Ed INVITED REVIEW

Shoulder & Elbow

Shoulder & Elbow
2019, Vol. 11(2) 140–148

DOI: 10.1177/1758573218790119
journals.sagepub.com/home/sel
SAGE

Journey of the glenoid in anatomic total shoulder replacement

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Abstract

Anatomic total shoulder arthroplasty (TSR) has been shown to generate good to excellent results for patients with osteoarthritis and a functioning rotator cuff. Many studies have reported that the glenoid component loosening and failure remain the most common long-term complication of total shoulder arthroplasty. The approach to glenoid component is critical because a surgeon should consider patient-specific anatomy, preserving bone stock and joint line restoration, for a good and durable shoulder function. Over the years, different glenoid design and materials have been tried in various configurations. These include cemented polyethylene, uncemented metal-backed and hybrid implants. Although advances in biomechanics, design and tribology have improved our understanding of the glenoid, the journey of the glenoid component in anatomic total shoulder arthroplasty has not yet reached its final destination. This article attempts to describe the evolution of the glenoid component in anatomic TSR and current practice.

Keywords

glenoid, total shoulder arthroplasty, polyethylene, metal back

Date received: 10th June 2018; accepted: 11th June 2018

Introduction

In 1974, Neer[1] introduced the first-generation total shoulder arthroplasty (TSA). This included a monoblock humeral stem and a cemented, all-polyethylene (PE) keeled glenoid prosthesis. In 1982, Neer et al.[2] published a case series showing significant improvements in shoulder range of motion and function following TSA. Since then, use of TSA has increased significantly around the world.

With time, TSA has shown to be a good long-term solution for degenerative shoulders and some types of post-traumatic proximal humerus fracture sequelae. Increasing use of glenoid component led to focus on implant survivorship beyond short-term clinical outcome. Many studies reported that the glenoid component loosening and failure was the most common long-term complication of TSA. This accounted for approximately 24% of all TSA complications.[3] Moreover, the rate of glenoid lucent lines after cemented all-PE implants in anatomic TSA has been reported to be up to 90%.

The causation of cemented PE glenoid loosening is multifactorial, including implant design, surgical technique including cement use, patient characteristics, rotator cuff function and infection. To improve survivorship, other glenoid designs and materials were developed through the years including uncemented metal-backed, hybrid and stepped designs.

Knowledge of the native glenoid anatomy and pathology, indications and technique of implantation, potential mechanisms of failure and the rationale behind various implant designs allow the surgeon to minimize risk of failure and maximize outcomes following TSA.

Glenoid preparation, fixation and radial mismatch

Proper implantation and fixation technique of glenoid implant is crucial to maximize long-term success of

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TSA. In general, the goal of glenoid component placement should be retroversion less of 10°, achieving 98% or greater glenoid component seating and maintaining high quality bone stock for support.

Glenoid reaming technique to create an optimal positioning of the implant is first in a series of relevant technical steps. Reaming tools have improved from hand-controlled to power-operated ones. Preservation of subchondral bone of the glenoid is very important to provide a rigid structural support to the implant and to prepare a perfectly congruent glenoid bone surface.[4] Furthermore, in case of excessive retroversion and/or biconcave glenoid and static posterior humeral head subluxation, excessive eccentric reaming to implant the glenoid component should be avoided. Ho et al.[5] showed that glenoid components implanted with more than 15° of retroversion are associated with fivefold increase of osteolysis around central peg. Cadaveric and computer simulation studies have indicated that approximately 15° is the amount of retroversion that can be successfully corrected by eccentric reaming without vault penetration.[6]

However, the degree of deformity correction using this method may be limited by the structure of the glenoid vault. The tapered morphology of the vault causes the diameter of the bone surface to decrease as it is reamed more medially. At some stage, bony support for the good prosthesis positioning is weakened. In severe cases of glenoid deformity, complete correction to achieve 'normal' anatomy may not be possible.

In addition, there are other potential downsides to progressive medial reaming. As reaming progresses medially, the glenoid vault narrows, thus decreasing the amount of bone stock available for implantation. Significant reaming may also result in implantation of a smaller glenoid component with substantial mismatch between the glenoid and humeral head. Severe medialization of the glenoid may also decrease tension in the rotator cuff, with possible detrimental functional consequences. Finally, excessive glenoid reaming to correct glenoid version may increase the risk of medial subsidence, as demonstrated by Walch et al.[7]

In these situations, different surgical strategies should be used to avoid failure of implant.

Another relevant issue that may influence glenoid implant survival is the radial mismatch.

In shoulder arthroplasty, mismatch is defined as the difference in the radius of diameter of curvature between the humeral head and the glenoid components. This parameter is defined as 'conformity'.

The ideal conformity and/or mismatch between humeral and glenoid implant is debatable. A fully conformed articulation, such as the original Neer prosthesis, may uniformly distribute stress at the implant–bone interface. However, translation can occur more freely with less conformity between the implants (however, contact pressures are not uniform). A previous study showed that the mismatch had a significant influence on the scores for the glenoid radiolucent lines (RLLs), which were best when the radial mismatch was between 6 and 10 mm; in particular larger is the mismatch and lesser is the conformity of the glenoid component, lower is the radiolucency.[8]

Glenoid design has been identified as an important risk factor for component loosening. A wide variety of options are currently available for the glenoid component. However, long-term outcome studies are poorly available for most of these implants, and the surgeon is left to rely on an understanding of the implant design rationale when choosing the appropriate implant.

In an attempt to improve the clinical results and reduce failures, different designs were developed. These can be classified into three basic types of glenoid fixation: all PE cemented, cementless metal-backed and more recently, hybrid designs.

Significant evolution has occurred in the last few decades in terms of design, primary fixation and materials to develop newer glenoid designs. With development of reverse TSR, considerations for possible future options of revision and conversion of anatomic TSR to reverse TSR had to also be considered.

Most PE implants require cementation. Cementing of glenoid components has evolved over time to include such techniques as burr curettage, lavage, meticulous glenoid drying and cement pressurization. These techniques have been shown to decrease rates of RLLs as well as improve implant survivorship with all-PE glenoid components.[9]

However, excessive cementation may also have detrimental effects. When the cement is curing, it reaches a maximum temperature well over the limit necessary for bone necrosis to occur.[10] In addition, removal of a large cement mantle during revision for loosening or infection may leave behind a large and complex bony glenoid defect.

Furthermore, fixation of a cemented prosthesis should be strongest at time zero, but the reported presence of RLLs at the glenoid bone–cement junction even at time zero suggests that such fixation is suboptimal.[11,12]

All-PE glenoid

Glenoid design evolution for PE implants

After the experience reported by Neer et al.[2] on cemented all-PE glenoid prostheses, the design and implantation of cemented all-PE prostheses has progressed through a number of different 'generations'. This was aided by understanding of significant issues outlined by

studies dealing with conformity, shape, fixation (peg versus keel) and cementation technique.

Conforming/non-conforming. In an attempt to improve gleno-humeral stability, some authors started to use glenoid implant with an artificial labrum. Biomechanical studies showed a decrease of the prosthetic head translation with the labrum design compared to the standard design and a small mismatch effect was noticed.[13] However, clinical studies reported that radiolucency surrounding these hooded glenoid components was higher than that associated with standard TSA implant.[14] This confirms observation of Walch et al.[8] that closer is the match between the humeral and glenoid component diameters of curvature, the higher is the prevalence of periprosthetic radiolucency on follow-up radiographs.

Non-conforming designs evolved with the idea of reducing rim loading and the rocking horse phenomenon which was the primary cause of loosening in glenoid implants. Non-conforming designs allow some translation and subluxation of the humeral head before rim loading. This may reduce rim loading and wear of the PE. On the other hand, the reduction in contact between the humeral head and glenoid will increase the contact stress on the PE.

Shape. Modern glenoid components are also available in different shapes and sizes. Some implants are pear shaped to mimic the shape of the normal glenoid, whereas others are elliptical. A pear shape implant offers the potential for less overhang superiorly and less uncovered bone inferiorly, but this shape has not been shown to be superior to elliptical designs. This may be because the arthritic glenoid is not often pear shaped and properly sized elliptical implants often fit well after reaming. The back of the components may be flat or convex; curved back designs seem to resist micromotions more effectively than flat designs. In a radiographic study comparing flat and convex poly glenoid components, some authors reported that, at two years of follow-up, glenoid designs with a curved back had better seating and significantly better radiolucency scores than did designs with a flat back.[15] In addition, most current designs have glenoid backside with different radius of curvature. This allows the implant to better adapt to the arthritic glenoid to avoid excessive reaming of the subchondral bone. More recently, inset glenoid designs have been developed. This design allows a partial resurfacing of the glenoid, leaving a peripheral rim of bone that potentially improves stability and resistance.

Keeled and pegged. *Keeled* components were at first introduced by Neer. Although the rate of survival is acceptable, in long-term follow-up studies, the rate of RLLs and radiographic loosening is still high. In a study by Walch et al.[16] on convex-back keeled glenoid components without using modern cementing techniques, the glenoid survival was 99.7% at five years and 98.3% at 10 years with endpoint defined as revision surgery for glenoid loosening, and 99.7% at five years and 51.5% at 10 years with endpoint defined as radiologic loosening. As discussed, modifications have been made, especially in bone preparation and cementing techniques, to reduce the amount of RLL and loosening.[17] Kasten et al. studied keeled component by preparing the bone by impaction and utilizing modern cementing techniques. They reached a survival rate of 100% in mean follow-up of 89 months. Radiological glenoid loosening was 9% after five years, and 33% after nine years.[18] However, the prevalence of immediate postoperative radiographic lucency is greater with keeled components than with other all-poly (pegged) glenoid component.[11,19] As immediate postoperative lucent lines are most likely related to bone necrosis due to cement curing than fluid or clot, it is not clear why they are more commonly observed in keeled components.

Pegged components were designed with the intention of resecting less subchondral bone and utilizing stronger peripheral bone for fixation than conventional keeled ones. Although there are a lot of studies in favour of pegged versus keeled components,[10,11,17,19,20] there is still controversy as to which is superior. The pegged design needs less cement for fixation (17% less in a study), lowering the risk of bone necrosis, which may be the cause of progressive RLLs.[10] In fact, the amount of heat from the exothermic reaction of cement curing is related to the volume of the cement that is used and is a particular cause for concern because the thermal insulation properties of PE do not allow dissipation of the heat.[10]

Also, greater resistance to off-center loads has been shown[20] in pegged designs, Mansat et al.[21] and Roche et al.[22] did not see any difference in peripheral loading between pegged and keeled design. In some studies, pegged designs have shown less incidence of early postoperative radiolucency compared to keeled components at short-term follow-up.[18] In a recent study, the same authors showed that at an average 7.9-year follow-up, all-poly, pegged glenoid implants were equivalent to keeled implants with respect to radiolucency, clinical outcomes and need for revision surgery.[23] Furthermore, other comparative studies investigating the use of pegged and keeled implants failed to show any difference in implant longevity or clinical outcomes, although higher rate of RLLs with keeled implants was reported.[11,24]

Long-term results will determine if the use of new cemented pegged glenoid component with multiple backside curvatures can provide better results in terms of RLL and revision rates.[25]

On the other side, McLeondon et al., in a long-term follow-up, with an average of 7.2 years, concluded that the risk of glenoid component failure (radiographic and clinical) was higher in pegged than keeled components. However, pegged design used in this study had liner pegs which are not the common design in the market. Advanced presurgical glenoid erosion and younger patient age seem to be the main risk factors for radiographic loosening.[26]

In summary, long-term fixation of the glenoid component remains an unsolved problem. Loosening of cemented PE glenoid component represents an important cause of failure in TSA.[27–29] For these reasons, in recent years, several modifications to the traditional all-poly pegged implants have been introduced with the goal of improving implant stability and longevity. Divergent pegs are used to provide additional stability against micromotion. All-poly curved components that allow for bone ingrowth onto an interference-fit central peg provide the theoretical possibility of long-term biologic fixation. These implants have demonstrated promising clinical results, particularly when radiographic density is observed between the flutes of the central peg, indicating bone ingrowth.[30] Biomechanical study by Anglin et al.[20] demonstrated that the maximum tensile edge displacement of the pegged glenoids was less than the keeled glenoids and the threaded pegs had lower distractions than the cylindrical pegs. Some authors did report early good to excellent clinical and radiographic results using this implant.[31,32] However, RLLs were seen most commonly around the inferior pegs of the prosthesis, and this may represent an incipient mode of failure.[31] More recently, Merolla et al., in a CT scan-controlled study showed that at a mean follow-up of 31 months, in TSA using these new glenoid components, there was satisfactory bone ingrowth around central peg. Osteolysis was observed in two of 30 cases.[33]

Inlay design: This implant represents another surgical option. An inlay component has less diameter than the onlay design. It is implanted in the glenoid within a bone socket. Therefore, bone support and cement fixation occur not only at the back but also circumferentially at the edge of the component. This may reduce the risk of loosening. Also, when rim loading occurs, as a primary cause of loosening, the humeral head articulates simultaneously with the glenoid component and the surrounding tissue sharing the load. Biomechanical research supports the idea that inlay components resist loosening better than onlay designs because there is a better resistance to the rocking horse phenomenon.[34,35] However, there is a concern that contact of the humeral head with native tissue may cause pain, as in hemiarthroplasty.[34] Although some studies showed good clinical outcome using this implant,[36] longer follow-up studies are awaited to better evaluate this design.

Metal-backed glenoid (MBG)

MBG implants were introduced in an attempt to improve glenoid fixation and to reduce glenoid lucent lines; uncemented fixation with porous coated or tissue ingrowth components have been developed with the aim of achieving more stable fixation to the bone and a corresponding increase in implant survival.[28,29] When considering the use of MBGs, in theory, there are a number of potential advantages over a cemented prosthesis. We know from literature that fixation of a cemented prosthesis is strongest at time zero; but the reported presence of RLLs at the glenoid bone–cement junction, even at time zero, in case of all-PE glenoid, suggests that such fixation is suboptimal.[11,12] In case of uncemented prosthesis, when the initial fixation is strong, the strength of fixation should improve with time due to progression of the bone ingrowth process and thus should, in theory, improve implant survival.

MBGs components are composed of two parts. It consists of a metal back and a PE insert. With the success of reverse prostheses, there is currently renewed interest in the development of 'universal' uncemented glenoid MB implants. These implants could be used for both anatomic (with a PE insert) and reverse (with a metallic sphere) shoulder arthroplasty. In theory, possible revision surgery should be less complicated because the glenoid baseplate does not need to be removed.

Different types of MBGs have been developed. Most of these implants were, of course, uncemented and fixed to the glenoid with screws. Despite the theoretical advantages of uncemented fixation, the reported data in the literature on MBGs have been disappointing.

Cofield et al., in 1994, reported results using a Smith and Nephew Cofield I prosthesis. They observed in 88 shoulders, with a follow-up of less than two years, 16% of complications relating to the glenoid component.[37] Tauton et al.[38] reviewed 83 of these 88 shoulders previously reported by Cofield et al., at a mean follow-up of 9.5 years and noted 39.7% of glenoid loosening and a 31.3% of glenoid revision. Torchia et al., in 1997, reported on the results of 113 Neer prosthesis. At 15 years of follow-up, they observed a survival rate of 87%. However, 44% of glenoid components showed radiographic signs of definite loosening.[39] Boileau et al, in a prospective double blind randomized study, compared MBGs versus all-poly glenoid. They concluded that at a minimum of three-year follow-up, survivorship of MBGs was inferior to a cemented all-poly and revision surgery was required in 15% of MBGs cases compared with 0% in the all-poly cemented

group, although RLLs were more frequent in all-poly group.[40]

In 2015, the same authors reported a retrospective multicentric study on 165 TSA performed using the same type of MB implant. They concluded that due to high failure rate (37%), uncemented MBGs were not a viable long-term option because of accelerated PE wear leading to early revision surgery. Furthermore, conservation of the MB tray with reinsertion of a new PE insert is rarely possible because of glenoid bone loss, implant loosening, soft tissue deficiency and prosthetic instability. It was also observed that younger patients and biconcave glenoids had a negative effect on implant survival.[41]

Martin et al. reported the outcome of an uncemented glenoid component (Kirshner II C) in 140 shoulders with a medium follow-up of 7.5 years. Clinical failures were reported in 11% of cases and in 11% of cases there were fractured screws. They reported two cases of fractured MBG component.[42] Using a different MBG implant (Zimmer Mark2), Levy and Copeland[43] reported three cases of glenoid revision in 42 patients at a mean follow-up of 7.6 years.

On the other hand, Wallace in 1999 analysed 32 cemented glenoids versus 26 uncemented glenoids using a Biomet bimodular system (Biomet, Warsaw, IN USA). They noted at a mean follow-up of five years a high rate of dissociation of PE component from metal tray.[44] However, Rosemberg in 2007 reported the results of the Nottingham uncemented glenoid components at 10 years after modification of implant. Before 1997, the MBGs consisted of titanium ingrowth surface only but after 1997, they used hydroxyapatite to coat the back side of the metal glenoid baseplate. The modification of the implant moved the survival of the implant from 66 to 93%.[45] In 2013, Montoya et al. reported a failure of 10% of cases on 42 patients. This was because of breakage of the cage screw using a Universal metal-backed implant (Arthrex, Karlsfeld, Germany).[46] Clement et al., in 2013, reported on 31 consecutive patients treated with screw-fixed porous-coated MBG (Biomodular TSR, Biomet, Warsaw, IN USA) implant with survivorship of 93% at 10 years.[47]

Fox reported in a recent review of Mayo clinic experience, the outcome of 1542 primary TSA performed from 1984 to 2004. They reviewed different implants, specifically all-poly Neer II, MBG Neer II, MBG Cofield I, all-poly Cofield I, all-poly Cofield II. Analysing the data of these five groups, they noted that revision in the all-poly glenoid was 1.5%, while in the MBG group was 10.8%. An interesting point was that the revision of PE or metal wear was seen only in the MBGs. They concluded that the cause of failure of the MBG relates to material wear, loosening and instability due to reduced thickness of the PE part of the glenoid component.[48]

Fucentese in 2010 reported clinical and radiographic results of a new soft MBG component (Zimmer sulmesh) made by multiple layers of highly porous titanium. At a minimum follow-up of two years, they reported on 22 shoulders, a 13% of failure rate with broken peg.[49]

Because of these heterogeneous results, a number of MBGs have been abandoned by the proponent surgeons. However, it is quite clear that the design of the glenoid overall in case of MBGs seems to be the key to the survival of the implant. This may explain the variation in the reported survivorship of the MBG. Changes in MBG design have been haphazard, with major differences between prostheses. These differences have included the shape, width and length of central pegs and keels; the design, angulation and strength of fixation screws or pegs; the ingrowth/ongrowth surface employed; and, perhaps most significantly, the 'capture mechanism' used to hold the PE insert on the metal baseplate. Moreover, the mechanism of failure appears to vary from prosthesis to prosthesis, not surprisingly enticing 'major redesigns'. Castagna et al.[50] reported outcomes of MBG in 35 consecutive patients with a mean follow-up of 75.7 months. They used second-generation SMR MBG (SMR System, Lima Corporate, Villanova, di San Daniele, Udine, Italy) as their prosthesis of choice. This glenoid differs from the previous design with an altered central peg, with potential to fit a reverse glenosphere component. Using this system, the authors reported no PE–glenoid dissociations, and no patients underwent implant failure-related revision surgery in the period of follow-up study. The good results were attributed to four main reasons: (1) The shape of the glenoid. There is evidence that curved-backed and less conforming implants have lower loosening rates. (2) A stiff and thick metal back (5 mm) that decreases the stress in the PE component which also reduces PE wear. (3) Good initial fixation of the glenoid with two screws and a hollow central peg that provides durable long-term stability. (4) The presence of HA on the peg and not only on MB baseplate. A disadvantage related to MBGs was related to the available thickness of the PE. Prostheses retrieved from clinically unsuccessful cases frequently revealed considerable PE wear, situated eccentrically on the articular surface (Bankart lesion effect).[48,51] The thicker the PE, the longer it takes to wear before revision is required. The MB baseplate of a MBG is usually at least 1.5 mm in thickness. This automatically commits the MBG to either a thinner PE insert or the risk of 'over stuffing' the joint.[52] In addition, several MBG designs use a continuous or interrupted lip or rim on the baseplate to help hold the PE liner in place. In cases

where the humeral head sits eccentrically on the glenoid these lips further reduce the 'available thickness' of PE. When an MBG fails, PE failure (wear or dissociation) results in metal-on-metal contact between the humeral head and glenoid baseplate. This causes severe and rapid metallosis and creates relative urgency to revision surgery. If this metal-on-metal contact damages the baseplate so as to make simple liner exchange inappropriate, the subsequent need for baseplate removal can pose major technical problems. The better the ingrowth, the more difficult and potentially more destructive the process. Another issue with modular MBGs has been the incidence of PE dissociation. Separation of the PE liner from the baseplate results in discomfort, reduced function and a disconcerting 'squeak'.[12,53] In addition, the central peg or keel of most metal-backed designs was quite large, requiring removal of a significant amount of glenoid bone stock. In patients with small glenoids, this can present problems.

Monoblock hybrid implant

In an attempt to improve initial fixation and to create better seating of the implant, newer implants have been designed. These new implants consist of a monoblock system formed by trabecular porous tantalum or titanium trabecular metal (TM), fully integrated with ultra-high molecular weight PE. The implant has a postconfiguration that provides initial fixation using a press fit between the implant and the bone. The friction coefficient between this material and cancellous bone improves initial fixation. Long-term fixation is provided by biological ingrowth into the TM material.

The first-generation TM glenoid consisted of a soft MBG, the Sulmesh (Zimmer, Winterthur, Switzerland). This component had layers of titanium mesh welded together to form four porous pegs covering the backside of a PE implant. Fucentese et al.[49] studied 22 patients using these implants and found a high failure rate of 13.6% at two-year follow-up. They found that the implant failed at the metal peg and component body interface. Despite the unacceptably high failure rate, the implants that survived showed good osteointegration and signs of loosening were virtually non-existent.

More recently, two companies have developed pegged glenoids that incorporate porous metal into their design. The first is the Zimmer 'trabecular metal' glenoid. Its monoblock design is composed of a PE glenoid face that is compression moulded to a porous tantalum keel. The keel is comprised of a five-peg cluster made of porous tantalum metal to promote stable bony ingrowth. A study in which this second-generation design was used showed good to excellent results regarding ingrowth in the porous tantalum keel. However, a high failure rate of 13.6 % was also observed. They found that the implant failed at the metal peg and component body interface.[12] Using different second-generation TM glenoids, other authors have reported better results.[54] None of these studies, however, report long-term follow-up results.

Despite the paucity of literature, use of porous metal in glenoid component appears promising with regards to osteointegration and has the potential to reduce glenoid loosening. However, due to the history of loosening and catastrophic failures seen in early use of these components, judicious use and close monitoring of these implants is currently recommended.

Augmented specific implant

One of the goals of anatomic TSA is the correction of bone deformity, particularly retroversion. Preoperative glenoid retroversion has been shown to be a negative predictive factor of clinical outcomes in both, hemiarthroplasty and TSA. As mentioned above, in cases of excessive glenoid retroversion (15°–20°), reaming of the anterior high side should be done. However, excessive eccentric reaming to implant the glenoid component (with less of 10° of retroversion, ideal implantation) should be avoided to preserve bone stock and avoid peg perforation and narrowing of the glenoid. Clavert et al.[6] were the first to quantify the limitations of eccentric anterior reaming, showing that correction of 15° by eccentric anterior reaming led to protrusion of at least one of the glenoid peg components through the vault of glenoid. Gillespie et al. then showed that correction of 15° left inadequate bone stock in more than 50% of specimens. This factor is particularly true in case of B2 biconcave glenoid or in case of glenoid with more than 20° of retroversion.[54] B2 glenoid with posterior humeral head subluxation has been associated with poorer TSA outcomes compared with other glenoid types. An average of 20% of radiographic loosening and 12% of revision of glenoid in such cases has been reported. In this series, asymmetric reaming was used to correct the joint line.[55] In a recent computational study, Chen et al. investigated the effect of glenoid version correction through eccentric reaming on glenoid bone quality. Using CT scans from patients with B2 glenoids, they virtually corrected retroversion in multiples of five, from 0 to 25°. However, 10° and higher of version correction resulted in a significant loss of quality of glenoid bone, specifically in the anterior region.[56] These results indicate that other options are needed to address bone deficiency and restore joint line in such situations. Some authors proposed use of humeral head autograft with all-poly glenoid to correct this retroversion. Clinical results were mixed and reported in several small cases series. In particular, high

radiolucency rates and complications relating to bone graft resorption were reported.[53] More recent studies have shown graft incorporation rates ranging from 83 to 100%.[42,57,58] Literature review analysis using this technique, however, reported complication rates of between 11 and 71%, with revision rate between 17 and 29%.[59]

In general, the indication for a posterior bone graft with an anatomic TSA is a type B2 glenoid with $> 20^{\circ}$ or 30° of retroversion in a young patient with a low demand level.

Augmented glenoid components provide a novel approach to correcting the glenoid that limits bone removal (and thus joint line medialization), while also eliminating the need for achieving union of a bone graft. Multiple designs exist for augmented glenoids including: stepped and posterior wedged glenoids. Finite element analysis has shown that posteriorly augmented glenoid components significantly reduce the stress at the bone–cement–implant interface and predict increased longevity compared to non-augmented glenoid components placed in retroversion.[60] Biomechanical studies have shown that stepped glenoids are significantly more stable than wedge-shaped designs.[61] Glenoids of posterior wedge design need less bone removal and thus there is a significantly greater residual glenoid bone density posteriorly.[62] While there is a paucity of long-term clinical literature, early published results have been promising. From a clinical point of view, different studies have shown that this is a reproducible technique that allows correction of alignment, preserves bone and limits medialization of the joint line. However, longer follow-up is needed to assess maintenance of correction, stability of implant over time and impaction component loosening.[63–65]

Conclusions

The journey of the glenoid component in TSA is long and has not yet reached its final destination.

Evolution in biomechanics, design and tribology has improved longevity of glenoid components. Furthermore, through the decades, new prosthetic designs have been introduced to deal with complex situations (i.e. reverse shoulder prosthesis for cuff deficiency or failure) thus increasing the spectrum of possible future revisions or conversion.

Declaration of Conflicting Interests

The author(s) declared no potential conflicts of interest with respect to the research, authorship, and/or publication of this article.

Funding

The author(s) received no financial support for the research, authorship, and/or publication of this article.

Ethical Review and Patient Consent

Not required for a Literature Review.

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