The Implant-Cement Interface in Total Hip Arthroplasty

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Introduction

Despite being in clinical use for more than 40 years, the detailed function of cemented femoral stems is still not completely understood. After insertion into the femur, the behavior of the bone/cement/stem construct can be expected to depend on a number of factors. A fundamental function performed by the bone cement is the transfer and distribution of the stress between the prosthesis and the bone. The success of cemented systems, in the long term, depends on many factors. The prosthesis itself can, depending on its material composition, shape, size, and surface finish, have a complex influence on its surroundings. Bone cement and its structural and mechanical characteristics in particular have a similar influence when combined with different stem designs. Thus, the quality and shape of the materials and interaction at the stem-cement interfaces are of great importance for long-term performance.

The development of clinical loosening has been attributed to micromovements, abrasive wear, leakage, and even the pumping of joint fluid out into the cement-bone interface. The initiation of this process may vary. There is, however, substantial evidence that debonding occurs between the cement and the stem, which may initiate loosening [1, 2]. Consequently, many stem designs are made to obtain firm fixation to the cement by the use of a collar, macrotexture, rough surface, or pre-coating with polymethylmethacrylate. A diametrically different opinion is that debonding of the stem is unavoidable and that the stem should therefore be designed to adapt to such an event. The polished double-tapered uncollared stem is an example of this [3].

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Effect of Prosthetic Design and Choice of Material on Stem-Cement Interface

Stem Material

There are three types of materials that are commonly used for cemented femoral stems. These are cobalt–chromium alloys, stainless steel, and titanium alloys. The use of a titanium alloy (usually Ti-6AI-4V or Ti-5AI-2.5Fe, Ti-6AI-7Nb) was attractive because its stiffness is closer to that of bone and bone cement than the other two alloys. The clinical performance of stems made of titanium alloy seems to be closely related to stem design and especially to the choice of surface designs. Those with rougher surface have shown high failure rates [4, 5]. Some smooth and polished stems [6, 7] seem to perform well,

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Fig. 4.1 This photograph shows an Exeter stem extracted 10 years postoperative. Note the severe crevice corrosion at the medial surface of the stem

but these are also sensitive to changes of design, especially if used with high offset or in small sizes [8]. Titanium–alumina–vanadium alloy has a stiffness of about 50 % relative to CoCr or stainless steel (elastic modulus $E = 230$ GPa for CoCr and 110 GPa for titanium). It is therefore more flexible, but it is uncertain whether this feature influences the frequency of stem–cement bonding in vivo. The higher flexibility of the stem will increase proximal cement stresses. For designs with small proximal dimensions, particularly in the M-L direction, cement stresses may become too high in the stem-cement interface resulting in debonding and cement failure. Titanium alloys have a high resistance against corrosion [9]. They are, however, comparatively soft and susceptible to abrasive wear of the oxide film as well as fretting and crevice corrosion. This type of corrosion is driven by the generation of a gap (the crevice) between the stem and the cement. It has often been reported to occur at the taper of modular implants. There are some reports of early failure of cemented titanium stems showing severe crevice corrosion at the stem-cement interface [10–12], but only a few studies show this type of corrosion for designs made of stainless steel [13], whereas reports of crevice corrosion of CoCr seem to be restricted to the head taper junction (Fig. 4.1). Even if it is not completely clear to what extent crevice corrosion between the stem and the cement has an influence on clinical failure rates, it seems to be evident that this phenomenon is of clinical relevance for some designs of cemented titanium implants. Taking this into consideration, titanium stems should not be advised when very small section shafts are required, particularly in heavy patients. If they are used they should have a polished or very smooth surface to avoid abrasive wear, fretting corrosion, and increased particle load in the joint space.

Cross-Sectional Shape

Femoral stems should be able to withstand axial, bending, and rotational forces. During activities such as rising from a chair or stair climbing, rotational forces increase [14]. The ability of a stem to withstand these forces is design dependent. The cross-sectional area (geometry, size) and shape (rounded, squared), the CCD angle, and offset should be considered, in addition to patient-related factors, as factors related to the stresses transferred to the interfaces. Several radiostereometric studies have demonstrated that well-functioning stems may also slowly displace into retroversion and varus and in most cases subside inside the cement mantle as an effect of these forces [15–18].

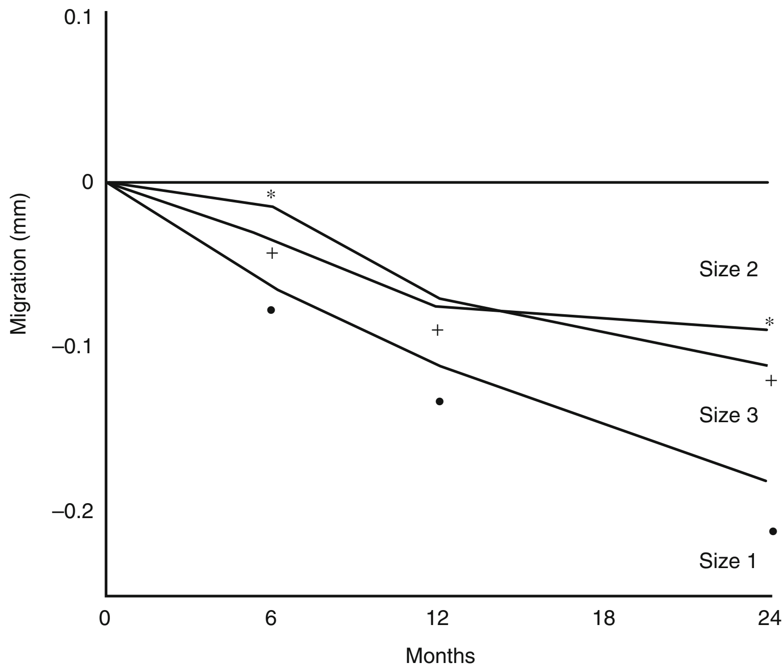

Fig. 4.2 Proximal (+)/distal (−) migration of the gravitational stem center related to stem size. Data are shown mean ± SE (Adapted from Olofsson et al. [17])

Thus, rotational resistance has become an important parameter of the design. Stems with a circular cross-sectional shape have smaller rotational stability at the stem-cement interface. The cross-sectional shape should therefore be rectangular or irregular to improve rotational stability. Stems with proximal–distal profiles along the surface also have an improved rotational stability. A potential disadvantage of a stem with a rectangular cross section is that great stresses are generated at its edges, resulting in debonding and even fracture of the cement. Most of these stems are polished and wedge shaped which reduces the risk of clinical complications due to abrasive wear. Rotational stability also depends on the cross-sectional size of the implant, which does not really explain why certain stem designs show deteriorating results with decreasing stem size.

In a retrospective study by Sylavain and coworkers, a rough-surfaced pre-coated stem was analyzed with an average of follow-up 36 months. These authors found a trend towards failure in smaller prosthesis sizes [19]. Kalairajah et al. [20] reported the clinical and radiographic outcomes of Taperloc arthroplasties. Patients who had smaller stems (7.5 mm or 10 mm) had a 27 % failure rate, whereas those patients who were implanted with stems equal to or greater than 12.5 mm had a 12 % failure rate. Thien et al. [21] studied design-related risk factors of 21,008 Exeter polished stems, 43,036 Lubinus SPII stems, and 7,140 Spectron EF primary stems in the Swedish Hip Arthroplasty Register inserted from 1999 through 2006. They found an increased failure rate due to loosening of the smallest version (extra-narrow) of the Lubinus stem and of the two smallest sizes of the Spectron EF primary stem. The crude revision rate of the Exeter stem was slightly higher than for the Lubinus stem, but with the Exeter the choice of stem size had no influence on the risk of revision due to loosening. Bourne et al. [22] reported that bone cement pressure during stem insertion increased when progressing from a small to a large stem, which could be expected to influence fixation at the stem-cement interface. The current authors have evaluated design variations on early migration measured with RSA of Spectron primary cemented stems [17]. Our results showed a higher but insignificant increase of stem migration in the cement mantle for the smallest stem size (Fig. 4.2) corresponding to findings by Thien et al [21]. This suggests that for at least for non-polished stem designs,

Fig. 4.3 This photograph shows a Spectron stem size one revised 8 years postoperative. The subsidence and rotation in the cement mantle had abrasive polishing effect on the matt surface on this particular stem leading to loosening and revision surgery

there is a certain lower limit for downsizing. If a matte or grit-blasted cemented stem becomes too small, it cannot withstand external forces sufficiently, resulting in abrasive wear, loosening, and osteolysis (Fig. 4.3). This limit probably varies depending on the activity of the patient and the time frame of the observation period. This means that special consideration should be given to active and heavy patients with a narrow femoral canal, especially if a large offset is required. For these cases, alternatives other than a small-sized non-polished cemented stem should be considered.

Stem Design Different Philosophies

The number of design variations of cemented femoral stems available today is partly a result of different design philosophies in combination with experience obtained from clinical studies and observations made in National Registers. It has become more and more obvious that it is the combination of shape and surface finish of the stem that is of importance for long-term results. The detailed interaction between these factors is, however, still not completely understood. Based largely on the surface finish of the stem, and also on the way the stem interacts with the cement mantle, two main philosophies of fixation have evolved, one based around a polished stem surface and the other based on a rough stem surface, with or without adjuvant fixation features. Huiskes et al. [23] have reintroduced the concept of shape-closed fixation versus force-closed modes of fixation for cemented femoral stems.

A shape-closed fixation design (Fig. 4.4) is one in which the stem achieves fixation at the stem-cement interface through a match in the shape of the surfaces of the stem and the cement with the cement gripping the surface of the stem. The aim is to achieve a rigid interlock between the stem and cement and thereby nullify movement at this interface. These designs have matte, grit-blasted, and beaded or porous surfaces into which the cement is intended to penetrate, thus achieving a solid bond between the stem and the cement. Attempts have been made to improve bonding between stem and cement by pre-coating the stem surface with cement applied onto the surface during manufacture of the stem [24]. This pre-coated stem is then inserted into the cement in the femoral canal, which binds with the pre-coated surface. Early clinical trials have suggested that the use of pre-coated stems is associated with favorable short- to medium-term survival in some studies [25, 26]. However, Callaghan et al. [27] have found that the use of a pre-coated stem is associated with poor survival,

Fig. 4.4 The Spectron EF and Exeter stems are examples for shape-closed and force-closed fixation system, respectively

with a loosening rate of 24 % at an average of only 8 years follow-up. These results of pre-coated femoral stems have been confirmed by several other clinical studies [28–30]. Therefore, it appears that, despite theoretical advantages, the use of pre-coating is detrimental to the long-term survival of femoral implants. The reason for this remains unclear. Debonding of the pre-coating from a comparatively rough undersurface over time, followed by abrasive wear and corrosion might be one explanation.

A force-closed system (Fig. 4.4) is one in which the fixation of the stem within the cement is achieved through the balance of forces without the need for the existence of a bond between the stem and the cement. A stem may act as a taper within the cement, in which case fixation is achieved through the balance of forces across the stem-cement interface and bonding between the stem and cement is neither necessary nor desirable. The balance of forces arises from the ability of a polished tapered stem to subside over short distances within the cement mantle. Retrieval analysis and laboratory experiments have shown that this subsidence is accommodated by cement creep. The subsidence of the polished taper within the cement means that this type of stem can maintain satisfactory fixation despite changes in the cement mantle over time. This ability to subside may allow the loading of the stem to be distributed evenly, especially in the proximal femur where remarkable preservation of calcar bone has been seen with this type of stem [31].

Milles [32] has investigated the effect of stem surface finish on cement stresses. He has shown that for polished stems the major load component is radial compression but for rough stems there is significant shear (Fig. 4.5). Studies evaluating the physical properties of acrylic cement [33, 34] have shown that cement is significantly stronger when loaded in compression compared to loading in tension or shear.

Surface Roughness

In attempts to improve the bonding between stem and cement, several authors have, in the laboratory, investigated the relationship between stem surface roughness and the shear strength achieved at the interface [35–37]. These studies have shown that increasing stem roughness leads to increased strength of the stem-cement interface, although in many cases the testing methods have been unrealistic and have disregarded the effects of cyclical loading [36]. Such experimental data have led to the belief that a rough surface finish is beneficial [24] and to the development of a number of femoral prostheses with roughened surfaces. Kärrholm et al. [15] used RSA to measure the migration of different femoral stem designs inside the cement mantle. They found that stem migration inside the mantle occurred with variable frequency for all designs studied, including two with comparatively rough surfaces. If these findings can be generalized, such stems should be used cautiously and probably only in sizes with a sufficiently large surface area to be able to counteract debonding and inducible displacements during activity as previously discussed. Verdonschot and coworkers [38] implanted

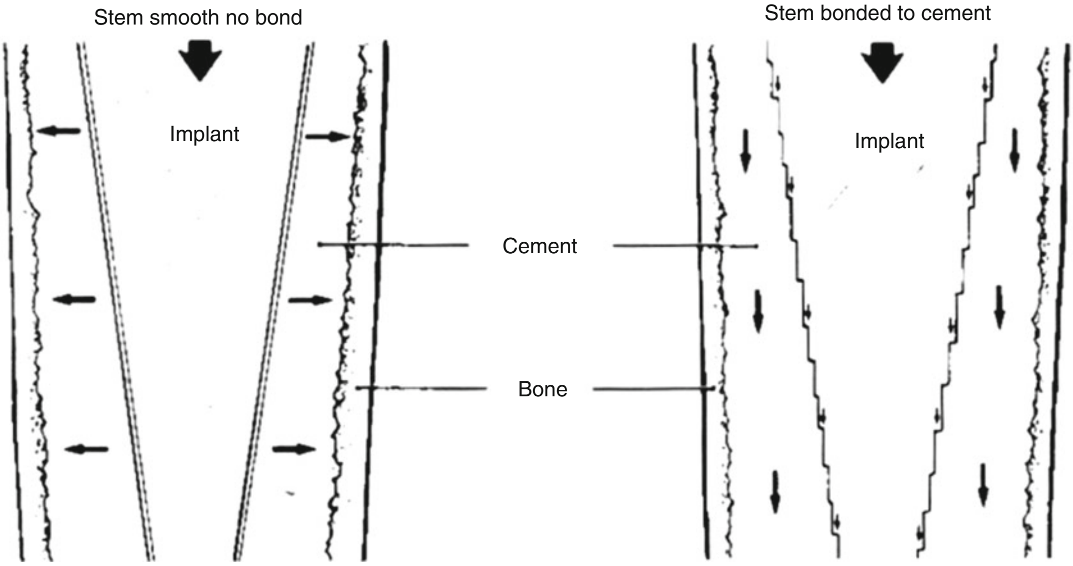

Fig. 4.5 In polished stems there is no bond to the cement, the load at the cement-bone interface is radial compression. If the stem is bonded to the cement, the loads at the cement-bone junction are shear loads

metal tapers into cement with three different surface roughness values (Ra's were 0.02, 1.1 and 11 μm) and exposed the tapers to a cyclic load. They measured migration and determined the amount of damage at the interface (abrasion) and in the cement (cracks) in sections using scanning electron microscopy (SEM). Although migration was less for the rough tapers, the amount of (abrasive) damage was larger for these components. The experimental study by Crowninshield et al. [37], who applied a cyclic displacement onto a rough piece of metal that was compressed against bone cement, is in agreement with these findings. Verdonschot and coworkers evaluated the relationship of surface roughness, cyclic micromotions, and stresses in the cement around the asperities of the roughness profile using finite element micromodels [39]. Micromotions reduced with increasing surface roughness. Despite the fact that cyclic micromotions were maximal for a surface roughness of 0 μm (theoretical case), the local cement stresses remained low due to the absence of asperities on the metal surface. At a roughness value of Ra = 15 μm, local cement stresses were very high, indicating a high abrasive mechanism. Interestingly, when the surface roughness was further increased, local cement stresses reduced again because of reduced cyclic motions caused by the better "grip" of the metal surface on the cement (Fig. 4.6). These studies by Verdonschot show the complexity of stem-cement interface mechanics and cement abrasion. The surface roughness beyond which the abrasive potential diminishes depends on many other factors such as prosthetic design, offset, loading conditions, location, cement characteristics, and other patient-related factors.

There are well-documented instances when roughened stems have been found to fail earlier than polished versions of the same implant. According to the Swedish Register, the Exeter matt stem with a surface finish of about Ra = 1.0 μm produced significantly worse results than the polished version (with a roughness of Ra = 0.02 μm). The Iowa stem is another example [40]. Race et al. [41] reported more gaps at the stem-cement interface with a grit-blasted cemented Charnley femoral stem (Ra 5.3 μm) than with a similar design with a satin surface (Ra 0.75 μm). On the other hand, Von Knoch et al. [42] evaluated the surface roughness of 11 femoral components (Ra = 1 μm) that were retrieved after 2–15 years. They found no abrasion or corrosion phenomena suggesting that the stem had been very stable. Spectron EF

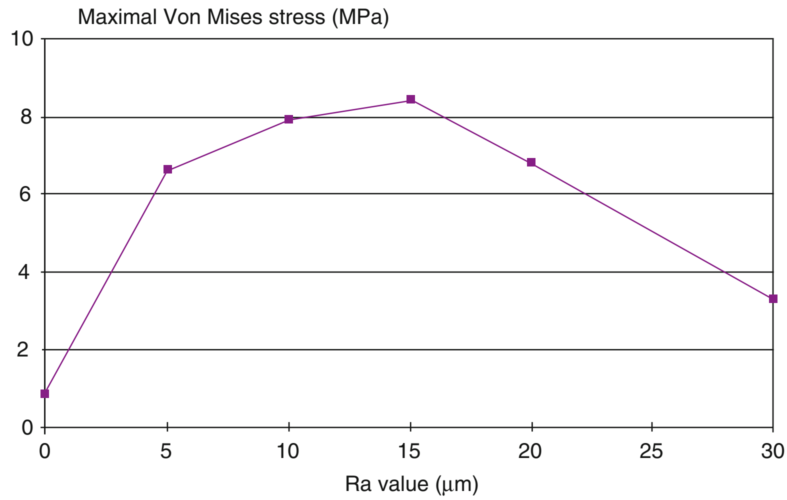

Fig. 4.6 Results from a finite element micro-analysis of the surface roughness of a straight-tapered unbonded stem. The local stresses around the asperities of the stem surface did show a maximum at 15 μm; beyond that value the local stresses were reduced (Adapted from Verdonschot et al. [39])

prosthesis which is a straight cobalt–chromium stem, proximally grit-blasted with an average surface roughness of 2.8 μm and distally smoother with an average roughness of 0.7 μm, has shown 97.5 % survival at 15 years in the Swedish hip registry [43].

There have been studies reporting good results using femoral stems with either smooth [40–49] or rough surfaces [50–54]. The optimum balance between these two factors still remains uncertain. Failure of THAs is of multifactorial origin (cementing, patient characteristics, component design); thus, valid comparisons cannot be made unless most of the factors are similar between the studies being compared.

Stem Migration and Wear

Perfect bonding between the stem and cement must be achieved and this bond must be durable to produce long-term success of rough stems. Poor adherence of the cement to the stem and inclusion of gas (air mixed with evaporation from the cement), shrinkage of the cement during curing, and creep may cause early debonding and gaps at the interfaces [55–57]. Additionally, both experimental [58] and in vivo [59, 60] studies have shown that during the lifetime of a hip replacement, the stem further debonds from the cement, opening up a gap between the stem and cement [38, 61]. Howell et al. [62] have reported an analysis of stem wear on the surface of 172 femoral stems of 23 different designs. They demonstrated that wear changes affected 93 % of stems in the study and this included 74 stems that were reported as being well fixed by the revising surgeon. The wear was often localized and was concentrated along the anterolateral and posteromedial borders of the stems. They found a fundamental difference in wear morphology on matt and polished stems. Matt stems were found to wear through abrasive polishing of the surface. Removal of debris was probably brought about by fluid in the stem-cement interface, and they found evidence of slurry wear of the matt stem surfaces caused by high-pressure fluid containing hard particles (Fig. 4.7). In contrast, the wear morphology of polished stems was typical of fretting wear and the stem surface surrounding the areas of wear was unaffected by the wear process.

Our group evaluated 97 hips that were randomized to receive Spectron primary stems fixed with either fluoride-containing cement or conventional cement [63]. Subsidence was measured with radiostereometric analysis. Two patients (three hips) underwent revision surgeries. Subsidence in the cement mantle of these hips was between 0.4 and 1.35 mm at revision surgery (Fig. 4.8).

When a rough stem subsides in the cement mantle over the acceptable level for this particular stem design, a high probability of abrasive polishing of

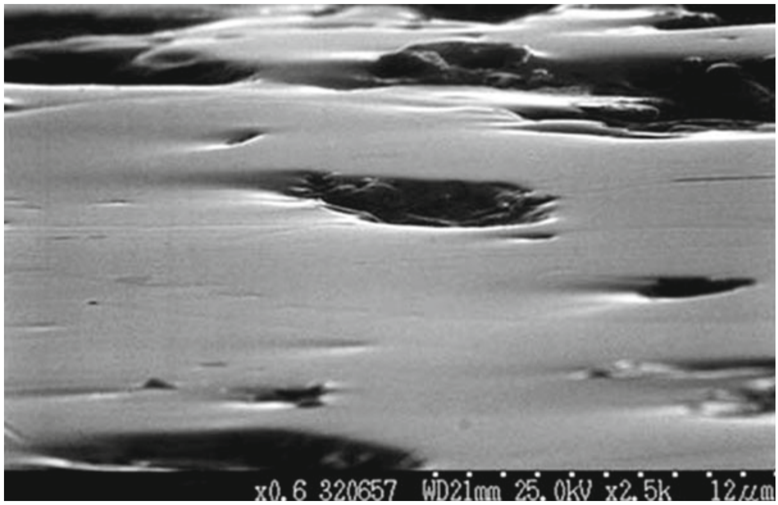

Fig. 4.7 Scanning electron micrograph of erosion or "slurry wear" seen on the surface of a matt stem. A comet tail appearance is seen on one side of each surface depression, a typical appearance of slurry wear caused by high-pressure fluid containing hard particles (Adapted from Howell et al. [62])

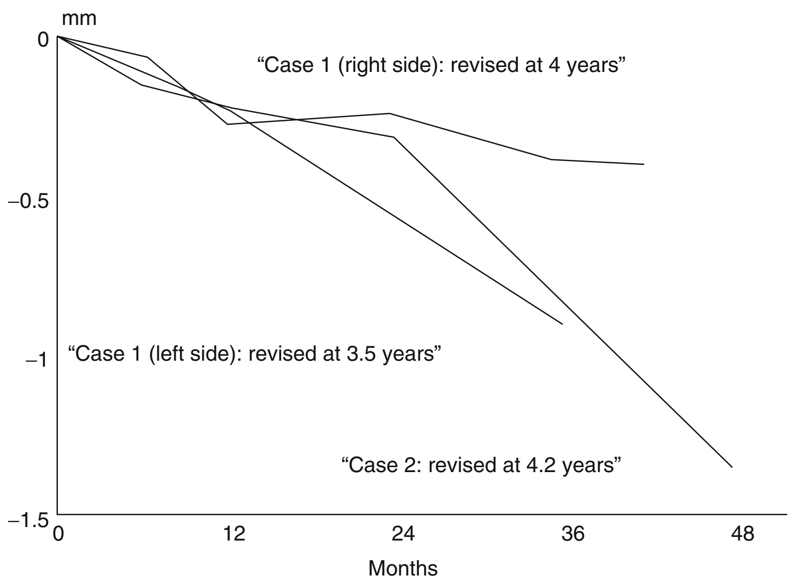

Fig. 4.8 This graph shows the distal migration in three stems (two patients) that had been revised (Adapted from Digas et al. [63])

the matt surface is indicated as well as abrasive wear of the cement mantle. This may have several important effects. It generates large numbers of particles that may contribute to third-body wear, both at the stem-cement interface and at the articulation. Another important effect of abrasive wear of the cement mantle is enlargement of the gap at the stem-cement interface, which leads to cyclical movements of the stem and further cement wear. The result is destabilization of the stem within its cement mantle and a slow but probably continuous enlargement of the space between the stem and the cement. On the other hand, the fretting wear of polished stems as shown by Howell et al. [62] occurs below the level of the original stem surface, leaving the surrounding stem unaffected. It is therefore likely that polished stem wear represents a more benign process. Furthermore, a stem with a polished surface and the correct geometry may function as a taper within the cement [31, 64, 65], allowing subsidence of the stem within the cement mantle, thus closing the stem-cement interface, preventing fluid migration and the dispersal of particulate debris. As a consequence, polished stems,

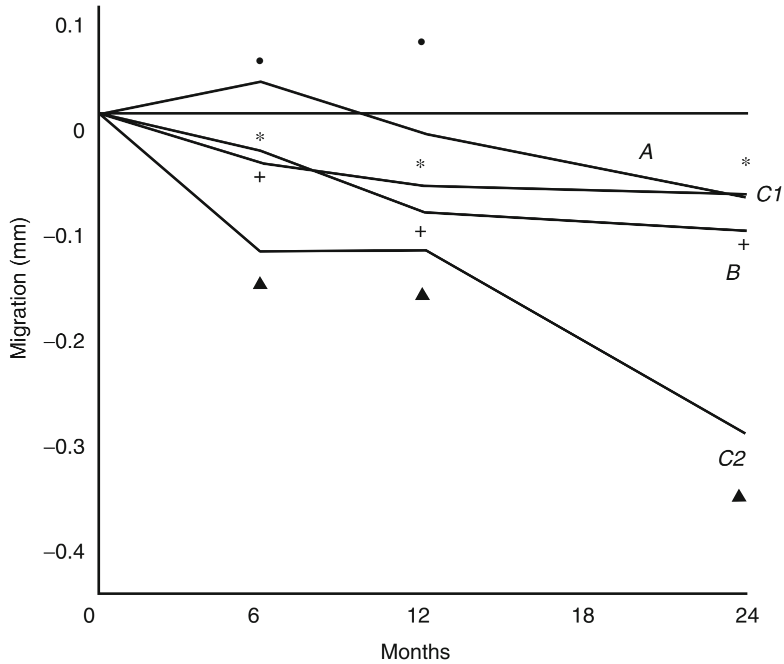

Fig. 4.9 Proximal (+)/distal (–) stem migration related to cement mantle quality is shown. Data are shown mean ± SE for Barrack groups A to C2 (Adapted from Olofsson et al. [17])

at least in theory, are less prone to the resultant third-body wear or the effects of debris and fluid flow. Stems with matt surface, such as the Lubinus SPII, have excellent survival rates in the Swedish hip registry (94.3 % at 19 years). Other stems with matt surface, such as the Müller CoCrNiMo straight stem [66] and MS-30 [67], have also shown good long-term performance (92.7 % at 15 years and 100 % at 10 years, respectively). The reason why these designs rarely seem to fail because of abrasive wear is not known but could perhaps be related to good fixation inside the mantle and reduced sensitivity to wear in cases where debonding occurs. A firmer fixation of the stem to the mantle might have other beneficial effects such as reducing the risk of periprosthetic fractures [68].

The Cement Mantle

The quality of the cement mantle is important for stem fixation [69–71]. Ramaniraka et al. [72] have evaluated micromovements at the bone-cement and stem-cement interfaces. They found that movements at the bone-cement interfaces were minimal if the cement mantle had a thickness of 3–4 mm but increased if it became wider. Abnormally high micromovements occurred when the cement was thinner than 2 mm. With use of contemporary cementing technique, initiation of loosening at the bone-cement interface is probably very rare. The production of an intact and durable cement mantle during an operation is, however, of fundamental importance.

As early as 1983 Carlsson et al. [73] observed scalloping around stems with broken cement mantles. They suggested the use of a centering device to avoid this complication. The present authors have evaluated the influence of design variation on the early migration of cemented stems with RSA [17]. We found that cases classified as C2 (presence of stem–cortex contact according to Barrack's classification) subsided more than those with a better quality of cement mantle (A-C1, Fig. 4.9). This observation suggests that patients with cortex–stem contact more easily debond from the mantle, which facilitates transport of joint fluid and debris from the joint to the interface. An inadequate cement mantle, with implant contact with the inner and distal femoral cortex, has been correlated with long-term loosening and femoral osteolysis [74].

Creep

Fatigue failure and creep are two critical factors in the endurance of bone cement. The bone cement creeps under dynamic and static loading conditions. As a result, stems which are debonded from the cement may gradually subside, depending on their shape and surface roughness followed by expansion of the cement mantle around the shaft. This phenomenon produces a redistribution of the stresses in the cement, which may have favorable or damaging effects on the entire prosthetic system. According to Verdonschot and Huiskes [58], the amount of stem subsidence which can be explained by creep is only around 0.05 mm. Kärrholm et al. [15] have shown that stems which subside less than 0.1 mm during the first 2 years have a low revision rate in the Swedish National Hip Arthroplasty Registry. Lee et al. [75] claim that the cement can tolerate a considerable amount of deformation if subjected to continuous pressure at body temperature over weeks or months. Two phenomena have been described which are correlated with creep in cement. The first is the debonding of the stem from the cement, which can induce locally increased stress resulting in fractures inside the mantle. The other is the plastic flow of the cement. It has been shown that cement creep relaxes cement stresses and creates a more favorable stress distribution at the interface [76]. Delayed injection time of acrylic bone cement increases creep compared with bone cement prepared according to standard injection procedures [77]. Creep therefore depends not only on the material properties but also on the handling of the cement by the surgeon. Waanders et al. [78] investigate how fatigue damage and cement creep separately affect the mechanical response of cement at various load levels in terms of plastic displacement and crack formation in FEA studies. They conclude that when cement is subjected to low stresses, plastic interface displacement is mostly caused by cement creep, while at higher loads cement fatigue cracking is the dominant factor. They conclude that cement creep can decrease crack formation in cement by up to 20 %. Cement creep does not decrease the stress levels in the bone with respect to its initial state, and cement fatigue damage only results in an increase in bone stresses. Vacuum mixing reduces the porosity of the cement and as a consequence volumetric creeping may increase from 3–5 % to 5–7 % in different cements [79]. Creeping at the cement-bone interface can be regarded as beneficial as some interface gaps allow for revascularization [80] and no studies have shown any detrimental effect on the stem-cement interface when cement with reduced porosity due to vacuum mixing is used [57]. The exact consequences of creeping are, however, still unknown, especially concerning its relation to aseptic failure and its effects when used with polished versus matte or rough surface finishes.

The Influence of Porosity at the Stem-Cement Interface

Extensive porosity at the stem-cement interface has been found in retrieved cement mantles and in laboratory-prepared specimens [81]. This interface porosity is caused by entrapment of air at the stem surface during stem insertion and by residual porosity in the cement. A further cause is the cement's shrinkage away from the colder stem surface which produces pores [81]. Although cement curing is chemically initiated, polymerization is thermally activated. Thus, cement curing starts at the warmer bone surface and progresses towards the cooler stem. Resultant pores as well as residual pores in the cement are driven towards the last polymerizing region of the stem. To counteract this effect, Jafri et al. [82] evaluated the effect of preheating the stem. They observed a dramatic reduction of porosity at the stem-cement interface. This effect was observed at a temperature difference between the bone and the stem of 3° and was most pronounced at a difference of 7°. They recommended preheating of the stem to 40° in clinical practice. Iesaka et al. [83] have shown that stems preheated to 37° had greater interface shear strength at stem-cement interface than stems at room temperature both initially (53 % greater strength) and after simulated aging (155 % greater strength). Fatigue lifetimes were also improved and there was a >99 % decrease in interface porosity. When

Table 4.1 The different stem design and patient materials

Group	Name	Material	Surface finish (μm) Prox.	Surface finish (μm) Dist.	Male/fem	Mean age
1	Lubinus SP II	CoCr alloy	1.5	1.5	8/12	67(52–78)
2	Lubinus SP II	TiALV alloy	1.0	1.0	9/14	65(51–76)
3a	Spectron EF	CoCr alloy	2.8	0.7	6/10	70(65–76)
3b	Spectron EF	CoCr alloy	2.8	0.7	10/11	58(42–70)
3c	Spectron EF	CoCr alloy	2.8	0.7	4/13	71(61–81)
4	Anatomic-Option	CoCr alloy	1.5	1.5	15/29	58(32–69)
5	Tifit	TiALV alloy	1.3	1.3	12/8	52(38–66)
6	SHP	CoCr alloy	3.8	2.0	8/12	67(55–78)
7	Exeter	Stainless steel	<0.5	<0.5	11/5	71(63–81)

cement is mixed under vacuum, cement porosity is significantly reduced, producing less porosity at the stem-cement interface [57, 81]. Various studies have shown that interface porosity affects the debonding energy of the interface [84], weakens the resistance of the cement to torsional load [85], and decreases fatigue life of the stem-cement interface [83]. Interface porosity has also been linked to the initiation of cement cracks [59, 86]. The evidence that reduction of interface porosity improves the strength of the interface, thereby increasing the longevity of cemented implants, is convincing.

Migration Pattern of Cemented Femoral Stems

Several studies [87–89] have shown that early migration precedes clinical loosening. Micromovements open up interfaces, increase abrasive wear, and may be an indirect indication of asymmetrical loading of the cement mantle, subsequently resulting in fracture.

Today, there are a number of studies which have measured the migration of different designs of cemented femoral stems using RSA [2, 89–95]. In these studies migration has always been measured in relation to the bone and sometimes also in relation to cement. These materials represent stems of different shapes, materials, and surface finishes. The main purpose of many of these studies has been to evaluate total migration (stem vs bone) and to what extent this migration occurs at the stem-cement interface. Kärrholm and coworkers have evaluated the micromotion of the most common stems used in Sweden [15]. The materials are presented in Table 4.1. Lubinus SP II is anatomic, double curved with anterior and posterior ridges and a wide collar. Lubinus SP II stems of CoCr alloy constitute the references in this study because of their thorough documentation in the Swedish National Registry. This stem design showed a small early subsidence with only minimal increase after 6 months (Fig. 4.10). Spectron EF is straight and has a medial collar. The first version had a stem length of 135 mm (3a). In the following version (3b-c), the stem length increases with increasing size. These stems showed no or almost no subsidence until after 6 month follow-up. There was a levelling of the curve after 2 years, suggesting a period of deformation of the cement or debonding in only a few of the cases followed by secondary stabilization (Fig. 4.10). The Tifit stem is straight and has a small medial collar. It has anterior and posterior longitudinal indentations. Its length increases with increasing size. Anatomic-Option is anatomic, double curved with proximal indentations. The stem length increases with increasing size. Both designs tended to show increasing subsidence after 6 month follow-up. The Tifit stems, followed for 5 years, migrated more slowly after the 2-year follow-up (Fig. 4.11). Scientific hip (SHP) has no collar and a teardrop-like appearance proximally but becomes more cylindrical and tapered distally. The stem has four proximal PMMA spacers. The tip is sharp. All have a CCD angle of 120°. The length of the SHP stem increases with increasing size. The Exeter stem is a polished straight, double-tapered, flat, collarless stem with a centralizer fixed

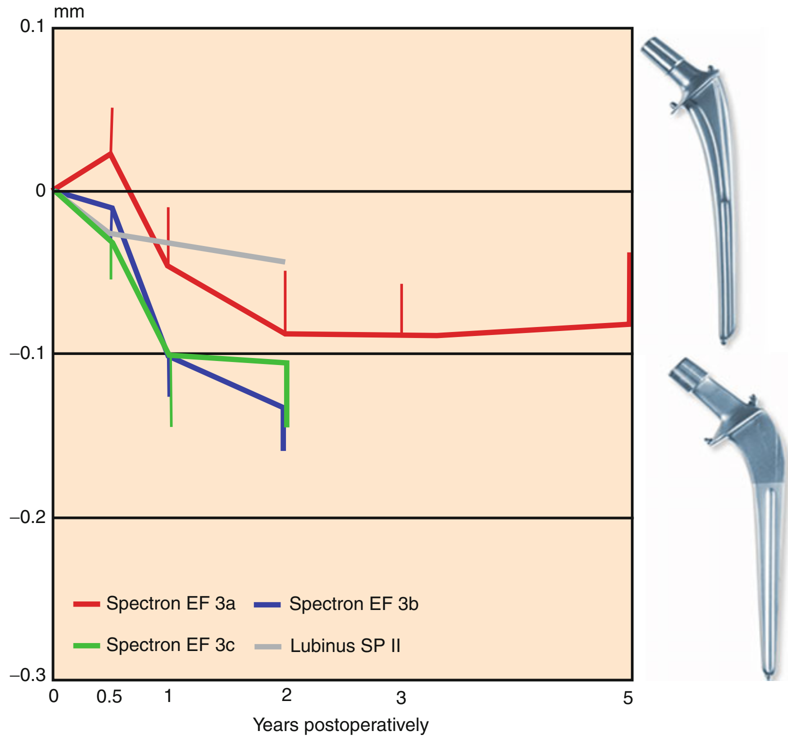

Fig. 4.10 Subsidence of the Lubinus SP II and Spectron EF stems

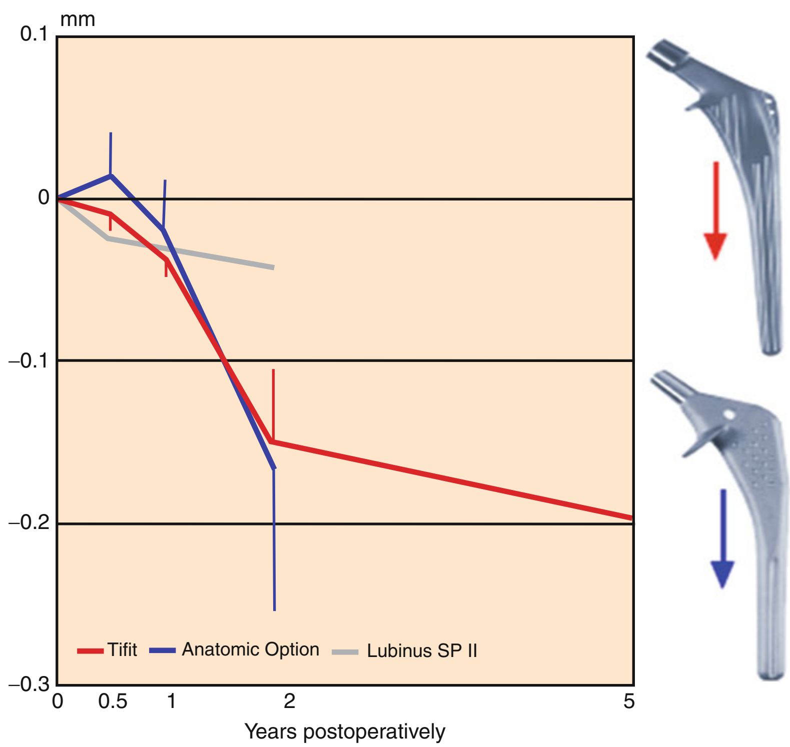

Fig. 4.11 Subsidence of the Tifit and Anatomic-Option stems

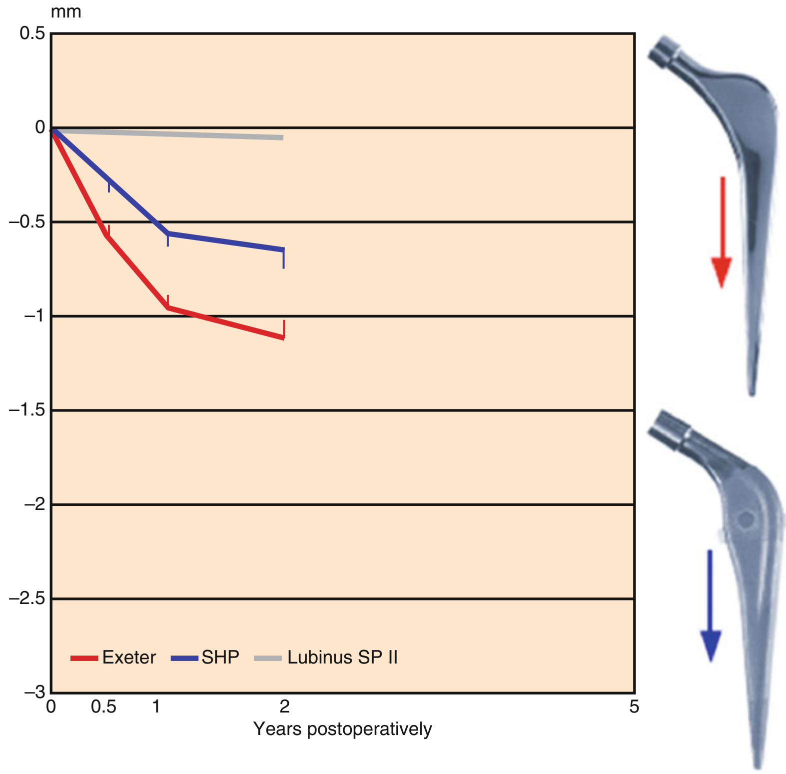

Fig. 4.12 Subsidence of the SHP and Exeter stems

to the tip. The collarless stems and especially the Exeter designs showed early and fast subsidence. The migration rate tended to decrease somewhat after 1 year (Fig. 4.12).

At 2 years control most stem designs showed retroversion. Median values exceeding 1° were noted for the Exeter (1.7) and SHP design (2.5). There was also a slight tendency to posterior tilt. The median varus/valgus tilt was close to zero. In all series stem subsidence was more common in the cement mantle (Table 4.2). Four stems that had been revised before the 5-year control showed more subsidence and retroversion than the remaining cases in each group up to 2-year follow-up. According to this study there is a close connection between stem geometry and recorded micromotion. Although subsidence of the stem and posterior displacement of the head are believed to be the most important predictors of early failure in cemented total hip arthroplasty [89], it has become generally accepted that early clinical migration values must be related to stem shape and surface finish. Thien et al. [96] used RSA in a prospective randomized study to evaluate fixation of 3 modifications of the Lubinus SP2 stem. These stems were 27 matte (standard design), 28 polymethylmethacrylate (PMMA) coated, and 29 collarless and polished. They have identical stem design, shape, and alloy. The only difference was the surface finish and the presence of a collar or not. The mean subsidence for the polished stem was 0.4 mm at 2 years, while for the other two groups the respective values were below 0.1 mm. Between 2 and 5 years, subsidence for the three groups was nearly equal (Fig. 4.13). This study shows the effect of surface finish on stem migration. Other RSA studies have shown similar behavior for the standard matte Lubinus design with low mean values regarding both subsidence and rotation [91, 97]. Twenty-two Exeter stems have been evaluated with RSA up to 5 years by Stefansdottir at al [94]. The median migration at 2 years was 1.34 mm and at 5 years 1.77 mm. A major part of the migration occurred within the first 4 months after surgery. There were no reoperations during the 5-year follow-up in the two studies mentioned above.

Table 4.2 Stem subsidence inside the cement mantle in mm

Group	Implant	Total	Significant subsidence ($p<0.01$)[a]	Cement mantle: Subsiding >significant value	Cement mantle: Value ranges	Stem inside mantle: Subsiding >significant value	Stem inside mantle: Value ranges
1	Lubinus SP II	4	0.11	0		4	–0.20 to –0.12
2	Lubinus SP II	13	0.18	2	–0.32 to –0.20	10	–0.46 to –0.18
3a	Spectron EF	11	0.18	0		4	–0.25 to –0.19
3b	Spectron EF	16	0.20	3	–0.30 to –0.23	4	–0.46 to –0.21
3c	Spectron EF	12	0.11	4	–0.25 to –0.17	5	–0.28 to –0.13
6	SHP	14	0.11	2	–0.42 to –0.32	14	–1.10 to –0.17
7	Exeter	16	0.20	0		16	–1.94 to –0.71

[a]Significant level ($p<0.01$) for individual cases in each study varies depending on technique related factors

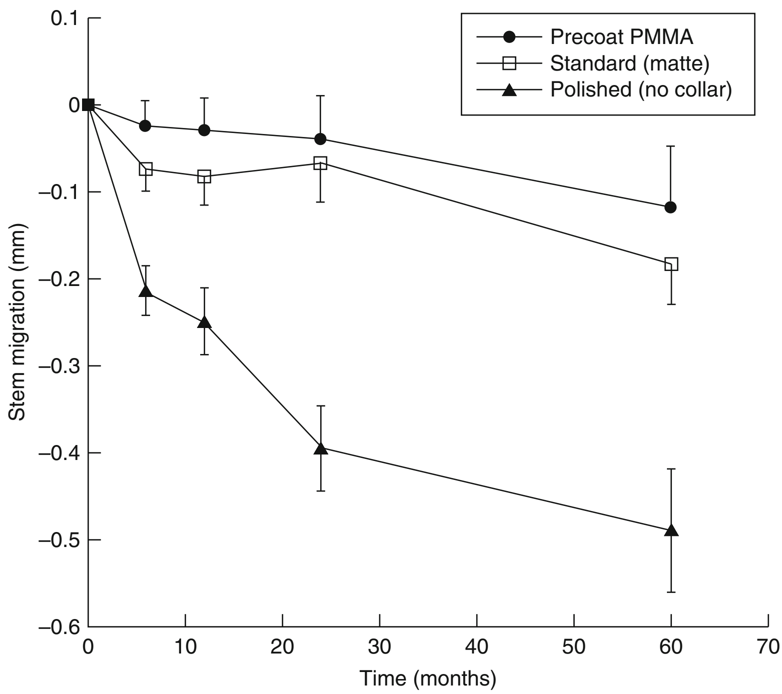

Fig. 4.13 This graph shows the proximal (+)/distal (–) migration of all 3 stem types vs the femoral bone in all cases. Mean and SEM (Adapted from Thien et al. [96])

According to the Swedish arthroplasty hip registry, the survival rate for Lubinus SPII stem is 94.3 % at 19 years, while for the Exeter stem 96.9 % at 12 years. From these RSA studies and the Swedish arthroplasty hip registry, it is evident that the amount of tolerable migration of the stem until clinical failure varies depending on the design and surface finish of the stem. Even if the magnitude of migration is design related, the pattern of motion in cases with impending clinical failures seems to be similar. The stem subsidence and rotation into retroversion are higher in failures than clinical successful uses of the same type. These implants with later clinical failure can be identified within first years of follow-up. One example is the Spectron stem which was introduced in the early 1980s. In 1995 a new version was introduced (Spectron EF Primary). The stem became narrower and shorter in the smallest sizes. In addition, a version with an increased offset, a polished neck, and a narrower cone was introduced. In the previously mentioned study [15], the older version of Spectron stem had a mean subsidence lower than 0.1 mm at 5-year follow-up (Fig. 4.10). A previous study of ours evaluating the new version of Spectron stem with RSA showed subsidence of 0.28 mm in the cement mantle at 5 years when Palacos cement had been used [63]. The higher early subsidence rate of the new Spectron design is mirrored in the annual report of the Swedish Hip Arthroplasty Register 2010 [43] which shows lower survival rate for this particular stem compared with the older design (new design 95 % at 12 year vs older design 97.5 % at 15 year, Fig. 4.14). The higher subsidence rate of this stem, which has a rough surface finish, may increase particle production due to abrasive wear. In some cases, and after a variable time period, osteolysis may develop and the mantle may fracture leading to clinical loosening and revision surgery.

RSA Studies Evaluating the Stem-Cement Interface

Accurate measurements of the cement mantle migration are more difficult than the corresponding measurements of the femoral stem because of problems of visualizing cement markers and obtaining sufficient marker scatter. This often means that reliable data can only be obtained for migration in the proximal/distal direction. Few RSA studies have evaluated micromotion in the stem-cement interface as part of total stem migration related to bone up to 5 years. In a prospective randomized study, Nivbrant et al. [6] evaluated two types of bone cement (bone cement with reduced amount of monomer Cemex Rx and Palacos R) that were used to fixate 47 Lubinus SP2 prostheses with 5 years of follow-up. All stems in this study were made of titanium alloy, and their surface was slightly smoother than the cobalt–chrome alloy version. In 28 cases subsidence of the cement mantle could be studied. In 14 of 16 cases where stem subsidence relative to bone exceeded 0.18 mm (the 99 % confidence limit of precision for this study), more than 50 % of this motion occurred inside the mantle. Stefandottir et al. [94] followed the migration of the Exeter stems in 22 primary hip arthroplasties for 5 years. The median migration at 5 years was 1.77 mm. The cement mantle could be evaluated in 12 cases. Five cement mantles migrated above the detection level (0.2 mm) between 0.20 and 0.64 mm. A correlation between distal migration and retroversion was found. They concluded that distal migration and rotations occur mainly inside the cement mantle. In a prospective study 97 hips were randomized to receive a Spectron EF stem fixed with fluoride-containing acrylic bone cement (Cemex F) or conventional bone cement (Palacos) [63]. Evaluation at 5 years revealed no differences in stem migration. In 61 cases (27 Cemex F, 34 Palacos) where proximal/distal migration between stem and cement could be studied, subsidence increased similarly in both groups. Subsidence between stem and bone exceeding 0.15 mm (the 99 % confidence limit of precision in this study) was observed in 35 cases (17 Cemex F, 18 Palacos). In 23 hips at least 50 % of this subsidence occurred inside the cement mantle. Four of the 28 cases (2C, 2P) showed distal migration of the cement mantle exceeding the detection limit for individual cases (0.16 mm; range, 0.17–0.37 mm). Eighty-four hips randomly received Lubinus SP2 stem with matte (M), polymethylmethacrylate coated (PG), or polished surface (uncollared) (P) [96]. The polished stems subsided more than the matte and PMMA-coated stems at 6 months and after 5 years (Fig. 4.14). Stem subsidence in relation to the cement could be evaluated in 37 cases (12 P, 12 M, 13 PC) at 5years. In 11 of the 12 polished stems, more than 50 % of this subsidence occurred inside the cement mantle (Fig. 4.15). In 10 of 12 matte stems and 8 of 13 PMMA-coated stems, less than 50 % of subsidence occurred in stem-cement interface.

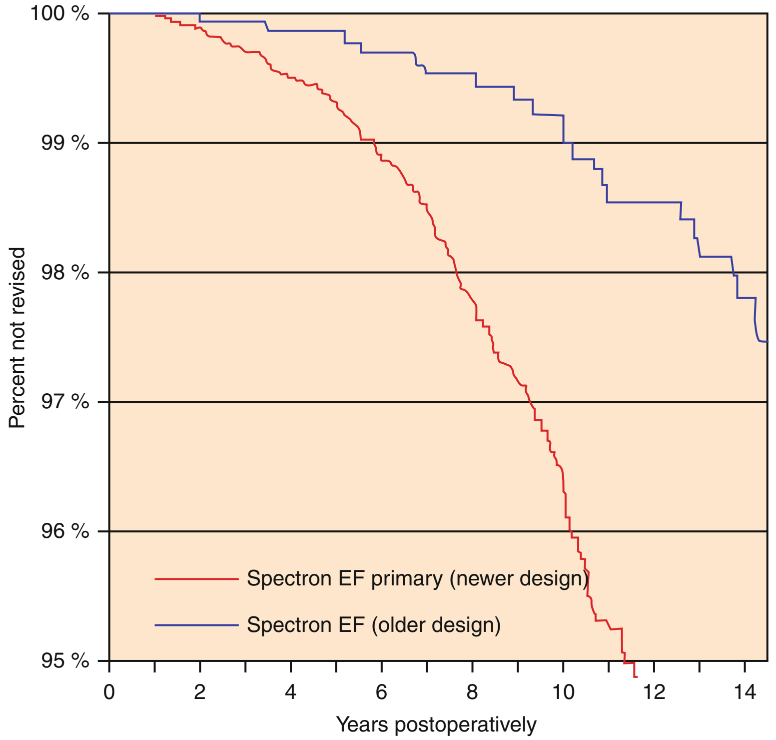

Fig. 4.14 Implant survival regarding stem revision for loosening/osteolysis with or without simultaneous cup revision

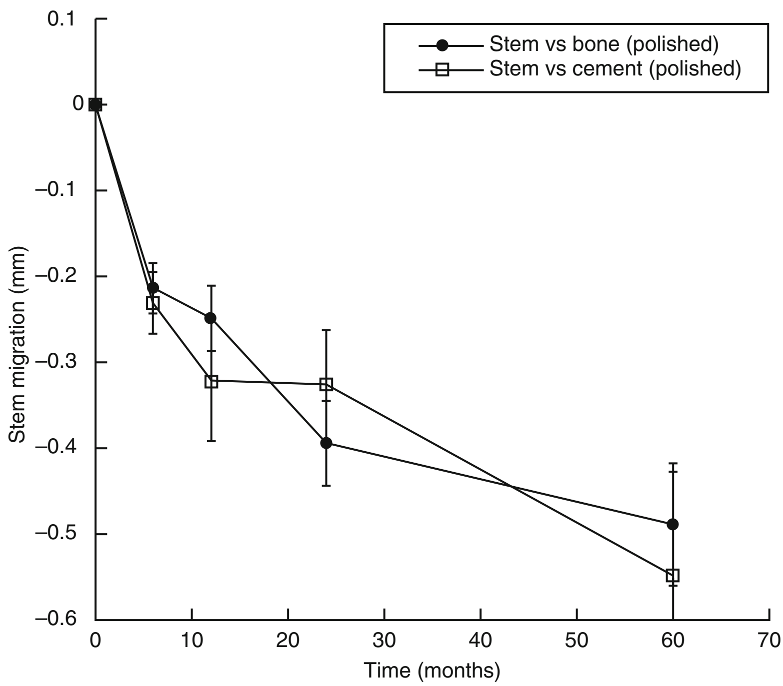

Fig. 4.15 This graph shows the proximal (+)/distal (−) migration over time of the polished stems vs the bone ($n = 25$ stems) and vs the cement ($n = 12$). Mean values and SEM (Adapted from Thien et al. [96])

In summary, RSA data have revealed that stem subsidence inside the mantle occurs with variable frequency and magnitude in almost all designs of THA regardless of the presence of a rough surface finish. Subsidence for polished stems is higher than for matt or rough stems and occurs mainly in the cement mantle, while stems with a matte or rough surface finish subside at both interfaces. The stem-cement interface might be stronger with a rough surface and may postpone debonding, but when they debond, rough stems may produce more cement debris than polished ones [38]. Some matt stems, for example, the Lubinus SP, seem to be comparatively resistant to abrasive wear. There seems, however, to be a lower size limit for this design, since the introduction of the smallest version (extra-narrow) resulted in an increased failure rate, mainly due to loosening. This particular size has, however, been used in small numbers and has, according to our knowledge, not been studied with RSA.

The increased subsidence of the polished stems in the cement mantle may be advantageous in the reduction of stresses at the bone-cement interface by facilitation of more even distribution of load at the interfaces. In the study by Thien et al. [96] mentioned above, the polished version showed significantly less loss of proximal bone mineral density, suggesting a more physiological loading of the bone. This effect, however, seemed to be temporary and mainly lasted for 2 years, whereas continued stem subsidence could be measured during the entire period of 5 years of observation. Subsidence below the acceptable limit for each stem may be advantageous if it increases stem–cement contact and stability, provided that this is not necessarily associated with inducible displacements during activity. Another effect of subsidence may be the maintenance of proximal load distribution [96]. As previously mentioned, polished stems have the disadvantage of being associated with increased risk of periprosthetic fracture. This has been well documented for the Exeter design but concerns may exist for many other designs of polished stems [98, 99]. The reason for this is unclear. It could be important to consider this in patient groups with increased risk of these complications, such as cases with previous femoral neck fractures, idiopathic femoral head necrosis, or an osteoporotic femur due to other causes.

The choice of optimum stem design may thus vary depending on patient characteristics. For the majority of patients, the choice between a well-documented matt or polished stem is not controversial. The experience of the surgeon with a particular design is probably the most important factor for the result. In active patients with a narrow femur, a polished design is probably preferable. In older patients, and especially those with the diagnoses mentioned above, a well-documented matt stem is preferable. Many stem designs used with cement have very good clinical records in the long term [43, 100, 101]. For older patients room for improvement is limited. It should, however, be emphasized that some of these stems have undergone modifications one or more times during the last 10–15 years. Furthermore, the indications for THA have tended to embrace more patients of younger age, and the demands on the implants might have increased with increasing general health and activity levels of the patient population in the older age groups. Further improvements could include minimization of particle production at the stem-cement interface to minimize third-body wear and the development of stems and fixation principles which are associated with less proximal bone loss. A still more reproducible fixation of the mantle might also be beneficial in young patients.

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