoC bearings have the disadvantage of producing an audible noise, described as a "squeak," in select patients. While squeaking is a phenomenon that occurs in all hard-on-hard bearings, it appears to be self-limiting in MoM bearings and usually resolves within the first 6 months [81, 82]. Conversely, the squeaking of CoC bearings occurs later in the lifetime of the prosthesis and usually persists [83, 84]. Many potential causes of this squeaking have been proposed including implant design, patient factors, and implant malposition. More specifically, edge-loading, third bodies, and certain stem designs have been associated with an increased risk of squeaking [85–87]. Interestingly, despite being a common cause for revision, squeaking is not known to contribute to implant loosening or even osteolysis as evidenced by two studies with minimum follow-ups of 2.5 and 10 years, respectively [88, 89]. Reports on the incidence of squeaking vary greatly between studies. A prospective observational study of 1486 CoC THAs with a mean follow-up of 5.5 years found that 6 % of patients suffered from an audible squeak [89]. The majority of squeaks in these patients occurred during walking, climbing stairs, and bending forward. A prospective, randomized, multicenter study by Capello et al. of 475 CoC THAs found that only 0.8 % of patients noticed an audible squeak, with a mean follow-up of 5 years [80]. Contrary to this, a smaller study of 43 CoC THA by Keurentjes et al. reported audible squeaks in 20 % of patients [90]. A meta-analysis revealed a mean incidence of 2.4 % (0.7–20 %) for CoC bearings [85]. Options for patients who find the squeaking intolerable include an exchange of all components or simply the liner. Before surgery is recommended patients should be counseled that there is a chance that the squeaking may reduce over time.

It is clear that CoC THA is a viable option for patients, especially the young, who remain increasingly active. However the risk of squeaking, while small, has dramatically reduced enthusiasm for this bearing. Still, if the incidence of this phenomenon can be minimized, it is possible that ceramics could become the bearing of choice in THA.

The Acetabular Component

The acetabular component represents the proximal articulation surface of the total hip replacement. Its function is to replace the native acetabulum with a synthetic bearing that interfaces with the femoral component. The types of acetabular components can loosely be divided into two groups: monoblock and modular. As the name implies monoblock components typically consist of a single piece of either polyethylene or

metal that is machined in such a way that it serves as an interface with the surrounding bone on the convex surface while articulating with the femoral head on the concave surface. By contrast modular cups consist of two pieces: a shell and a liner. The metallic shell contains interfaces with the surrounding bone and contains a locking mechanism on the concave surface that is able to accept the liner. Similarly to the mechanisms of femoral component fixation, the acetabular components can either utilize cement or osseointegration to provide rigid and lasting attachment to the surrounding bone.

Cemented Versus Uncemented Acetabular Components

The initial design of acetabular components utilized cement as the means of fixation between a monoblock polyethylene cup and the underlying bone. Cementless modular acetabular components were introduced in the 1980s as a response to the idea that cement was the principal cause of loosening of the acetabular component. The term "cement disease" was used to describe the process of microscopic cement particles inducing osteolysis and resulting in eventual loosening. Since then, this concept has been challenged by studies that demonstrated that the major causes of osteolysis are reactions to polyethylene wear particles and hydrostatic fluid flow [91, 92]. Additionally, the increased degree of osteolysis seen in cementless cups as compared with cemented ones has caused surgeons to re-examine the decision to move away from cemented acetabular fixation [93, 94]. Still, few studies have compared the long-term results of cemented versus uncemented acetabular and the optimal fixation method has not yet been decided.

It is important to note that there is a huge variety of cemented and cementless acetabular components that all have multiple aspects of their designs that can be either beneficial or detrimental to implant survivorship. Therefore, overarching conclusions are sometimes difficult to draw due to the confounding effect created when comparing two systems that differ in more than just their fixation method. Regardless of these factors, a thorough review of the literature can describe trends that are useful when deciding whether to use cement or osseointegration as the fixation method in a given patient.

Multiple meta-analyses exist that compare cemented and uncemented fixation for acetabular components. Many of these suffer from heterogeneities in patient cohorts, bearing surfaces used, and other aforementioned confounders which serve to limit the generalizability of this data. Early systematic reviews pooled studies with short- and long-term outcomes which can skew data. One meta-analysis of 20 articles included studies with follow-up of as little as 1 year. Another study used similar follow-up criteria to examine cemented and uncemented acetabular components at short and intermediate follow-up. Both meta-analyses failed to show a better survival of cementless as compared with cemented [95, 96]. A meta-analysis of the literature published by Toossi et al. in 2013 examined survivorship or revision rate of primary total hip arthroplasty at a minimum of 10 years follow-up [97]. It analyzed 81 articles that examined the outcomes of cementless, cemented or both types of acetabular components and ultimately included 13,509 cemented components and 13,067 uncemented acetabular components in its analysis. Initially, the study did not reveal any

effect of the type of acetabular component fixation on either survivorship or revision rate, however a regression analysis showed that the estimated odds ratio for survivorship of a cemented acetabular component was 1.6 (95 % CI 1.32–2.4 $p=0.002$) when adjusted for age, sex, and mean duration of follow-up [97].

Osseointegration of Uncemented Acetabular Components

Modern uncemented acetabular components rely on a type of biologic fixation known as osseointegration for adherence to the surrounding bone. The mechanism of osseointegration seen in uncemented implants is classified as either ongrowth or ingrowth. Traditionally, ingrowth is bone deposition in the interstices of a porous surface and ongrowth is bone attachment to a flat implant surface. The difficulty with these definitions is that ultimately porosity is an arbitrary definition and so ingrowth may exist in situations where it is smaller than the resolution of current imaging. Furthermore, even in porous-coated implants, there are flat surfaces which facilitate ongrowth. Thus, both ongrowth and ingrowth can play a role in the fixation of an acetabular component. One criterion commonly used to distinguish these two processes describes ingrowth as growth into pores which are visible under light microscopy and ongrowth as a situation in which "no surface macroporosity at the level of a light microscope is visible and the bone appears to be directly attached to the implant material" [98].

Interestingly, early uncemented acetabular components relied on neither ingrowth nor ongrowth for fixation and were plagued by failures. These designs utilized the geometric shape of the implant, large pegs, or threaded rings for mechanical fixation. In a retrieval study by Bobyn et al. it was found that there was radiographic and histological evidence of fibrous tissue filling the threaded grooves with as little as 9 % of the threaded (fixating) component surface area in contact with the bone [99]. It was concluded that the lack of "micro-interlock" or osseointegration among other reasons was the cause of acetabular component migration and eventual failure. These failures led to the development of the second-generation threaded cups which added design features meant to support ingrowth through the use of porous coated metallic surfaces or ongrowth using the rough surfaces of grit-blasted metals. The superiority of porous over first generation threaded implants was shown in a matched-pair analysis of otherwise identical cup designs. Porous threaded implants performed significantly better at 2–4 years of follow-up with a 0 % loosening/revision rate noted in the porous group compared with a 29 % incidence of loosening and a 10.7 rate of revision noted in the first generation design [100]. Other studies with longer follow-up supported these findings [101, 102].

Bony ingrowth requires more than just a porous surface to occur successfully. The surface must also be consist of a biocompatible material, have optimal pore size, be in intimate contact with viable bone, and have adequate initial stability for osseointegration to occur successfully. In several studies, the ideal diameter of the pores was found to be between 150 and 450 μm [103, 104]. Studies in canines have demonstrated that less than 20 μm of micromotion allowed for optimal bone ingrowth into a porous titanium mesh while over 150 μm of micromotion resulted

in fibrous tissue at the bone–implant interface [105]. This fibrous tissue contributes to osteolysis, implant instability, and eventual failure [106, 107]. Although initial apposition of a porous implant with the surrounding bone surface is not absolutely necessary for successful osseointegration, the rate and degree of mineralization is enhanced when the initial post-surgical gap width is less than 0.5 mm [108, 109].

These findings along with clinical data have prompted manufacturers to favor press-fit (under-reamed) acetabular fixation as opposed to line-to-line designs [110]. Additional fixation strength could also be achieved with the addition of supplemental screw fixation. One cadaveric study comparing line-to-line with press-fit both with and without screw fixation found that press fit with screw fixation resulted in the greatest stability [111]. Other designs have utilized hydroxyapatite (HA) coated components as an osteoconductive material to improve bone ingrowth and ongrowth. Although successful on femoral components, the results of HA coated acetabular components have been mixed. Smooth HA coated designs have proven to have high revision rates and porous HA-coated implants have shown identical clinical results to non-HA coated porous designs except for a decrease in polar radiolucencies at 2 years [102, 112–114]. An 8-year randomized controlled trial found no superior survival or rate of revision of HA-coated porous acetabular cups over similar non-HA-coated implants [115].

Femoral Stem Design

The femoral component is designed to mimic function of the proximal portion of the femur. This component is the most diverse in terms of design options available to the surgeon owing to the complexity and nuances encountered in human femoral anatomy. This section will discuss the effects of the use of cement, different implant geometries, modularity, femoral head size, and bone preserving hip replacement techniques, i.e. resurfacing.

Femoral Head Size and Dislocation Rates

Native femoral head sizes are typically larger than those used in total hip arthroplasty. Although total hip arthroplasty has largely been successful even when utilizing smaller heads, problems with impingement free range of motion and dislocation have left room for improvement. In vitro data prompted researchers to experiment with larger femoral head diameters in order to increase stability after arthroplasty [116–118]. These designers sought to decrease the rate of dislocation by increasing the head-to-neck ratio, the jump distance prior to dislocation, and the tension on the surrounding soft tissues. Cadaveric studies showed that the range of motion increased significantly when larger heads were employed and in samples with larger head sizes the limitation in range of motion was due to bone impingement rather than component impingement (Table 11.1) [119].

Table 11.1 Average range of motion in cadavers with 32, 40, and 44 mm femoral heads [Reprinted from Cross MB, Nam D, Mayman DJ. Ideal femoral head size in total hip arthroplasty balances stability and volumetric wear. HSS J. 2012;8(3):270–4. With permission from Springer-Verlag]

Head Size	Limitations to range of motion	Flexion	Extension	Internal rotation	External rotation	Abduction	Adduction	Flexed internal rotation	Flexed external rotation
32 mm	Component impingement	119.0 (±4.0)	50.7 (±1.3)	128.9 (±3.0)	55.1 (±2.7)	56.3 (±2.7)	54.3 (±3.2)	32.3 (±0.8)	92.4 (±1.4)
40 mm	Component impingement	124.7 (±4.6)	59.1 (±1.8)	137.1 (±3.7)	63.7 (±3.2)	60.3 (±2.7)	58.4 (±3.4)	39.9 (±0.4)	100.4 (±1.1)
44 mm	Component impingement	127.5 (±4.9)	62.0 (±1.4)	139.5 (±4.9)	66.5 (±4.9)	61.5 (±3.5)	60.5 (±3.5)	42.0 (±0.0)	102.5 (±0.7)
	Bone impingement	124.9 (±8.9)	108.0 (±18.5)	135.6 (±6.5)	40.7 (±9.4)	76.0 (±6.4)	24.6 (±7.6)	44.9 (±5.3)	105.7 (±7.6)

As a result of this research, there has been a trend towards increasing femoral head sizes in total hip arthroplasty in an effort to decrease dislocation rates and thus improve the stability. This has largely been successful, especially for surgeons who perform total hip arthroplasty through a posterior approach which has been traditionally associated with dislocation. Population based registry studies undertaken in Finland and Sweden found that larger diameter femoral heads resulted in a decreased dislocation rate. The Finnish study examined patients with femoral heads that were 32, 36, and greater than 36 mm and compared them with 28 mm heads. The results showed a significantly decreased relative risk of dislocation of 0.4, 0.4, and 0.09, respectively [120]. A Swedish registry study found similar results, noting that the relative risk of revision of 28 mm heads compared with 22 mm heads was 0.5 [121] A recent systematic review which incorporated 24 randomized controlled trials found that larger femoral head size (36 mm vs. 28 mm) was associated with a decreased risk of implant dislocation [122].

Unfortunately, larger femoral heads come at the cost of increased wear properties. As mentioned in the earlier section, larger femoral head sizes result in a greater volumetric wear and therefore increased osteolysis, implant loosening, and eventual failure. A hip simulator study performed by Clarke et al. focusing on UHMWPE found a proportional increase in volumetric wear of approximately 7.8 % for every millimeter that the head diameter increased in MoP implants [7]. An in vivo radiographic study found that there was a 74 % increase in volumetric wear when 28 and 32 mm MoP bearings were compared [11]. A revision retrieval study of loose MoP acetabular components found an increase of 5.1 mm^3/year for each millimeter increase in the radius of the head [12]. Studies comparing different sized femoral heads articulating with HXLP found no difference in linear wear but significantly increased volumetric wear associated with increasing head sizes [16–18]. Similarly, MoM bearings have all shown increased wear associated with larger head diameters [123]. MoM articulations have the added risk of producing severe adverse soft tissue reaction and releasing serum ions as a result of wear [49–52]. Thus, the use of large diameter heads in the context of MoM bearings is currently not recommended [70].

When considering what size femoral head to use in a total joint arthroplasty ultimately the improved range of motion and dislocation characteristics of the larger heads must be weighed against the propensity of these components to produce more wear. Johnson et al. found that in order to perform activities of daily living, one must have hip flexion of 120°, hip abduction of 20°, and hip external rotation of 20° [124]. The greatest range of motion was seen in cadaveric samples with a femoral head size of 44 mm, however a 32 mm head appears to be sufficiently large to provide adequate range of motion for day-to-day functioning. Thus one study concluded that for the average patient, a 28 or 32 mm cobalt chrome on highly cross-linked polyethylene is a safe, durable, and effective bearing surface that balances the risk of increased osteolysis with a decreased propensity to produce significant wear [119].

Cemented Femoral Components

The total hip arthroplasties popularized by Sir John Charnley utilized components that were fixed to the surrounding bone using a self-curing acrylic bone cement. In these early trials, there was a wide variability in the success of cemented femoral components [125, 126]. This inconsistency can at least partially be explained by the evolution of cementing techniques over the past half century. Originally the cement distribution technique involved finger packing the bone cement into an unplugged femoral canal. Modern cementing techniques involve cleansing the canal with pulsatile lavage, inserting cement in a retrograde fashion, porosity reduction via vacuum mixing, and cement pressurization within the canal. Additionally, the stem is centralized proximally and distally in order to ensure an adequate and symmetric stem mantle. Thus, when critically examining clinical data, it is important to determine which cement technique was used when placing the component.

The mechanisms of failure of cemented femoral components are typically the result of mechanical factors initiating femoral loosening. Debonding, or separation of the cement from the stem, occurs followed by high stresses produced in the cement mantle proximally and at the distal tip of the implant [127–130]. These stresses then initiate crack formation which further destabilizes the implant and produces debris which results in an inflammatory reaction, bone resorption, and a soft tissue membrane which is commonly encountered in aseptic loosening. Interestingly, unlike uncemented components, this fibrous membrane forms late in the loosening process of cemented components and is not thought to play a significant role in the initial loosening process [107, 131]. The factors that led to the initial debonding were examined in numerous studies. Radiolucencies in the cement mantle signifying poor distribution of cement were found to predict later failure [132]. Varus implant position was also found to be associated with a higher risk of aseptic loosening and was thought to result in adverse outcomes due to its propensity to create a poor cement mantle [133, 134]. Thus in an effort to evaluate cement mantles, a grading system was created by Barrack et al. [135] which distinguished complete filling of the proximal diaphysis (A), near complete filling (B), incomplete filling with either greater than 50 % demonstrating radiolucencies (C1) or less than 1 mm of mantle present (C2) and gross deficiencies in the mantle with no cement distal to the tip or multiple large voids (D). Aseptic loosening was associated with C and D mantles with the latter having the greatest amount of implant failures. Other variables that contributed to failure included increased weight of patient, younger age of patient, male sex, and patients with post-traumatic osteoarthritis. Interestingly, the type of implant and surgeon did not correlate with a change in incidence of revision in this study [136].

Clinical outcomes of cemented femoral components have improved with advances in cement techniques. Distal plugging of femoral canal, extensive lavage, and retrograde injection of bone cement resulted in reported mechanical failure rate of 1–5 % in studies performed by Ranawat et al., Madey et al., and Smith et al. at 15 years

follow-up [137–139]. Contradictory results were reported by Sanchez-Sotelo et al. who found that the mechanical failure rate was 10 %. Stratification revealed a 4.3 % failure rate in patients older than 50 years and a 27.7 % failure rate in patients younger than 50 years [140]. Further advances in cementing technique including pressurization and vacuum preparation of the cement have resulted in further decreases in failures. A 13.5-year follow-up of 204 THAs placed with this cementing technique showed only 4 revisions: one for osteolysis, two for recurrent dislocations, and one delayed infection [141].

Uncemented Femoral Components

Uncemented implants were originally created to deal with the issue of "cement disease" or the lysis of periprosthetic bone. Cementless femoral components rely on osseointegration, the structural and functional connection between bone and implant, without intervening soft tissue, for their fixation strength. The mechanism of osseointegration seen in uncemented implants is classified as either ongrowth or ingrowth. Traditionally, ingrowth is bone deposition in the interstices of a porous surface and ongrowth is bone attachment to a flat implant surface. As noted earlier, bony ingrowth requires more than just a porous surface to occur successfully. The surface must also be consist of a biocompatible material, have optimal pore size, be in intimate contact with viable bone, and have adequate initial stability for bony ingrowth to occur successfully [103–105, 108, 109]. The ideal values for these parameters are discussed earlier in the chapter when describing the osseointegration of uncemented acetabular components.

While uncemented components have enjoyed great success, it is currently recognized that "cement disease" is a misnomer and is instead referred to as osteolysis. This process is known to cause loosening and eventual failure in both cemented and uncemented implants. Osseointegration of cementless porous coated femoral stems has proven to be a reliable and successful form of fixation but some features of cementless stems such as stem geometry, surface properties of the porous surface, and the extent to which the porous surface is applied to the stem continue to be a source of debate. One way of categorizing femoral components is by stem geometry. The two most common types of geometries are anatomical and tapered. The differences between these include their shape, metallurgy, head-neck design, and ongrowth–ingrowth surface.

Anatomic and Tapered Designs Early hip arthroplasty designs featured anatomic stems. These designs were constructed from cobalt-chrome alloys and achieved fixation by osseointegration of the proximal femoral component. Unfortunately, patients receiving these replacements frequently complained of thigh pain due to some combination of modular mismatch, endosteal irritation, and lack of ingrowth. Efforts to improve this design have led to the development of tapered femoral components.

The design rationale behind tapered components is geared towards promoting long-lasting osseointegration between the component and the surrounding bone. This requires rigid initial fixation until full osseointegration can be achieved. This initial stability is provided by the taper when it is forced into the medullary canal thereby producing circumferential hoop stresses that do not allow an axially loaded tapered stem to advance any further [142]. Rotational stability is provided by a rectangular cross section of the tapered stem although circular cross sections also exist and have shown highly satisfactory results as well [143]. An advantage of the circular cross section over the rectangular one is the ability to correct femoral anteversion if necessary. The tapered design has been shown to reduce stress shielding and lead to a decreased prevalence of thigh pain [144–147]. Femoral fixation requires osseointegration of the femoral component, typically in the proximal stem because of the maximal contact between bone and implant afforded by this area. Once osseointegrated, the stem is rigidly and lastingly fixed to the surrounding bone.

Clinical studies have favored the use of tapered designs over anatomic ones. In one study of 311 Porous Coated Anatomic (PCA) stems the overall survival rate of the femoral component at 14 years was 95 %. Unfortunately this design also showed a 36 % prevalence of thigh pain and 42 % of individuals had significant amount of femoral osteolysis as evidenced by radiography [148]. One study with 10- to 13-year follow-up of tapered total hip replacements in 283 patients found a 99 % survival rate of the femoral component with osteolysis seen in only 6.2 % of cases [149]. Activity related thigh pain was also low, noted in only 3 % of patients [149]. Very long-term studies of greater than 20 years have also shown positive results for tapered designs. The 20- to 25-year survival rate for tapered total hip replacements with follow-up ranging from 20 to 25 years was found to be between 86 and 95 % [150–152]. A long-term study of 47 obese patients with 18- to 27-year follow-up found a 94 % survival [153]. Risks for failure in a study of 326 patients receiving tapered THA with a follow-up of 22 years were found to be undersized stems, and hips where the cup was already revised. This study had a high (38 %) rate of cup revision due to the use of smooth-threaded cementless sockets [151].

Proximal and Extensive Porosity Another point of contention among designers of femoral components is the ideal extent of porous coating that should be applied to a femoral implant. In general, uncemented press-fit designs fall into one of two categories: fully porous-coated cylindrical stems that achieve distal fixation and proximally porous-coated tapered stems that achieve proximal fixation.

Extensively porous coated prostheses are defined as those with porous coating of more than 80 % of the surface area of the stem. The most common of these is the Anatomic Medullary Locking (AML) stem produced by DePuy. A critical design feature of the AML is its straight, cylindrical, non-tapered distal stem geometry. Thus, the stem does not wedge in place and fixation depends on a "scratch fit" between the rough external surface of the implant and a similar shaped bone canal. The theoretical advantage of the extensive coating is that it allows for osseointegration over the entire length of the stem. In this design, distal porosity is particularly important because the distal part of the stem most consistently contacts with the

cortical bone. Interestingly, even in cases where successful osseointegration does not occur and a fibrous tissue membrane forms between the bone and stem, adequate fixation is still achieved and there is sufficient radiographic stability as well as patient satisfaction owing to the extensively coated surface [154]. Conversely, proximally porous coated femoral implants can be either tapered or cylindrical and were intended to achieve biologic fixation solely in the femoral metaphysis. This property was intended to reduce proximal stress shielding and preserve bone stock.

Both designs have proponents that cite multiple publications with reproducible long-term follow-up. The reported incidence of thigh pain in patients receiving THA with a fully coated femoral component has varied between 3 and 20 % [155–157]. Today, most surgeons presume that proximally coated stems and cemented stems are less predisposed to causing thigh pain than fully coated stems although this idea remains controversial [158–161]. Similarly, femoral bone loss as a result of stress shielding around well-fixed femoral components is thought to occur more frequently in extensively porous coated stems. In theory, bone loss in the femur could lead to greater and lesser trochanteric avulsion fractures and make future revisions more difficult. However, currently, it is important to note that stress shielding is more of a radiographic finding than a diagnosis and the clinical ramifications of this process have not yet been demonstrated. A prospective randomized blinded clinical trial followed 388 patients receiving either proximally porous coated and fully porous coated femoral components (Table 11.2). A minimum follow-up of 2 years with a mean of 6 years was used and found that post-operative clinical outcome scores were similar at all follow-up intervals. There were no differences in incidence of thigh pain at any time although bone density reduction was greater in the fully coated stem as compared with the proximally coated [162].

Modular Femoral Components

Despite the success of monoblock implants, these designs were limited by their inability to fine-tune two properties which can differ greatly between patients: offset and leg length. Since dislocations remain one of the most frequent post-operative complications after THA, designers sought to reduce the dislocation rate by allowing surgeons to make adjustments in the offset and neck length of implants. Thus modular femoral stems were intended to allow a more accurate reproduction of patient anatomy. The use of a modular head-neck junction allowed the surgeon greater freedom in adjusting for leg length discrepancies as well as optimizing the function of the abductors [163]. Modular head-neck components also allowed for easier revision of femoral components when the femoral stem is clearly fixed and requires no further fixation. Unfortunately, increased customizability came at a cost and these implants suffer from an increased risk of fatigue failure, fretting, and crevice corrosion.

There is a vast array of modular femoral stems that allow for retention and sacrifice of various components of hip anatomy. While most modular femoral components are used in revision total hip arthroplasty, some complex primary total hip

Table 11.2 Incidence and severity of thigh pain in Synergy™ (proximally coated) and Prodigy™ (fully porous coated) femoral stem groups [Reprinted from MacDonald SJ, Rosenzweig S, Guerin JS, McCalden RW, Bohm ER, Bourne RB et al. Proximally versus fully porous-coated femoral stems: a multicenter randomized trial. Clin Orthop Relat Res. 2010;468(2):424–32 with permission from Springer Verlag]

	Incidence			Severity*		
Time	Synergy™	Prodigy™	*p* Value	Synergy™	Prodigy™	*p* Value
Preoperative	71 %	69 %	0.75	80.74 (±20.24)	80.96 (±20.70)	0.84
6 months	11 %	15 %	0.254	38.67 (±31.96)	37.67 (±23.01)	0.94
1 year	11 %	14 %	0.512	42.82 (±25.06)	47.42 (±27.99)	0.683
2 years	9 %	6 %	0.527	42.25 (±34.76)	33 (±20.05)	0.661

[a]Values are expressed as mean ± SD on a 100-mm visual analog scale

arthroplasties can benefit from the use of these devices as well. Modularity can occur at various points throughout these femoral stems and great emphasis has been placed on the association between modularity positioning and implant failure. Systems such as the Zimmer ZMR utilize a mid stem modularity allowing independent selection of the sizing and positioning of the proximal and distal components. The DePuy S-ROM is a proximal modular femoral implant that utilizes a titanium stem with distal spines to achieve initial fit and rotational stability. Standard and calcar replacement options are available with variable offset options. A separate sleeve that incorporates a "step-like" geometry to convert shear to compressive forces is available in both porous and hydroxyapatite coated designs. Between the proximal and distal stem segments is a tapered section that engages this enveloping sleeve. The sleeve is positioned first and the stem is placed through the sleeve and engaged. This allows the S-ROM stem to separate hip biomechanics and component fixation. The sleeve achieves fixation whilst the stem allows for adjustments in length and offset. In all, the S-ROM design allows for 10,398 different reconstructive possibilities [164]. Unfortunately, stems with distal modularity have underperformed other systems and most have been withdrawn from the market.

Modular stems have produced good clinical outcomes in certain situations. Restrepo et al. examined initial distal fixation, femoral offset restoration, leg length equalization, and hip stability in 118 patients who underwent revision with Stryker Restoration Stem [165]. This system consists of a fluted, titanium conical distal stem, which attaches to a proximal body. Adequate bone ingrowth and fixation was obtained in 100 % of patients and the offset was corrected in 66 % while leg length discrepancy was corrected in 78 %. Ultimately, stability was achieved in 97 % of patients who received this implant. Initial concerns were raised regarding failure of the modular junction but in this study with a 4–7 years follow-up, no failure/fracture was observed in a total of 118 patients [165].

When two components lock into each other, wear and corrosion are an inevitable consequence. The production of wear particles can lead to osteolysis and corrosion can lead to implant failure. More alarmingly, metallosis and adverse local tissue responses (ALTR) similar to that seen in metal-on-metal bearing couples are being

reported in patients with modular hip arthroplasties [166]. One subtype of these adverse responses is termed aseptic lymphocyte-dominated vasculitis-associated lesions (ALVAL). This lymphocyte-dominated reaction is very severe and can result in not only implant loosening but also severe soft tissue necrosis and pseudotumor formation [53]. Modular stems that incorporate a double taper (head-neck and neck-stem) add additional sites for failure and corrosion, with the added risk of increased wear particles.

Kop et al. examined 57 retrieved modular stems of 7 different designs. Of these, three were cobalt-chromium-molybdenum based and four titanium based [167]. The aim of this retrieval study was to assess whether the same degradation mechanism was present at the head-neck and neck-stem junction, whether or not the additional junction contributed to the revision, if the implant alloy affected the extent of degradation, and if the trunion machine finish affected the degradation mechanisms. Corrosion and fretting were both lowest in the titanium components with 62 % of the Co-Cr-Mo components having corrosion of the trunion, and 90 % fretting. In contrast, 30 % of Ti-based components showed corrosion, with 50 % exhibiting fretting. However cold welding of the titanium components did occur. It was concluded that titanium modular components may reduce the amount of degradation, but at the expense of an increased risk of cold welding.

Kop et al. also used a retrieval study of 16 modular components to examine the relationship between corrosion, material, and implant time. Of the retrieved implants 6 tapers showed fretting corrosion, with the average implant time being 39 months. No corrosion was shown in the remaining 10 tapers, which had an average implantation time of 2.7 months. It was concluded that even with modern materials and taper designs corrosion is still a concern with added modularity [168].

Gilbert et al. examined 148 retrieved modular hip prostheses. Significant corrosion was noted on 16 % of necks and 35 % of heads [169]. Concerns have also been raised regarding the elevated serum metal ions levels produced due to corrosion of modular stems. Once implanted, a protective surface oxide layer forms on the implant (modular implants are typically composed of titanium or Cobalt-chrome). While this film provides added corrosion resistance to the implant, it is subjected to repeated disruption as a result of stresses applied to the prosthesis. As the film reforms it reduces oxygen in the surrounding soft tissue. This process repeats itself and results in a reduced ability of the film to protect the implant [29].

Jacobs et al. describe corrosion is also affecting the structural integrity of the implant. This has been implicated in isolated incidents of fracture. Wright et al. report a case of fracture of the modular neck in a 49-year-old man. The fracture occurred while bending forward to tie his shoes, in the context of a fall onto his hip 2 months previously. On examination with light microscopy marked fretting and corrosion were noted along with debris [170].

Despite the intra-operative advantages offered by modular stems, the added junctions with the implant can lead to increased corrosion and wear particle production. While implants such as the S-ROM and modular taper stems have revolutionized revision surgery the use of modularity in primary THA should be avoided due to the risk of fatigue fracture and corrosion.

Hip Resurfacing

While THA is the treatment of choice for osteoarthritis of the hip there is an increasing cohort of patients who are requiring replacements earlier in life in order to remain physically active. While THA offers good symptomatic relief for these patients, they are likely to require revision in the future as a result of the increased physical demands placed on the implants, their increased life expectancy, and the insufficient longevity of traditional total hip arthroplasty. Hip resurfacing was developed as one solution to this problem and involves the preservation of the femoral neck through the use of a cap over the femoral head. This approach conserves femoral bone and theoretically allows for easier revision in the future, should it be needed.

The results of early resurfacing procedures in the 1970s and 1980s were poor. Accelerated wear and large volumes of biologically active wear debris resulted from the combination of a large articulating surface with a thin polyethylene liner and led to high rates of implant loosening and bone loss. With the development of new, more wear resistant materials, resurfacing has been reintroduced, this time incorporating a metal-on-metal (MoM) articulating surface [171]. As such, the downsides of MoM bearing couples which have been discussed earlier in this chapter must be weighed when deciding whether to pursue this reconstruction option.

As the stability of the femoral head is a key component in the effectiveness of resurfacing, severe bone loss, cysts or osteonecrosis of the femoral head or neck are contraindications for resurfacing. Retention of the femoral neck also means that unlike THA, femoral neck fractures may occur. Marker et al. used a prospective cohort study to identify the incidence of femoral neck fracture. 550 resurfacings performed by a single surgeon were studied. It was shown that 14 (2.5 %) had resulted in fracture of the femoral neck. Of these, 12 had occurred in the first 69 resurfacings performed, with women and obese patients shown to have a higher cumulative incidence of fracture. It was concluded that the risk of femoral neck fracture is multifactorial, associated with both the surgical learning curve and patient selection [172].

In vitro studies have suggested more limitations to hip resurfacing. Bengs et al. and Kluess et al. have both demonstrated reduced range of motion (ROM) in resurfacing compared to THA. Kluess examined the ROM of eight resurfacing prosthetics using 3D CAD models. The ROM of the resurfacing systems was found to be substantially less than that of total hip prosthetics with the large diameter of the femoral neck leading to impingements in all maneuvers analyzed [173]. Bengs examined the ROM of eight different hip replacement designs implanted into composite femurs and pelvises. It was found that compared to the THA prosthetics, resurfacings showed reduced ranges of motion, with, once again, early impingement of the femoral neck [174]. This is important since hip resurfacing is largely being indicated for younger and more active patients who require an increased ROM for various activities. Fortunately, despite these studies demonstrating reduced ROM in resurfacing systems, these reductions have not been seen when applied to the clinical setting. Clinical studies have shown similar ROM for both resurfacing and THA systems. Le Duff et al. examined 35 patients who had undergone bilateral

surgery receiving THA on one side and resurfacing on the other, with a mean follow-up of 88 months. They found no difference in ROM between the two systems [175]. Shimmin et al. suggested that while THA shows greater ROM in laboratory studies, this cannot be recreated in patients with normal flexibility leading to similar ROM in both THA and resurfacing [176]. Shimmin also reports that of the nine papers comparing functional outcomes between THA and resurfacing, eight showed consistently similar outcomes [176].

Springer et al. used a large meta-analysis to compare the results of 3269 resurfacings with 6408 cementless THAs. Femoral revision for mechanical failure was used as an endpoint and was found to be 1.3 % in the THA group with a mean follow-up of 8.4 years compared to 2.6 % in the resurfacing group with a mean follow-up of 3.9 years [177]. Johanson et al. used the Nordic Arthroplasty Register Association database to examine the non-septic 2-year revision risk of 1638 resurfacings and compared to 172,554 THAs. By 2 years the revision rate for resurfacings was 2.4 % compared to 1.1 % for THA [178].

It is clear that not all resurfacing systems yield similar results. Seppanen et al. examined the Nordic Joint registry between 2001 and 2009. During this time 4401 hip resurfacings were performed. When comparing the Articulating Surface Replacement (ASR) Hip resurfacing system (DePuy Orthopedics, Warsaw, Indiana), to the Birmingham resurfacing system (BHR) it was found that the ASR had inferior outcomes to the BHR with a relative revision risk of 1.8 (CI: 1.2–2.7) [179]. The ASR Hip resurfacing system was recalled by DePuy voluntarily in August 2010, based on unpublished data from the National Joint Registry of England and Wales that showed a 12 % revision rate at 5 years [180]. In 2007, the Australian Orthopedic Association National Joint Replacement Registry (AOANJRR) reported that the ASR revision system had higher than expected revision rates (3.0 revisions per 100 observed component years), that were twice that of other resurfacing systems with a cumulative percent revision at 2 years of 5.16 % [181]. De Steiger et al. reviewed the AOANJRR between 2003 and December 2009 identifying 1167 ASR resurfacing procedures. It was found that the cumulative revision rate at 5 years for the ASR resurfacing system was 10.9 % compared to 4.0 % with all other hip resurfacing prosthesis [180].

While the literature shows that resurfacing is a viable alternative to THA, especially in younger patients, it is clear that adequate surgical experience must be combined with careful selection of the patient and prosthesis in order to achieve optimal results. Additionally, it is important to note that hip resurfacings utilize a MoM bearing couple which carries inherent issues as discussed earlier in the chapter.

Surgical Approaches in Hip Arthroplasty

Numerous surgical approaches to the hip exist for use in hip arthroplasty. Although evidence in the literature can be found for the use of certain approaches over others for given indications, realistically most surgeons will use the approach that they are

most comfortable with for the vast majority of their cases. In certain situations, however, there are definite advantages to either modifying the surgeon-preferred approach or using a different one altogether. The need to access a particular anatomic region either because of bony deficiency or in order to remove hardware from a previous surgery are two such examples. Additionally, modifications of existing approaches must be utilized in certain populations. Patients with difficult anatomy, such as obese or muscular patients, require larger incisions, longer retractors, and more surgical assistants in order to have a satisfactory clinical outcome.

Recently there has been a surge in interest within the orthopedic community to perform THAs using minimally invasive surgery (MIS). Mini-incisions are either smaller versions of conventional approaches or novel incisions which are used to gain access to the acetabulum and femur. *Ultimately, the invasiveness of a procedure is more dependent on the amount of soft tissue damage that it causes rather than the size of the incision*. MIS appears to be less disruptive to soft tissues and advocates point to the potential for reduced intra-operative blood loss, decreased muscle damage, shorter length of hospitalization, reduced post-operative pain, and improved cosmesis that these methods offer. Conversely, some believe that the current practices in THA already produce excellent results with low complication rate, significant improvement in patient function, and excellent long-term prognosis. They argue that smaller incisions impair intra-operative visualization and lead to implant malposition, increased risk of intra-operative fractures and the potential for increased muscle damage. Concerns have also been expressed over the risk of neurovascular injury and poor implant fixation [182].

Posterolateral

The posterior or Moore Southern approach is currently the most popular technique for total hip arthroplasty. Many surgeons favor the posterolateral approach as it is less technically demanding than other methods and results in limited muscle damage while allowing for simple extension of the incision if needed. This method does not violate the abductor mechanism and is therefore thought to result in a lower incidence of post-operative Trendelenberg gait [183–186]. The major disadvantage of this approach is the risk of posterior dislocation due to the need for the release of the external rotators. This risk can be minimized by utilizing a careful repair of the posterior soft tissue structures [187–191] or through the utilization of larger diameter femoral heads. Despite these modifications, dislocation remains the main post-operative concern of the posterolateral approach [192–194].

The approach is performed with a 10–15 cm curved incision centered on the posterior aspect of the greater trochanter. The fascia lata is split in line with the incision and the fibers of the gluteus maximus dissected bluntly to reveal the short external rotators. These are then detached close to the femoral insertion and reflected thus exposing the posterior aspect of the hip joint and capsule. An incision is then made in the joint capsule and internal rotation of the thigh is used to dislocate the

femoral head and thus expose the joint. Proximal and distal extensions are possible and can be used to visualize the ilium or middle-distal femur, respectively. In obese or muscular patients, this approach can be modified in order to obtain adequate exposure. Often, the release of either the quadratus femoris, gluteal sling, reflected head of the rectus femoris, or the anterior capsule can be used to mobilize the femur and provide adequate exposure to ream the acetabulum. Failure to do this in obese or muscular patients can result in excess retroversion and likely contributes to the increased incidence of dislocation that is associated with this approach [195]. The MIS adaptation of the posterolateral approach utilizes a smaller incision (8–10 cm) along with minimal quadratus femoris release and a less invasive dissection.

Studies on the clinical outcomes of the posterolateral approach, whether traditional or MIS as compared with other approaches have been difficult to interpret due to confounding variables and conflicting results. A Cochrane review by Jolles and Bogoch examining the merits of a posterior versus lateral surgical approach for total hip arthroplasty in adults with osteoarthritis found the quantity and quality of trials to be insufficient to make a recommendation [196]. One study of 1793 primary THRs performed by either the posterolateral or direct anterior approach found that the former resulted in a 1.2 % decrease in wound infection compared with the latter (7/505 vs. 3/1288) [197]. Studies have found a longer duration of rehabilitation, greater blood loss, increased use of transfusion, greater narcotic usage, and a longer hospital discharge associated with the posterolateral approach [198–201]. Other studies conducted to compare parameters such as gait have found similar results regardless of approach [202]. A prospective nonrandomized multicenter study of 1089 THAs found no difference in Oxford hip scores, dislocation rates, or revision rates between anterolateral versus posterior hip replacements at 5 years follow-up [203]. Unfortunately most of these studies were nonrandomized and so suffer from selection bias which limits the generalizability of the results.

Others have compared the traditional posterolateral approach to the MIS adaptation and have favored the adoption of the minimally invasive option. Sculco et al. reported on 1500 procedures that utilized the MIS adaptation and found that the complication rate for dislocation was 1.2 %, with femoral fracture and sciatic neuropraxia rates both at 0.3 % [204]. A randomized controlled trial found that patients who underwent a minimally invasive total hip arthroplasty demonstrated decreased blood loss and limped less at 6-week follow-up [205]. A systemic review by Cheng et al. compared the operative outcomes between standard and MIS in THA. It was found that operative time and blood loss were significantly reduced in the MIS group for patients with a posterolateral incision. There were no statistically significant differences reported in post-operative outcomes between the standard and MIS groups [206]. Berstock et al. conducted a systematic review and meta-analysis of the standard versus mini-incision posterior approach to total hip arthroplasty and found that the mini-incision posterior approach was associated with an early improvement in the Harris hip score, reduced operating time (by 5 min), reduced hospital stay (by 14 h), and reduced intra-operative and total blood loss (by 63 and 119 mL, respectively). There was no difference noted in the incidence of dislocation, nerve injury, infection, or venous thromboembolic events [207].

Direct Anterior

Recently there has been widespread interest in the direct anterior approach to the hip for THA partially due to the belief that it reduces the risk of posterior dislocation by preserving the external rotators. This approach is a minimally invasive modification of the Smith-Peterson method that begins with an 8–10 cm incision from the anterior superior iliac spine in the direction of the lateral patella. It then exploits the inter-muscular and inter-nervous plane between the tensor fasciae latae (supplied by the superior gluteal nerve), and the sartorius muscle (innervated by branches of the femoral nerve). The rectus femoris is retracted medially and the iliopsoas is dissected away from the joint capsule. An arthrotomy is then performed to gain access to the joint. The obvious benefit of this approach is that no muscles are incised including the posterior structures that are important in the stability of the hip. Concerns surrounding this approach relate to its limited exposure of the femur. Detractors contend that this may lead to malposition of femoral implants or the use of implant designs that offer less bone fixation as compared with the conventional posterior approaches.

The majority of approaches to the hip require resection/splitting of muscle. The theoretical benefits of the anterior approach come mainly as a result of muscle preservation. By preserving the posterior structures and external rotators post-operative dislocation rates are theoretically reduced. The high degree of soft tissue preservation means that the normal post-operative hip precautions are more relaxed and restoration of function is earlier. A comparison of minimally invasive direct anterior versus posterior total hip arthroplasty found that serum inflammation and muscle damage markers were decreased in the direct-anterior-approach group as compared with the posterolateral approach group [208]. Menghini et al. used 12 cadaver hips to compare the degree of muscle damage caused by the anterior and posterior approaches. While the posterior approach caused damage to the gluteus medius and minimus (18 % vs 8 %), the anterior approach demonstrated a high degree of damage to the tensor fasciae latae muscle (mean of 31 %). There was also a need to transect the piriformis or conjoined tendon in 50 % of the anterior approaches to mobilize the femur, thus causing damage to precisely those structures that the anterior approach is designed to avoid [209].

Clinical outcome data has been mixed and has suggested short-term outcome improvement associated with the use of the direct anterior approach as compared with other approaches. Nakata et al. used a clinical comparative study of the direct anterior with mini-posterior approach for 195 hips. It was found that patients who received the direct anterior approach had a quicker recovery for hip function and gait stability [198]. Other studies confirmed these findings and also found decreased blood loss, less narcotic use, decreased pain scores after surgery, and less use of walking aids with the direct approach [200, 201, 210–213]. Other studies have contradicted these findings and reported increased or equivalent operating time, blood loss, and length of recovery [214–216]. A prospective randomized study by Restrepo et al. of 100 patients compared a modified Smith-Peterson approach to the direct

lateral approach. It was found that at 1 year the anterior approach group showed significantly better improvement in mental and physical health dimensions for the Short Form-36 and Western Ontario McMaster Osteoarthritis Index, however at 2 years these results were the same for both groups [217]. No study to date has proven that the long-term functional results of the direct anterior approach are superior to any other approach.

Whilst the anterior approach to the hip reduces the risk of damage to the muscles and sciatic nerve, there is a high intra-operative risk of damage to the lateral femoral cutaneous nerve. Goulding et al. followed 132 patients who underwent an anterior approach to the hip and found that 81 % reported varying degrees of lateral femoral cutaneous nerve neuropraxia. There was a higher risk of neuropraxia in those undergoing hip resurfacing as opposed to THA: 91 % and 67 %, respectively. Whilst only a small number of patients reported complete resolution of the neuropraxia, no patients reported functional limitation and the symptoms of neuropraxia were eventually reduced over time [218].

Reports of hip dislocation vary. Matta et al. studied 437 patients (494 hips), undergoing an anterior approach to primary THA and found the dislocation rate to be 0.61 % [219]. Sariali et al. used a prospective study of 1764 primary THA using the anterior approach and found the dislocation rate to be 1.5 % [220]. These numbers are comparable to the reported dislocation rate when using the posterolateral approach (1.2 %) [204]. More data is needed before one can say with certainty that the anterior approach reduces dislocation rates.

Concerns have been raised with regard to the exposure attained when using the anterior approach. Femoral exposure is limited and one study noted that periprosthetic femoral fractures went unnoticed during 1.65 % of procedures utilizing the direct anterior approach [221]. The result is that the use of intra-operative fluoroscopy is recommended in some centers which has the potential to increase operative times and raises the risk of contamination of the surgical field. Additionally, specialist tables are recommended for this approach which are costly and not widely available [204].

Anterolateral

The anterolateral approach is also commonly utilized in THR. This method provides an inter-muscular plane between the tensor fasciae lata and the gluteus medius. It is important to note that both of these muscles are innervated by the superior gluteal nerve and therefore this is not a true inter-nervous approach. One study found that at a median of 9.3 months follow-up 74 % of patients exhibited either atrophy or hypertrophy of the tensor fasciae latae and 42 % exhibited fat replacement on MRI [222]. The approach begins with an incision starting posterior and distal to the anterior superior iliac spine and running distal to become centered over the tip of the greater trochanter. After incising the fascia, an interval is developed between the tensor fasciae lata and the gluteus medius. The abductor mechanism, and the reflected head of the rectus femoris are incised while the psoas

tendon is retracted after which a capsulotomy is performed and the joint visualized. The theoretical advantages of this approach include a decreased risk of dislocation owing to the limited disruption of posterior structures and good visualization of the acetabulum.

Clinical data has again failed to show clear superiority or inferiority compared with other approaches. A previously mentioned nonrandomized clinical trial comparing anterolateral and posterior hip approaches at 5 years follow-up failed to note any differences in Oxford hip scores, dislocation, or revision rates between groups [203]. A randomized clinical trial comparing anterolateral and lateral approaches found improved gait mechanics at 6 weeks post-surgery but no difference in functional outcomes after 12 weeks [223]. Other case series have shown similar results although one suggested that the risk of varus femoral stem malalignment was higher with anterolateral as compared with lateral approaches [224]. Lateral approaches are similar to anterolateral but result in a split in the gluteus medius rather than exploiting the inter-muscular plane between the gluteus medius and tensor fascia lata. Other studies noted an improvement in patient-reported outcomes such as pain and limping in patients undergoing the anterolateral approach as compared with the direct lateral approach [225]. The incidence of dislocation in a meta-analysis of studies comparing various approaches of studies was approximately 2.18 % for this approach which puts it in line with other approaches [190]. The anterolateral group showed increased range of motion as compared to the transtrochanteric approach [226]. One observational study of the Swedish Hip Arthroplasty Register noted an increased risk of revision due to aseptic loosening of THAs which were implanted using the anterolateral approach as compared with a posterolateral approach (RR 1.3 CI 1.0–1.6) [227]. Other studies have noted an increased risk for abductor muscle avulsion using this approach with a subsequent need for reattachment [228–230].

Similarly to other approaches, minimally invasive options exist for the anterolateral approach. Comparative studies of conventional versus minimally invasive options were conflicting with regard to surgical time and blood loss [231, 232]. Studies comparing functional outcomes found that during the first year after surgery, patients with the mini-incision THA had significantly better hip muscle strength, walking speed, and functional score but after 1 year, the performance characteristics studied were statistically equivalent [233–235]. A study of gait mechanics comparing direct lateral, posterior, and anterolateral approaches failed to find significant differences between groups in stride length, step length, peak hip extension, and walking speed after total hip arthroplasty at 6 weeks or 1 year after surgery [236, 237].

Studies comparing the various minimally invasive approaches (two-incision, mini- posterior, and mini-anterolateral found no difference between the three minimally invasive approaches in early hospital discharge or early functional recovery utilizing a rapid rehabilitation protocol [238]. Similarly a study comparing a mini-incision anterolateral and direct lateral approaches found differences in muscle strength recovery and blood inflammatory markers in short-term follow-up but did not find any difference in the Harris hip score, pain visual analog scale, the Western

Ontario and McMaster Universities Osteoarthritis Index, and Medical Outcomes Study Short Form 36 score between the two groups throughout the 1-year study period [239, 240].

Summary

Many approaches exist to gain access to the hip joint, with each having their own advantages and disadvantages. Some require steep learning curves and so operative results between surgeons in these approaches differ greatly. While each surgeon has their own views on each approach, the posterior approach remains the gold standard, as it is easier to master and allows for increased exposure to the hip. However it is clear that all approaches are successful in experienced hands, with surgical ability having a great effect on patient outcomes. Care must be taken to ensure that adequate exposure and familiarity to the procedure are attained so that complication risk and patient morbidity can be kept to minimal.

Conclusion

Hip reconstruction is the subject of ongoing efforts to improve clinical outcomes. It is especially challenging to improve on a treatment that has already produced excellent results. THA is successful in 85–95 % of cases. Given this finding, many will ask: why fix what isn't broken? One reason is that the surgical volume of hip replacements is staggering. Over 500,000 THAs are performed annually in the United States [241]. Even if only 5 % of these fail, the result is a significant amount of burden on the healthcare system but more importantly on those patients who are unlucky enough to have a poor clinical result. Secondly, implant failure that results in revision is costly, technically difficult, and more likely to fail than a primary procedure. Lastly, while THR is used often, there are many debilitated patients who are currently not candidates for this procedure due to their young age. Significant improvements in implant longevity can have a tremendous impact on the lives of these individuals by returning function to their joints earlier and allowing them to resume their normal way of life. Still, the multitude of new technological options that exist for total hip arthroplasty greatly exceed the evidence supporting their use. New designs should be tried but all should be tested rigorously in order to come to find the optimal combination of principal components. Lastly, despite the emphasis on technology, one of the main contributors to the success of THA is surgical ability. Component wear, soft tissue damage, and implant stability have all been shown to be affected by surgical technique. Thus, improvement in total hip arthroplasty must come from advances in implant design, biomechanics, and surgical technique.

References

1. Chapman RH, Kowal SL, Cherry SB, Ferrufino CP, Roberts CS, Chen L. The modeled lifetime cost-effectiveness of published adherence-improving interventions for antihypertensive and lipid-lowering medications. Value Health. 2010;13(6):685–94.
2. Jenkins PJ, Clement ND, Hamilton DF, Gaston P, Patton JT, Howie CR. Predicting the cost-effectiveness of total hip and knee replacement: a health economic analysis. Bone Joint J. 2013;95-B(1):115–21.
3. Malchau H, Herberts P, Eisler T, Garellick G, Soderman P. The Swedish Total Hip Replacement Register. J Bone Joint Surg Am. 2002;84-A Suppl 2:2–20.
4. Bozic KJ, Kurtz SM, Lau E, Ong K, Chiu V, Vail TP, et al. The epidemiology of revision total knee arthroplasty in the United States. Clin Orthop Relat Res. 2010;468(1):45–51.
5. Charnley J. Low friction arthroplasty of the hip: theory and practice. Berlin: Springer; 1979.
6. Wroblewski BM, Siney PD, Dowson D, Collins SN. Prospective clinical and joint simulator studies of a new total hip arthroplasty using alumina ceramic heads and cross-linked polyethylene cups. J Bone Joint Surg Br. 1996;78(2):280–5.
7. Clarke IC, Good V, Anissian L, Gustafson A. Charnley wear model for validation of hip simulators–ball diameter versus polytetrafluoroethylene and polyethylene wear. Proc Inst Mech Eng H. 1997;211(1):25–36.
8. Sochart DH, Porter ML. The long-term results of Charnley low-friction arthroplasty in young patients who have congenital dislocation, degenerative osteoarthrosis, or rheumatoid arthritis. J Bone Joint Surg Am. 1997;79(11):1599–617.
9. Dowson D. New joints for the millennium: wear control in total replacement hip joints. Proc Inst Mech Eng H. 2001;215(4):335–58.
10. Archard JF. Contact and rubbing of a flat surface. J Appl Phys. 1953;24:981.
11. Livermore J, Ilstrup D, Morrey B. Effect of femoral head size on wear of the polyethylene acetabular component. J Bone Joint Surg Am. 1990;72(4):518–28.
12. Hall RM, Siney P, Unsworth A, Wroblewski BM. The association between rates of wear in retrieved acetabular components and the radius of the femoral head. Proc Inst Mech Eng H. 1998;212(5):321–6.
13. Galvin A, Kang L, Tipper J, Stone M, Ingham E, Jin Z, et al. Wear of crosslinked polyethylene under different tribological conditions. J Mater Sci Mater Med. 2006;17(3):235–43.
14. McKellop H, Shen FW, Lu B, Campbell P, Salovey R. Development of an extremely wear-resistant ultra high molecular weight polyethylene for total hip replacements. J Orthop Res. 1999;17(2):157–67.
15. Muratoglu OK, Bragdon CR, O'Connor DO, Jasty M, Harris WH. A novel method of cross-linking ultra-high-molecular-weight polyethylene to improve wear, reduce oxidation, and retain mechanical properties. Recipient of the 1999 HAP Paul Award. J Arthroplasty. 2001;16(2):149–60.
16. Geller JA, Malchau H, Bragdon C, Greene M, Harris WH, Freiberg AA. Large diameter femoral heads on highly cross-linked polyethylene: minimum 3-year results. Clin Orthop Relat Res. 2006;447:53–9.
17. Bragdon CR, Greene ME, Freiberg AA, Harris WH, Malchau H. Radiostereometric analysis comparison of wear of highly cross-linked polyethylene against 36- vs 28-mm femoral heads. J Arthroplasty. 2007;22(6 Suppl 2):125–9.
18. Lachiewicz PF, Heckman DS, Soileau ES, Mangla J, Martell JM. Femoral head size and wear of highly cross-linked polyethylene at 5 to 8 years. Clin Orthop Relat Res. 2009;467(12):3290–6.
19. Sakoda H, Voice AM, McEwen HM, Isaac GH, Hardaker C, Wroblewski BM, et al. A comparison of the wear and physical properties of silane cross-linked polyethylene and ultra-high molecular weight polyethylene. J Arthroplasty. 2001;16(8):1018–23.

20. Sorimachi T, Clarke IC, Williams PA, Gustafson A, Yamamoto K. Third-body abrasive wear challenge of 32 mm conventional and 44 mm highly crosslinked polyethylene liners in a hip simulator model. Proc Inst Mech Eng H. 2009;223(5):607–23.
21. McKellop H, Shen FW, DiMaio W, Lancaster JG. Wear of gamma-crosslinked polyethylene acetabular cups against roughened femoral balls. Clin Orthop Relat Res. 1999;369:73–82.
22. Nevelos J, Ingham E, Doyle C, Streicher R, Nevelos A, Walter W, et al. Microseparation of the centers of alumina-alumina artificial hip joints during simulator testing produces clinically relevant wear rates and patterns. J Arthroplasty. 2000;15(6):793–5.
23. Komistek RD, Dennis DA, Ochoa JA, Haas BD, Hammill C. In vivo comparison of hip separation after metal-on-metal or metal-on-polyethylene total hip arthroplasty. J Bone Joint Surg Am. 2002;84-A(10):1836–41.
24. Williams S, Butterfield M, Stewart T, Ingham E, Stone M, Fisher J. Wear and deformation of ceramic-on-polyethylene total hip replacements with joint laxity and swing phase microseparation. Proc Inst Mech Eng H. 2003;217(2):147–53.
25. Harris WH. Edge loading has a paradoxical effect on wear in metal-on-polyethylene total hip arthroplasties. Clin Orthop Relat Res. 2012;470(11):3077–82.
26. Green TR, Fisher J, Stone M, Wroblewski BM, Ingham E. Polyethylene particles of a 'critical size' are necessary for the induction of cytokines by macrophages in vitro. Biomaterials. 1998;19(24):2297–302.
27. Green TR, Fisher J, Matthews JB, Stone MH, Ingham E. Effect of size and dose on bone resorption activity of macrophages by in vitro clinically relevant ultra high molecular weight polyethylene particles. J Biomed Mater Res. 2000;53(5):490–7.
28. McKellop HA, Campbell P, Park SH, Schmalzried TP, Grigoris P, Amstutz HC, et al. The origin of submicron polyethylene wear debris in total hip arthroplasty. Clin Orthop Relat Res. 1995;311:3–20.
29. Endo MM, Barbour PS, Barton DC, Fisher J, Tipper JL, Ingham E, et al. Comparative wear and wear debris under three different counterface conditions of crosslinked and non-crosslinked ultra high molecular weight polyethylene. Biomed Mater Eng. 2001;11(1):23–35.
30. Kuzyk PR, Saccone M, Sprague S, Simunovic N, Bhandari M, Schemitsch EH. Cross-linked versus conventional polyethylene for total hip replacement: a meta-analysis of randomised controlled trials. J Bone Joint Surg Br. 2011;93(5):593–600.
31. Kurtz SM, Gawel HA, Patel JD. History and systematic review of wear and osteolysis outcomes for first-generation highly crosslinked polyethylene. Clin Orthop Relat Res. 2011;469(8):2262–77.
32. Thomas GE, Simpson DJ, Mehmood S, Taylor A, McLardy-Smith P, Gill HS, et al. The seven-year wear of highly cross-linked polyethylene in total hip arthroplasty: a double-blind, randomized controlled trial using radiostereometric analysis. J Bone Joint Surg Am. 2011;93(8):716–22.
33. Reynolds SE, Malkani AL, Ramakrishnan R, Yakkanti MR. Wear analysis of first-generation highly cross-linked polyethylene in primary total hip arthroplasty: an average 9-year follow-up. J Arthroplasty. 2012;27(6):1064–8.
34. Semlitsch M, Lehmann M, Weber H, Doerre E, Willert HG. New prospects for a prolonged functional life-span of artificial hip joints by using the material combination polyethylene/aluminium oxide ceramin/metal. J Biomed Mater Res. 1977;11(4):537–52.
35. Clarke IC, Good V, Williams P, Schroeder D, Anissian L, Stark A, et al. Ultra-low wear rates for rigid-on-rigid bearings in total hip replacements. Proc Inst Mech Eng H. 2000;214(4):331–47.
36. Clarke IC, Gustafson A. Clinical and hip simulator comparisons of ceramic-on-polyethylene and metal-on-polyethylene wear. Clin Orthop Relat Res. 2000;379:34–40.
37. Meftah M, Klingenstein GG, Yun RJ, Ranawat AS, Ranawat CS. Long-term performance of ceramic and metal femoral heads on conventional polyethylene in young and active patients: a matched-pair analysis. J Bone Joint Surg Am. 2013;95(13):1193–7.
38. Kadar T, Hallan G, Aamodt A, Indrekvam K, Badawy M, Skredderstuen A, et al. Wear and migration of highly cross-linked and conventional cemented polyethylene cups with cobalt

chrome or Oxinium femoral heads: a randomized radiostereometric study of 150 patients. J Orthop Res. 2011;29(8):1222–9.
39. Evangelista GT, Fulkerson E, Kummer F, Di Cesare PE. Surface damage to an Oxinium femoral head prosthesis after dislocation. J Bone Joint Surg Br. 2007;89(4):535–7.
40. McCalden RW, Charron KD, Davidson RD, Teeter MG, Holdsworth DW. Damage of an Oxinium femoral head and polyethylene liner following 'routine' total hip replacement. J Bone Joint Surg Br. 2011;93(3):409–13.
41. Jaffe WL, Strauss EJ, Cardinale M, Herrera L, Kummer FJ. Surface oxidized zirconium total hip arthroplasty head damage due to closed reduction effects on polyethylene wear. J Arthroplasty. 2009;24(6):898–902.
42. Wiles P. The surgery of the osteoarthritic hip. Br J Surg. 1958;45(193):488–97.
43. Fisher J, Jin Z, Tipper J, Stone M, Ingham E. Tribology of alternative bearings. Clin Orthop Relat Res. 2006;453:25–34.
44. Scholes SC, Unsworth A. The tribology of metal-on-metal total hip replacements. Proc Inst Mech Eng H. 2006;220(2):183–94.
45. Isaac GH, Thompson J, Williams S, Fisher J. Metal-on-metal bearings surfaces: materials, manufacture, design, optimization, and alternatives. Proc Inst Mech Eng H. 2006;220(2): 119–33.
46. Doorn PF, Campbell PA, Worrall J, Benya PD, McKellop HA, Amstutz HC. Metal wear particle characterization from metal on metal total hip replacements: transmission electron microscopy study of periprosthetic tissues and isolated particles. J Biomed Mater Res. 1998;42(1):103–11.
47. Jacobs JJ, Campbell PA, T Konttinen Y, Implant Wear Symposium 2007 Biologic Work Group. How has the biologic reaction to wear particles changed with newer bearing surfaces? J Am Acad Orthop Surg. 2008;16 Suppl 1:S49–55.
48. Lohmann CH, Schwartz Z, Koster G, Jahn U, Buchhorn GH, MacDougall MJ, et al. Phagocytosis of wear debris by osteoblasts affects differentiation and local factor production in a manner dependent on particle composition. Biomaterials. 2000;21(6):551–61.
49. Davies AP, Willert HG, Campbell PA, Learmonth ID, Case CP. An unusual lymphocytic perivascular infiltration in tissues around contemporary metal-on-metal joint replacements. J Bone Joint Surg Am. 2005;87(1):18–27.
50. Willert HG, Buchhorn GH, Fayyazi A, Flury R, Windler M, Koster G, et al. Metal-on-metal bearings and hypersensitivity in patients with artificial hip joints. A clinical and histomorphological study. J Bone Joint Surg Am. 2005;87(1):28–36.
51. Ingram JH, Stone M, Fisher J, Ingham E. The influence of molecular weight, crosslinking and counterface roughness on TNF-alpha production by macrophages in response to ultra high molecular weight polyethylene particles. Biomaterials. 2004;25(17):3511–22.
52. Doorn PF, Mirra JM, Campbell PA, Amstutz HC. Tissue reaction to metal on metal total hip prostheses. Clin Orthop Relat Res. 1996(329 Suppl):S187–205.
53. Watters TS, Cardona DM, Menon KS, Vinson EN, Bolognesi MP, Dodd LG. Aseptic lymphocyte-dominated vasculitis-associated lesion: a clinicopathologic review of an underrecognized cause of prosthetic failure. Am J Clin Pathol. 2010;134(6):886–93.
54. Mahendra G, Pandit H, Kliskey K, Murray D, Gill HS, Athanasou N. Necrotic and inflammatory changes in metal-on-metal resurfacing hip arthroplasties. Acta Orthop. 2009;80(6):653–9.
55. Glyn-Jones S, Pandit H, Kwon YM, Doll H, Gill HS, Murray DW. Risk factors for inflammatory pseudotumour formation following hip resurfacing. J Bone Joint Surg Br. 2009;91(12): 1566–74.
56. Urban RM, Jacobs JJ, Tomlinson MJ, Gavrilovic J, Black J, Peoc'h M. Dissemination of wear particles to the liver, spleen, and abdominal lymph nodes of patients with hip or knee replacement. J Bone Joint Surg Am. 2000;82(4):457–76.
57. Doherty AT, Howell RT, Ellis LA, Bisbinas I, Learmonth ID, Newson R, et al. Increased chromosome translocations and aneuploidy in peripheral blood lymphocytes of patients having revision arthroplasty of the hip. J Bone Joint Surg Br. 2001;83(7):1075–81.

58. Daniel J, Ziaee H, Pradhan C, McMinn DJ. Six-year results of a prospective study of metal ion levels in young patients with metal-on-metal hip resurfacings. J Bone Joint Surg Br. 2009;91(2):176–9.
59. Dunstan E, Sanghrajka AP, Tilley S, Unwin P, Blunn G, Cannon SR, et al. Metal ion levels after metal-on-metal proximal femoral replacements: a 30-year follow-up. J Bone Joint Surg Br. 2005;87(5):628–31.
60. Eswaramoorthy V, Moonot P, Kalairajah Y, Biant LC, Field RE. The Metasul metal-on-metal articulation in primary total hip replacement: clinical and radiological results at ten years. J Bone Joint Surg Br. 2008;90(10):1278–83.
61. Randelli F, Banci L, D'Anna A, Visentin O, Randelli G. Cementless Metasul metal-on-metal total hip arthroplasties at 13 years. J Arthroplasty. 2012;27(2):186–92.
62. Migaud H, Putman S, Krantz N, Vasseur L, Girard J. Cementless metal-on-metal versus ceramic-on-polyethylene hip arthroplasty in patients less than fifty years of age: a comparative study with twelve to fourteen-year follow-up. J Bone Joint Surg Am. 2011;93 Suppl 2:137–42.
63. Dorr LD, Wan Z, Longjohn DB, Dubois B, Murken R. Total hip arthroplasty with use of the Metasul metal-on-metal articulation. Four to seven-year results. J Bone Joint Surg Am. 2000;82(6):789–98.
64. Saito S, Ishii T, Mori S, Hosaka K, Ootaki M, Tokuhashi Y. Long-term results of metasul metal-on-metal total hip arthroplasty. Orthopedics. 2010;33(8).
65. Haddad FS, Thakrar RR, Hart AJ, Skinner JA, Nargol AV, Nolan JF, et al. Metal-on-metal bearings: the evidence so far. J Bone Joint Surg Br. 2011;93(5):572–9.
66. Angadji A, Royle M, Collins SN, Shelton JC. Influence of cup orientation on the wear performance of metal-on-metal hip replacements. Proc Inst Mech Eng H. 2009;223(4):449–57.
67. Campbell P, Beaule PE, Ebramzadeh E, Le Duff MJ, De Smet K, Lu Z, et al. The John Charnley Award: a study of implant failure in metal-on-metal surface arthroplasties. Clin Orthop Relat Res. 2006;453:35–46.
68. Hart AJ, Buddhdev P, Winship P, Faria N, Powell JJ, Skinner JA. Cup inclination angle of greater than 50 degrees increases whole blood concentrations of cobalt and chromium ions after metal-on-metal hip resurfacing. Hip Int. 2008;18(3):212–9.
69. Bosker BH, Ettema HB, Boomsma MF, Kollen BJ, Maas M, Verheyen CC. High incidence of pseudotumour formation after large-diameter metal-on-metal total hip replacement: a prospective cohort study. J Bone Joint Surg Br. 2012;94(6):755–61.
70. Skinner J, Kay P. Commentary: metal on metal hips. BMJ. 2011;342:d3009.
71. Nevelos JE, Ingham E, Doyle C, Nevelos AB, Fisher J. Wear of HIPed and non-HIPed alumina-alumina hip joints under standard and severe simulator testing conditions. Biomaterials. 2001;22(16):2191–7.
72. Al-Hajjar M, Leslie IJ, Tipper J, Williams S, Fisher J, Jennings LM. Effect of cup inclination angle during microseparation and rim loading on the wear of BIOLOX(R) delta ceramic-on-ceramic total hip replacement. J Biomed Mater Res B Appl Biomater. 2010;95(2):263–8.
73. Tipper JL, Hatton A, Nevelos JE, Ingham E, Doyle C, Streicher R, et al. Alumina-alumina artificial hip joints. Part II: characterisation of the wear debris from in vitro hip joint simulations. Biomaterials. 2002;23(16):3441–8.
74. Wimmer MA, Nassutt R, Sprecher C, Loos J, Tager G, Fischer A. Investigation on stick phenomena in metal-on-metal hip joints after resting periods. Proc Inst Mech Eng H. 2006;220(2):219–27.
75. Brockett C, Williams S, Jin Z, Isaac G, Fisher J. Friction of total hip replacements with different bearings and loading conditions. J Biomed Mater Res B Appl Biomater. 2007;81(2): 508–15.
76. Bizot P, Nizard R, Hamadouche M, Hannouche D, Sedel L. Prevention of wear and osteolysis: alumina-on-alumina bearing. Clin Orthop Relat Res. 2001;393:85–93.
77. Choi IY, Kim YS, Hwang KT, Kim YH. Incidence and factors associated with squeaking in alumina-on-alumina THA. Clin Orthop Relat Res. 2010;468(12):3234–9.

78. Hannouche D, Nich C, Bizot P, Meunier A, Nizard R, Sedel L. Fractures of ceramic bearings: history and present status. Clin Orthop Relat Res. 2003;417:19–26.
79. Petsatodis GE, Papadopoulos PP, Papavasiliou KA, Hatzokos IG, Agathangelidis FG, Christodoulou AG. Primary cementless total hip arthroplasty with an alumina ceramic-on-ceramic bearing: results after a minimum of twenty years of follow-up. J Bone Joint Surg Am. 2010;92(3):639–44.
80. Capello WN, D'Antonio JA, Feinberg JR, Manley MT, Naughton M. Ceramic-on-ceramic total hip arthroplasty: update. J Arthroplasty. 2008;23(7 Suppl):39–43.
81. Brockett CL, Harper P, Williams S, Isaac GH, Dwyer-Joyce RS, Jin Z, et al. The influence of clearance on friction, lubrication and squeaking in large diameter metal-on-metal hip replacements. J Mater Sci Mater Med. 2008;19(4):1575–9.
82. Back DL, Dalziel R, Young D, Shimmin A. Early results of primary Birmingham hip resurfacings. An independent prospective study of the first 230 hips. J Bone Joint Surg Br. 2005;87(3):324–9.
83. Jarrett CA, Ranawat AS, Bruzzone M, Blum YC, Rodriguez JA, Ranawat CS. The squeaking hip: a phenomenon of ceramic-on-ceramic total hip arthroplasty. J Bone Joint Surg Am. 2009;91(6):1344–9.
84. Matar WY, Restrepo C, Parvizi J, Kurtz SM, Hozack WJ. Revision hip arthroplasty for ceramic-on-ceramic squeaking hips does not compromise the results. J Arthroplasty. 2010;25(6 Suppl):81–6.
85. Stanat SJ, Capozzi JD. Squeaking in third- and fourth-generation ceramic-on-ceramic total hip arthroplasty: meta-analysis and systematic review. J Arthroplasty. 2012;27(3):445–53.
86. Brockett CL, Williams S, Jin Z, Isaac GH, Fisher J. Squeaking hip arthroplasties: a tribological phenomenon. J Arthroplasty. 2013;28(1):90–7.
87. Walter WL, O'Toole GC, Walter WK, Ellis A, Zicat BA. Squeaking in ceramic-on-ceramic hips: the importance of acetabular component orientation. J Arthroplasty. 2007;22(4): 496–503.
88. Chevillotte C, Pibarot V, Carret JP, Bejui-Hugues J, Guyen O. Hip squeaking: a 10-year follow-up study. J Arthroplasty. 2012;27(6):1008–13.
89. Restrepo C, Matar WY, Parvizi J, Rothman RH, Hozack WJ. Natural history of squeaking after total hip arthroplasty. Clin Orthop Relat Res. 2010;468(9):2340–5.
90. Keurentjes JC, Kuipers RM, Wever DJ, Schreurs BW. High incidence of squeaking in THAs with alumina ceramic-on-ceramic bearings. Clin Orthop Relat Res. 2008;466(6):1438–43.
91. Emms NW, Stockley I, Hamer AJ, Wilkinson JM. Long-term outcome of a cementless, hemispherical, press-fit acetabular component: survivorship analysis and dose–response relationship to linear polyethylene wear. J Bone Joint Surg Br. 2010;92(6):856–61.
92. De Man FH, Tigchelaar W, Marti RK, Van Noorden CJ, Van der Vis HM. Effects of mechanical compression of a fibrous tissue interface on bone with or without high-density polyethylene particles in a rabbit model of prosthetic loosening. J Bone Joint Surg Am. 2005;87(7): 1522–33.
93. McCombe P, Williams SA. A comparison of polyethylene wear rates between cemented and cementless cups. A prospective, randomised trial. J Bone Joint Surg Br. 2004;86(3):344–9.
94. Wilkinson JM, Hamer AJ, Stockley I, Eastell R. Polyethylene wear rate and osteolysis: critical threshold versus continuous dose–response relationship. J Orthop Res. 2005;23(3): 520–5.
95. Morshed S, Bozic KJ, Ries MD, Malchau H, Colford Jr JM. Comparison of cemented and uncemented fixation in total hip replacement: a meta-analysis. Acta Orthop. 2007;78(3):315–26.
96. Yahiro MA, Gantenberg JB, Nelson R, Lu HT, Mishra NK. Comparison of the results of cemented, porous-ingrowth, and threaded acetabular cup fixation. A meta-analysis of the orthopaedic literature. J Arthroplasty. 1995;10(3):339–50.
97. Toossi N, Adeli B, Timperley AJ, Haddad FS, Maltenfort M, Parvizi J. Acetabular components in total hip arthroplasty: is there evidence that cementless fixation is better? J Bone Joint Surg Am. 2013;95(2):168–74.

98. Sumner DR, Virdi AS, Leven RM, Healy KE. Enhancing cementless fixation. In: Shanbhag A, Rubash HE, Jacobs JJ, editors. Joint replacement and bone resorption: pathology, biomaterials, and clinical practice. Boca Raton: CRC Press; 2005. p. 727–54.
99. Bobyn JD, Engh CA, Glassman AH. Radiography and histology of a threaded acetabular implant. One case studied at two years. J Bone Joint Surg Br. 1988;70(2):302–4.
100. Pupparo F, Engh CA. Comparison of porous-threaded and smooth-threaded acetabular components of identical design. Two- to four-year results. Clin Orthop Relat Res. 1991; 271:201–6.
101. Aigner C. 10 years results with the corund-blasted Zweymuller titanium alloy threaded acetabular cup. Z Orthop Ihre Grenzgeb. 1998;136(2):110–4.
102. Manley MT, Capello WN, D'Antonio JA, Edidin AA, Geesink RG. Fixation of acetabular cups without cement in total hip arthroplasty. A comparison of three different implant surfaces at a minimum duration of follow-up of five years. J Bone Joint Surg Am. 1998;80(8):1175–85.
103. Welsh RP, Pilliar RM, Macnab I. Surgical implants. The role of surface porosity in fixation to bone and acrylic. J Bone Joint Surg Am. 1971;53(5):963–77.
104. Bobyn JD, Pilliar RM, Cameron HU, Weatherly GC. The optimum pore size for the fixation of porous-surfaced metal implants by the ingrowth of bone. Clin Orthop Relat Res. 1980;150:263–70.
105. Bragdon CR, Burke D, Lowenstein JD, O'Connor DO, Ramamurti B, Jasty M, et al. Differences in stiffness of the interface between a cementless porous implant and cancellous bone in vivo in dogs due to varying amounts of implant motion. J Arthroplasty. 1996;11(8):945–51.
106. Goldring SR, Jasty M, Roelke MS, Rourke CM, Bringhurst FR, Harris WH. Formation of a synovial-like membrane at the bone-cement interface. Its role in bone resorption and implant loosening after total hip replacement. Arthritis Rheum. 1986;29(7):836–42.
107. Goldring SR, Schiller AL, Roelke M, Rourke CM, O'Neil DA, Harris WH. The synovial-like membrane at the bone-cement interface in loose total hip replacements and its proposed role in bone lysis. J Bone Joint Surg Am. 1983;65(5):575–84.
108. Sandborn PM, Cook SD, Spires WP, Kester MA. Tissue response to porous-coated implants lacking initial bone apposition. J Arthroplasty. 1988;3(4):337–46.
109. Dalton JE, Cook SD, Thomas KA, Kay JF. The effect of operative fit and hydroxyapatite coating on the mechanical and biological response to porous implants. J Bone Joint Surg Am. 1995;77(1):97–110.
110. Schmalzried TP, Wessinger SJ, Hill GE, Harris WH. The Harris-Galante porous acetabular component press-fit without screw fixation. Five-year radiographic analysis of primary cases. J Arthroplasty. 1994;9(3):235–42.
111. Stiehl JB, MacMillan E, Skrade DA. Mechanical stability of porous-coated acetabular components in total hip arthroplasty. J Arthroplasty. 1991;6(4):295–300.
112. Capello WN, D'Antonio JA, Manley MT, Feinberg JR. Hydroxyapatite in total hip arthroplasty. Clinical results and critical issues. Clin Orthop Relat Res. 1998;355:200–11.
113. Overgaard S, Lind M, Glerup H, Grundvig S, Bunger C, Soballe K. Hydroxyapatite and fluorapatite coatings for fixation of weight loaded implants. Clin Orthop Relat Res. 1997;336:286–96.
114. Thanner J, Karrholm J, Herberts P, Malchau H. Porous cups with and without hydroxylapatite-tricalcium phosphate coating: 23 matched pairs evaluated with radiostereometry. J Arthroplasty. 1999;14(3):266–71.
115. Valancius K, Soballe K, Nielsen PT, Laursen MB. No superior performance of hydroxyapatite-coated acetabular cups over porous-coated cups. Acta Orthop. 2013;84(6):544–8.
116. Amstutz HC, Lodwig RM, Schurman DJ, Hodgson AG. Range of motion studies for total hip replacements. A comparative study with a new experimental apparatus. Clin Orthop Relat Res. 1975;111:124–30.
117. Chandler DR, Glousman R, Hull D, McGuire PJ, Kim IS, Clarke IC, et al. Prosthetic hip range of motion and impingement. The effects of head and neck geometry. Clin Orthop Relat Res. 1982;166:284–91.

118. Kiguchi K, Horie T, Yamashita A, Ueno M, Kobayashi T, Mawatari M, et al. A study of the effect of the femoral head diameter on prosthetic hip joint dislocation using a hip-joint motion simulator. Conf Proc IEEE Eng Med Biol Soc. 2009;2009:6058–61.
119. Cross MB, Nam D, Mayman DJ. Ideal femoral head size in total hip arthroplasty balances stability and volumetric wear. HSS J. 2012;8(3):270–4.
120. Kostensalo I, Junnila M, Virolainen P, Remes V, Matilainen M, Vahlberg T, et al. Effect of femoral head size on risk of revision for dislocation after total hip arthroplasty: a population-based analysis of 42,379 primary procedures from the Finnish Arthroplasty Register. Acta Orthop. 2013;84(4):342–7.
121. Hailer NP, Weiss RJ, Stark A, Karrholm J. The risk of revision due to dislocation after total hip arthroplasty depends on surgical approach, femoral head size, sex, and primary diagnosis. An analysis of 78,098 operations in the Swedish Hip Arthroplasty Register. Acta Orthop. 2012;83(5):442–8.
122. Tsertsvadze A, Grove A, Freeman K, Court R, Johnson S, Connock M, et al. Total hip replacement for the treatment of end stage arthritis of the hip: a systematic review and meta-analysis. PLoS One. 2014;9(7), e99804.
123. Dyrkacz RMR, Turgeon T, Ojo O, Brandt JM, Wyss U, editor. Head size affects corrosion behavior in artificial hip joints. 2012 Orthopaedic Research Society; 2012.
124. Johnston RC, Smidt GL. Hip motion measurements for selected activities of daily living. Clin Orthop Relat Res. 1970;72:205–15.
125. Dorr LD, Kane 3rd TJ, Conaty JP. Long-term results of cemented total hip arthroplasty in patients 45 years old or younger. A 16-year follow-up study. J Arthroplasty. 1994;9(5):453–6.
126. Jaffe WL, Jarolem KL. Normalized and proportionalized cemented femoral stem designs. A 10-year clinical study. J Arthroplasty. 1995;10 Suppl:S39–44.
127. Ayers D, Mann K. The importance of proximal cement filling of the calcar region: a biomechanical justification. J Arthroplasty. 2003;18(7 Suppl 1):103–9.
128. Harrigan TP, Kareh JA, O'Connor DO, Burke DW, Harris WH. A finite element study of the initiation of failure of fixation in cemented femoral total hip components. J Orthop Res. 1992;10(1):134–44.
129. Jasty M, Maloney WJ, Bragdon CR, O'Connor DO, Haire T, Harris WH. The initiation of failure in cemented femoral components of hip arthroplasties. J Bone Joint Surg Br. 1991;73(4):551–8.
130. Maloney WJ, Jasty M, Burke DW, O'Connor DO, Zalenski EB, Bragdon C, et al. Biomechanical and histologic investigation of cemented total hip arthroplasties. A study of autopsy-retrieved femurs after in vivo cycling. Clin Orthop Relat Res. 1989;249:129–40.
131. Jasty M, Jiranek W, Harris WH. Acrylic fragmentation in total hip replacements and its biological consequences. Clin Orthop Relat Res. 1992;285:116–28.
132. Tapadiya D, Walker RH, Schurman DJ. Prediction of outcome of total hip arthroplasty based on initial postoperative radiographic analysis. Matched, paired comparisons of failed versus successful femoral components. Clin Orthop Relat Res. 1984;186:5–15.
133. Callaghan JJ, Salvati EA, Pellicci PM, Wilson Jr PD, Ranawat CS. Results of revision for mechanical failure after cemented total hip replacement, 1979 to 1982. A two to five-year follow-up. J Bone Joint Surg Am. 1985;67(7):1074–85.
134. Ebramzadeh E, Sarmiento A, McKellop HA, Llinas A, Gogan W. The cement mantle in total hip arthroplasty. Analysis of long-term radiographic results. J Bone Joint Surg Am. 1994;76(1):77–87.
135. Barrack RL, Mulroy Jr RD, Harris WH. Improved cementing techniques and femoral component loosening in young patients with hip arthroplasty. A 12-year radiographic review. J Bone Joint Surg Br. 1992;74(3):385–9.
136. Schurman DJ, Bloch DA, Segal MR, Tanner CM. Conventional cemented total hip arthroplasty. Assessment of clinical factors associated with revision for mechanical failure. Clin Orthop Relat Res. 1989;240:173–80.
137. Ranawat CS, Deshmukh RG, Peters LE, Umlas ME. Prediction of the long-term durability of all-polyethylene cemented sockets. Clin Orthop Relat Res. 1995;317:89–105.

138. Madey SM, Callaghan JJ, Olejniczak JP, Goetz DD, Johnston RC. Charnley total hip arthroplasty with use of improved techniques of cementing. The results after a minimum of fifteen years of follow-up. J Bone Joint Surg Am. 1997;79(1):53–64.
139. Smith SW, Estok 2nd DM, Harris WH. Total hip arthroplasty with use of second-generation cementing techniques. An eighteen-year-average follow-up study. J Bone Joint Surg Am. 1998;80(11):1632–40.
140. Sanchez-Sotelo J, Berry DJ, Harmsen S. Long-term results of use of a collared matte-finished femoral component fixed with second-generation cementing techniques. A fifteen-year-median follow-up study. J Bone Joint Surg Am. 2002;84-A(9):1636–41.
141. Rasquinha VJ, Dua V, Rodriguez JA, Ranawat CS. Fifteen-year survivorship of a collarless, cemented, normalized femoral stem in primary hybrid total hip arthroplasty with a modified third-generation cement technique. J Arthroplasty. 2003;18(7 Suppl 1):86–94.
142. Mallory TH, Head WC, Lombardi Jr AV. Tapered design for the cementless total hip arthroplasty femoral component. Clin Orthop Relat Res. 1997;344:172–8.
143. Wagner H, Wagner M. Cone prosthesis for the hip joint. Arch Orthop Trauma Surg. 2000;120(1–2):88–95.
144. Rorabeck CH, Bourne RB, Laupacis A, Feeny D, Wong C, Tugwell P, et al. A double-blind study of 250 cases comparing cemented with cementless total hip arthroplasty. Cost-effectiveness and its impact on health-related quality of life. Clin Orthop Relat Res. 1994;298:156–64.
145. Aldinger PR, Breusch SJ, Lukoschek M, Mau H, Ewerbeck V, Thomsen M. A ten- to 15-year follow-up of the cementless spotorno stem. J Bone Joint Surg Br. 2003;85(2):209–14.
146. Capello WN, D'Antonio JA, Feinberg JR, Manley MT. Ten-year results with hydroxyapatite-coated total hip femoral components in patients less than fifty years old. A concise follow-up of a previous report. J Bone Joint Surg Am. 2003;85-A(5):885–9.
147. Grubl A, Chiari C, Gruber M, Kaider A, Gottsauner-Wolf F. Cementless total hip arthroplasty with a tapered, rectangular titanium stem and a threaded cup: a minimum ten-year follow-up. J Bone Joint Surg Am. 2002;84-A(3):425–31.
148. Kawamura H, Dunbar MJ, Murray P, Bourne RB, Rorabeck CH. The porous coated anatomic total hip replacement. A ten to fourteen-year follow-up study of a cementless total hip arthroplasty. J Bone Joint Surg Am. 2001;83-A(9):1333–8.
149. Bourne RB, Rorabeck CH, Patterson JJ, Guerin J. Tapered titanium cementless total hip replacements: a 10- to 13-year followup study. Clin Orthop Relat Res. 2001;393:112–20.
150. Kolb A, Grubl A, Schneckener CD, Chiari C, Kaider A, Lass R, et al. Cementless total hip arthroplasty with the rectangular titanium Zweymuller stem: a concise follow-up, at a minimum of twenty years, of previous reports. J Bone Joint Surg Am. 2012;94(18):1681–4.
151. Streit MR, Innmann MM, Merle C, Bruckner T, Aldinger PR, Gotterbarm T. Long-term (20- to 25-year) results of an uncemented tapered titanium femoral component and factors affecting survivorship. Clin Orthop Relat Res. 2013;471(10):3262–9.
152. Ateschrang A, Weise K, Weller S, Stockle U, de Zwart P, Ochs BG. Long-term results using the straight tapered femoral cementless hip stem in total hip arthroplasty: a minimum of twenty-year follow-up. J Arthroplasty. 2014;29(8):1559–65.
153. McLaughlin JR, Lee KR. Uncemented total hip arthroplasty using a tapered femoral component in obese patients: an 18–27 year follow-up study. J Arthroplasty. 2014;29(7):1365–8.
154. Sychterz CJ, Engh CA. The influence of clinical factors on periprosthetic bone remodeling. Clin Orthop Relat Res. 1996;322:285–92.
155. Callaghan JJ, Templeton JE, Liu SS, Warth LC, Chung YY. Improved results using extensively coated THA stems at minimum 5-year followup. Clin Orthop Relat Res. 2006;453:91–6.
156. Engh CA, Massin P, Suthers KE. Roentgenographic assessment of the biologic fixation of porous-surfaced femoral components. Clin Orthop Relat Res. 1990;257:107–28.
157. Kronick JL, Barba ML, Paprosky WG. Extensively coated femoral components in young patients. Clin Orthop Relat Res. 1997;344:263–74.

158. Faraj AA, Yousuf M. Anterior thigh pain after cementless total hip arthroplasty. Int Orthop. 2005;29(3):149–51.
159. Kinov P, Radl R, Zacherl M, Leithner A, Windhager R. Correlation between thigh pain and radiological findings with a proximally porous-coated stem. Acta Orthop Belg. 2007;73(5): 618–24.
160. Loupasis G, Hyde ID, Morris EW. The Furlong hydroxyapatite-coated femoral prosthesis. A 4- to 7-year follow-up study. Arch Orthop Trauma Surg. 1998;117(3):132–5.
161. McAuley JP, Culpepper WJ, Engh CA. Total hip arthroplasty. Concerns with extensively porous coated femoral components. Clin Orthop Relat Res. 1998;355:182–8.
162. MacDonald SJ, Rosenzweig S, Guerin JS, McCalden RW, Bohm ER, Bourne RB, et al. Proximally versus fully porous-coated femoral stems: a multicenter randomized trial. Clin Orthop Relat Res. 2010;468(2):424–32.
163. Srinivasan A, Jung E, Levine BR. Modularity of the femoral component in total hip arthroplasty. J Am Acad Orthop Surg. 2012;20(4):214–22.
164. Spitzer AI. The S-ROM cementless femoral stem: history and literature review. Orthopedics. 2005;28(9 Suppl):s1117–24.
165. Restrepo C, Mashadi M, Parvizi J, Austin MS, Hozack WJ. Modular femoral stems for revision total hip arthroplasty. Clin Orthop Relat Res. 2011;469(2):476–82.
166. Cooper HJ, Urban RM, Wixson RL, Meneghini RM, Jacobs JJ. Adverse local tissue reaction arising from corrosion at the femoral neck-body junction in a dual-taper stem with a cobalt-chromium modular neck. J Bone Joint Surg Am. 2013;95(10):865–72.
167. Kop AM, Keogh C, Swarts E. Proximal component modularity in THA—at what cost? An implant retrieval study. Clin Orthop Relat Res. 2012;470(7):1885–94.
168. Kop AM, Swarts E. Corrosion of a hip stem with a modular neck taper junction: a retrieval study of 16 cases. J Arthroplasty. 2009;24(7):1019–23.
169. Gilbert JL, Buckley CA, Jacobs JJ. In vivo corrosion of modular hip prosthesis components in mixed and similar metal combinations. The effect of crevice, stress, motion, and alloy coupling. J Biomed Mater Res. 1993;27(12):1533–44.
170. Wright G, Sporer S, Urban R, Jacobs J. Fracture of a modular femoral neck after total hip arthroplasty: a case report. J Bone Joint Surg Am. 2010;92(6):1518–21.
171. Grigoris P, Roberts P, Panousis K, Jin Z. Hip resurfacing arthroplasty: the evolution of contemporary designs. Proc Inst Mech Eng H. 2006;220(2):95–105.
172. Marker DR, Seyler TM, Jinnah RH, Delanois RE, Ulrich SD, Mont MA. Femoral neck fractures after metal-on-metal total hip resurfacing: a prospective cohort study. J Arthroplasty. 2007;22(7 Suppl 3):66–71.
173. Kluess D, Zietz C, Lindner T, Mittelmeier W, Schmitz KP, Bader R. Limited range of motion of hip resurfacing arthroplasty due to unfavorable ratio of prosthetic head size and femoral neck diameter. Acta Orthop. 2008;79(6):748–54.
174. Bengs BC, Sangiorgio SN, Ebramzadeh E. Less range of motion with resurfacing arthroplasty than with total hip arthroplasty: in vitro examination of 8 designs. Acta Orthop. 2008;79(6):755–62.
175. Le Duff MJ, Wisk LE, Amstutz HC. Range of motion after stemmed total hip arthroplasty and hip resurfacing – a clinical study. Bull NYU Hosp Jt Dis. 2009;67(2):177–81.
176. Shimmin AJ, Bare JV. Comparison of functional results of hip resurfacing and total hip replacement: a review of the literature. Orthop Clin North Am. 2011;42(2):143–51. 2.
177. Springer BD, Connelly SE, Odum SM, Fehring TK, Griffin WL, Mason JB, et al. Cementless femoral components in young patients: review and meta-analysis of total hip arthroplasty and hip resurfacing. J Arthroplasty. 2009;24(6 Suppl):2–8.
178. Johanson PE, Fenstad AM, Furnes O, Garellick G, Havelin LI, Overgaard S, et al. Inferior outcome after hip resurfacing arthroplasty than after conventional arthroplasty. Evidence from the Nordic Arthroplasty Register Association (NARA) database, 1995 to 2007. Acta Orthop. 2010;81(5):535–41.

179. Seppanen M, Makela K, Virolainen P, Remes V, Pulkkinen P, Eskelinen A. Hip resurfacing arthroplasty: short-term survivorship of 4,401 hips from the Finnish Arthroplasty Register. Acta Orthop. 2012;83(3):207–13.
180. de Steiger RN, Hang JR, Miller LN, Graves SE, Davidson DC. Five-year results of the ASR XL Acetabular System and the ASR Hip Resurfacing System: an analysis from the Australian Orthopaedic Association National Joint Replacement Registry. J Bone Joint Surg Am. 2011;93(24):2287–93.
181. Australian Orthopaedic Association National Joint Replacement Registry. Annual report. Adelaide: Australian Orthopaedic Association 2007.
182. Berry DJ, Berger RA, Callaghan JJ, Dorr LD, Duwelius PJ, Hartzband MA, et al. Minimally invasive total hip arthroplasty. Development, early results, and a critical analysis. Presented at the Annual Meeting of the American Orthopaedic Association, Charleston, South Carolina, USA, June 14, 2003. J Bone Joint Surg Am. 2003;85-A(11):2235–46.
183. Bertin KC, Rottinger H. Anterolateral mini-incision hip replacement surgery: a modified Watson-Jones approach. Clin Orthop Relat Res. 2004;429:248–55.
184. Baker AS, Bitounis VC. Abductor function after total hip replacement. An electromyographic and clinical review. J Bone Joint Surg Br. 1989;71(1):47–50.
185. Gore DR, Murray MP, Sepic SB, Gardner GM. Anterolateral compared to posterior approach in total hip arthroplasty: differences in component positioning, hip strength, and hip motion. Clin Orthop Relat Res. 1982;165:180–7.
186. Ritter MA, Harty LD, Keating ME, Faris PM, Meding JB. A clinical comparison of the anterolateral and posterolateral approaches to the hip. Clin Orthop Relat Res. 2001;385: 95–9.
187. Pellicci PM, Bostrom M, Poss R. Posterior approach to total hip replacement using enhanced posterior soft tissue repair. Clin Orthop Relat Res. 1998;355:224–8.
188. Chiu FY, Chen CM, Chung TY, Lo WH, Chen TH. The effect of posterior capsulorrhaphy in primary total hip arthroplasty: a prospective randomized study. J Arthroplasty. 2000;15(2): 194–9.
189. White Jr RE, Forness TJ, Allman JK, Junick DW. Effect of posterior capsular repair on early dislocation in primary total hip replacement. Clin Orthop Relat Res. 2001;393:163–7.
190. Masonis JL, Bourne RB. Surgical approach, abductor function, and total hip arthroplasty dislocation. Clin Orthop Relat Res. 2002;405:46–53.
191. Woolson ST, Rahimtoola ZO. Risk factors for dislocation during the first 3 months after primary total hip replacement. J Arthroplasty. 1999;14(6):662–8.
192. Woo RY, Morrey BF. Dislocations after total hip arthroplasty. J Bone Joint Surg Am. 1982;64(9):1295–306.
193. Paterno SA, Lachiewicz PF, Kelley SS. The influence of patient-related factors and the position of the acetabular component on the rate of dislocation after total hip replacement. J Bone Joint Surg Am. 1997;79(8):1202–10.
194. Li E, Meding JB, Ritter MA, Keating EM, Faris PM. The natural history of a posteriorly dislocated total hip replacement. J Arthroplasty. 1999;14(8):964–8.
195. McCollum DE, Gray WJ. Dislocation after total hip arthroplasty. Causes and prevention. Clin Orthop Relat Res. 1990;261:159–70.
196. Jolles BM, Bogoch ER. Posterior versus lateral surgical approach for total hip arthroplasty in adults with osteoarthritis. Cochrane Database Syst Rev. 2006;3, CD003828.
197. Christensen CP, Karthikeyan T, Jacobs CA. Greater prevalence of wound complications requiring reoperation with direct anterior approach total hip arthroplasty. J Arthroplasty. 2014;29:1839–41.
198. Nakata K, Nishikawa M, Yamamoto K, Hirota S, Yoshikawa H. A clinical comparative study of the direct anterior with mini-posterior approach: two consecutive series. J Arthroplasty. 2009;24(5):698–704.
199. Schweppe ML, Seyler TM, Plate JF, Swenson RD, Lang JE. Does surgical approach in total hip arthroplasty affect rehabilitation, discharge disposition, and readmission rate? Surg Technol Int. 2013;23:219–27.

200. Taunton MJ, Mason JB, Odum SM, Springer BD. Direct anterior total hip arthroplasty yields more rapid voluntary cessation of all walking aids: a prospective, randomized clinical trial. J Arthroplasty. 2014;29:169–72.
201. Zawadsky MW, Paulus MC, Murray PJ, Johansen MA. Early outcome comparison between the direct anterior approach and the mini-incision posterior approach for primary total hip arthroplasty: 150 consecutive cases. J Arthroplasty. 2014;29(6):1256–60.
202. Rathod PA, Orishimo KF, Kremenic IJ, Deshmukh AJ, Rodriguez JA. Similar improvement in gait parameters following direct anterior & posterior approach total hip arthroplasty. J Arthroplasty. 2014;29(6):1261–4.
203. Palan J, Beard DJ, Murray DW, Andrew JG, Nolan J. Which approach for total hip arthroplasty: anterolateral or posterior? Clin Orthop Relat Res. 2009;467(2):473–7.
204. Sculco TP. Anterior approach in THA improves outcomes: opposes. Orthopedics. 2011;34(9):e459–61.
205. Chimento GF, Pavone V, Sharrock N, Kahn B, Cahill J, Sculco TP. Minimally invasive total hip arthroplasty: a prospective randomized study. J Arthroplasty. 2005;20(2):139–44.
206. Cheng T, Feng JG, Liu T, Zhang XL. Minimally invasive total hip arthroplasty: a systematic review. Int Orthop. 2009;33(6):1473–81.
207. Berstock JR, Blom AW, Beswick AD. A systematic review and meta-analysis of the standard versus mini-incision posterior approach to total hip arthroplasty. J Arthroplasty. 2014;29: 1970–82.
208. Bergin PF, Doppelt JD, Kephart CJ, Benke MT, Graeter JH, Holmes AS, et al. Comparison of minimally invasive direct anterior versus posterior total hip arthroplasty based on inflammation and muscle damage markers. J Bone Joint Surg Am. 2011;93(15):1392–8.
209. Meneghini RM, Pagnano MW, Trousdale RT, Hozack WJ. Muscle damage during MIS total hip arthroplasty: Smith-Petersen versus posterior approach. Clin Orthop Relat Res. 2006;453:293–8.
210. Alecci V, Valente M, Crucil M, Minerva M, Pellegrino CM, Sabbadini DD. Comparison of primary total hip replacements performed with a direct anterior approach versus the standard lateral approach: perioperative findings. J Orthop Traumatol. 2011;12(3):123–9.
211. Parvizi J, Rasouli MR, Jaberi M, Chevrollier G, Vizzi S, Sharkey PF, et al. Does the surgical approach in one stage bilateral total hip arthroplasty affect blood loss? Int Orthop. 2013;37(12):2357–62.
212. Barrett WP, Turner SE, Leopold JP. Prospective randomized study of direct anterior vs postero-lateral approach for total hip arthroplasty. J Arthroplasty. 2013;28(9):1634–8.
213. Goebel S, Steinert AF, Schillinger J, Eulert J, Broscheit J, Rudert M, et al. Reduced postoperative pain in total hip arthroplasty after minimal-invasive anterior approach. Int Orthop. 2012;36(3):491–8.
214. Spaans AJ, van den Hout JA, Bolder SB. High complication rate in the early experience of minimally invasive total hip arthroplasty by the direct anterior approach. Acta Orthop. 2012;83(4):342–6.
215. Poehling-Monaghan KL, Kamath AF, Taunton MJ, Pagnano MW. Direct anterior versus miniposterior THA with the same advanced perioperative protocols: surprising early clinical results. Clin Orthop Relat Res. 2015;473:623–31.
216. Klausmeier V, Lugade V, Jewett BA, Collis DK, Chou LS. Is there faster recovery with an anterior or anterolateral THA? A pilot study. Clin Orthop Relat Res. 2010;468(2):533–41.
217. Restrepo C, Parvizi J, Pour AE, Hozack WJ. Prospective randomized study of two surgical approaches for total hip arthroplasty. J Arthroplasty. 2010;25(5):671–9. 5.
218. Goulding K, Beaule PE, Kim PR, Fazekas A. Incidence of lateral femoral cutaneous nerve neuropraxia after anterior approach hip arthroplasty. Clin Orthop Relat Res. 2010;468(9): 2397–404.
219. Matta JM, Shahrdar C, Ferguson T. Single-incision anterior approach for total hip arthroplasty on an orthopaedic table. Clin Orthop Relat Res. 2005;441:115–24.
220. Sariali E, Leonard P, Mamoudy P. Dislocation after total hip arthroplasty using Hueter anterior approach. J Arthroplasty. 2008;23(2):266–72.

221. De Geest T, Vansintjan P, De Loore G. Direct anterior total hip arthroplasty: complications and early outcome in a series of 300 cases. Acta Orthop Belg. 2013;79(2):166–73.
222. Unis DB, Hawkins EJ, Alapatt MF, Benitez CL. Postoperative changes in the tensor fascia lata muscle after using the modified anterolateral approach for total hip arthroplasty. J Arthroplasty. 2013;28(4):663–5.
223. Landgraeber S, Quitmann H, Guth S, Haversath M, Kowalczyk W, Kecskemethy A, et al. A prospective randomized peri- and post-operative comparison of the minimally invasive anterolateral approach versus the lateral approach. Orthop Rev (Pavia). 2013;5(3), e19.
224. Bernasek TL, Lee WS, Lee HJ, Lee JS, Kim KH, Yang JJ. Minimally invasive primary THA: anterolateral intermuscular approach versus lateral transmuscular approach. Arch Orthop Trauma Surg. 2010;130(11):1349–54.
225. Amlie E, Havelin LI, Furnes O, Baste V, Nordsletten L, Hovik O, et al. Worse patient-reported outcome after lateral approach than after anterior and posterolateral approach in primary hip arthroplasty. Acta Orthop. 2014;85:463–9.
226. Cashman JP, Cashman WF. Comparison of complications in transtrochanteric and anterolateral approaches in primary total hip arthroplasty. Orthopedics. 2008;31(11):1085.
227. Lindgren V, Garellick G, Karrholm J, Wretenberg P. The type of surgical approach influences the risk of revision in total hip arthroplasty: a study from the Swedish Hip Arthroplasty Register of 90,662 total hip replacements with 3 different cemented prostheses. Acta Orthop. 2012;83(6):559–65.
228. Weber M, Berry DJ. Abductor avulsion after primary total hip arthroplasty. Results of repair. J Arthroplasty. 1997;12(2):202–6.
229. Tan J, Chen H, Chen C, Liang X, Huang W. The strength and function of hip abductors following anterolateral minimally invasive total hip arthroplasty. Chin J Traumatol. 2014;17(2):73–8.
230. Lübbeke A, Kampfen S, Stern R, Hoffmeyer P. Results of surgical repair of abductor avulsion after primary total hip arthroplasty. J Arthroplasty. 2008;23(5):694–8.
231. Howell JR, Masri BA, Duncan CP. Minimally invasive versus standard incision anterolateral hip replacement: a comparative study. Orthop Clin North Am. 2004;35(2):153–62.
232. Higuchi F, Gotoh M, Yamaguchi N, Suzuki R, Kunou Y, Ooishi K, et al. Minimally invasive uncemented total hip arthroplasty through an anterolateral approach with a shorter skin incision. J Orthop Sci. 2003;8(6):812–7.
233. Lin DH, Jan MH, Liu TK, Lin YF, Hou SM. Effects of anterolateral minimally invasive surgery in total hip arthroplasty on hip muscle strength, walking speed, and functional score. J Arthroplasty. 2007;22(8):1187–92.
234. Walde TA, Blattgerste D, Sehmisch S, Kuttler W, Walde HJ, Koster G. Early results and patient satisfaction after total hip arthroplasty using a minimally invasive anterolateral approach. Hip Int. 2009;19(4):367–71.
235. Repantis T, Bouras T, Korovessis P. Comparison of minimally invasive approach versus conventional anterolateral approach for total hip arthroplasty: a randomized controlled trial. Eur J Orthop Surg Traumatol. 2015;25(1):111–6.
236. Queen RM, Appleton JS, Butler RJ, Newman ET, Kelley SS, Attarian DE, et al. Total hip arthroplasty surgical approach does not alter postoperative gait mechanics one year after surgery. PM R. 2014;6(3):221–6. quiz 6.
237. Queen RM, Butler RJ, Watters TS, Kelley SS, Attarian DE, Bolognesi MP. The effect of total hip arthroplasty surgical approach on postoperative gait mechanics. J Arthroplasty. 2011;26(6 Suppl):66–71.
238. Meneghini RM, Smits SA. Early discharge and recovery with three minimally invasive total hip arthroplasty approaches: a preliminary study. Clin Orthop Relat Res. 2009;467(6): 1431–7.
239. Inaba Y, Kobayashi N, Yukizawa Y, Ishida T, Iwamoto N, Saito T. Little clinical advantage of modified Watson-Jones approach over modified mini-incision direct lateral approach in primary total hip arthroplasty. J Arthroplasty. 2011;26(7):1117–22.

240. Martin R, Clayson PE, Troussel S, Fraser BP, Docquier PL. Anterolateral minimally invasive total hip arthroplasty: a prospective randomized controlled study with a follow-up of 1 year. J Arthroplasty. 2011;26(8):1362–72.
241. Kurtz S, Ong K, Lau E, Mowat F, Halpern M. Projections of primary and revision hip and knee arthroplasty in the United States from 2005 to 2030. J Bone Joint Surg Am. 2007;89(4):780–5.