3 Bone-Cement Interface in Total Joint Arthroplasty

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The bone-cement interface represents a complex structure of acrylic bone cement interdigitating with and filling up trabecular marrow spaces, creating in this way an interlock between cement and bone. This interface thus provides the fixation of the whole cement mantle into the femur or the acetabulum. Obviously, the stability of the cement mantle and the longevity of the implants are directly dependent on the mechanical behavior of the bone-cement interface. Polymethylmethacrylate (PMMA) or otherwise "acrylic cement" was used in the industry for the first time in 1843. The first report on its use in humans was in dentistry in 1941 [1] and in orthopedic surgery in 1945 [2, 3]. However, the first report of its use in hip arthroplasty surgery was by Habouche in 1953 [4].

Although cement fixation of implants is common in a variety of joint arthroplasties, such as total knee, shoulder, and elbow arthroplasty, and in specific cases of fracture and tumor surgery, most of the principles of cementation (both experimental and clinical data) have been studied in total hip arthroplasty , because of its unique biomechanical characteristics and patterns of load transfer on the implants and on the bone-cement interface.

The use of PMMA for the fixation of implants in total hip arthroplasties was popularized by the pioneer work of Sir John Charnley in the early 1960s and has lasted up to now [5] (Fig. 3.1). Charnley believed that acrylic cement was necessary not only for the stabilization of the implants but also for the smoother transmission of the loads to the bone. Although the use of cementless implants has grown significantly over the years, a number of meta-analyses and reports of national joint registries [6, 7] have suggested that the long-term cemented fixation of hip replacement components is durable and successful.

Charnley rightly believed that the fixation of the implant to the bone by means of acrylic cement is obtained not through adhesion (glue) of the cement onto the bone but through interdigitation of the cement into trabecular bone. If the amount of acrylic cement penetration into bone is increased, the mechanical bond will improve, leading to a higher interface shear strength and fracture toughness [8]. Thus, successful long-term fixation requires stability on both interfaces, the implant-cement interface and the bone-cement interface. The stability of the one can directly affect the stability of the other [9]. The long-term survival of the bone-cement interface has been the subject of many studies. Experiments with specimens from the bone-cement interface have suggested that the interface degrades over time by fatigue loading [10].

Many parameters can influence the biomechanical properties of bone cement and affect the stability of the bone-cement interface: (1) cementing technique, (2) thickness of the cement mantle, (3) surface texture of the femoral component, (4) shape of the femoral component, and 5. manufacturing-metallurgy (Fig. 3.2).

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T. Karachalios (ed.), *Bone-Implant Interface in Orthopedic Surgery*,
DOI 10.1007/978-1-4471-5409-9_3, © Springer-Verlag London 2014

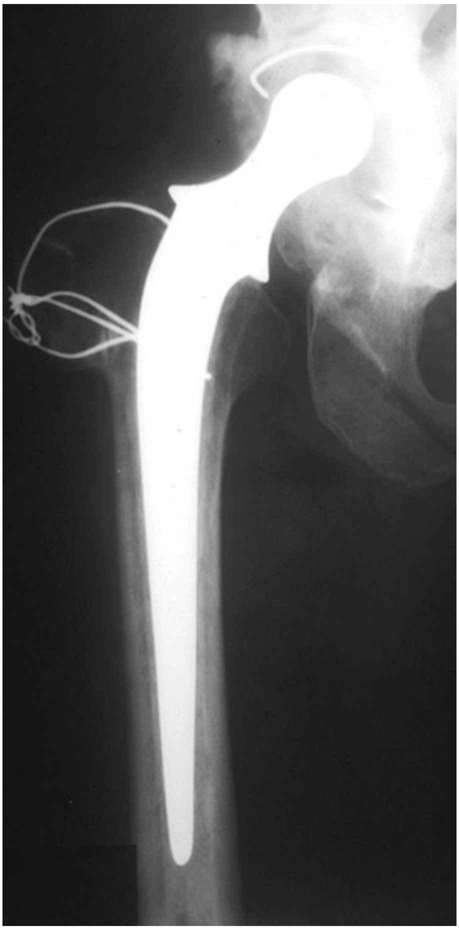

Fig. 3.1 Satisfactory radiological results at 32 years follow-up of a very early example of Charnley THA

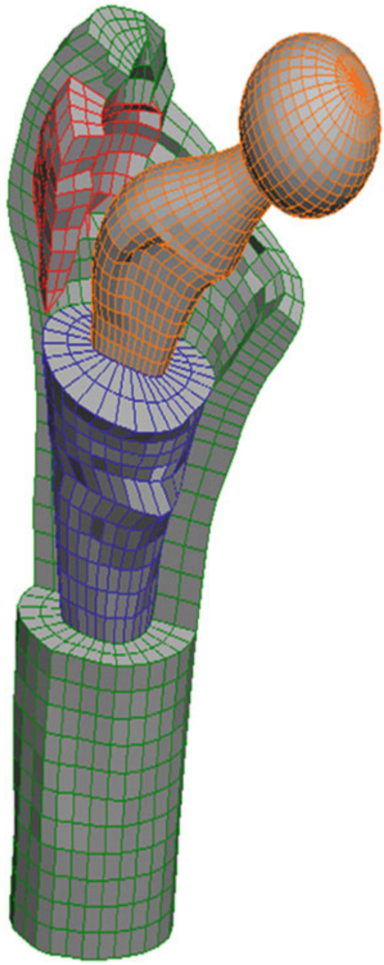

Fig. 3.2 FEA model which incorporates all variables influencing the initial mechanical behavior of the implant-bone-cement interfaces

Cementing Technique

Cementing technique is pivotal for the survival of a stable bone-cement interface. Over the years, a number of cementing techniques have been tried before universal agreement had been reached on the current "third-generation" technique (Fig. 3.3). In the early stages of its use, Sir John Charnley believed that the cement should be introduced by finger pressure. Since then the technique has evolved through three generations and now it is universally accepted that the femoral canal should be plugged to avoid distal migration of the cement and increase the interdigitation pressure and consequently the interface surface area and strength. The bone should be meticulously prepared; washed, preferably with pulsating lavage; and kept dry with specially designed instruments. The cement should be inserted in the "nonsticky" phase, by means of a special cement gun, which in certain preparation sets can be used for mixing the cement also, without the need for a mixing bowl. The insertion of the cement into the bone should start certain minutes after the commencement of mixing, according to the brand of the cement. Following the introduction of the cement, proximal seals (Fig. 3.4) are used to keep the cement under pressure and the implant is inserted when the cement is in a much less viscous state. Horne et al. studied the histology

Fig. 3.3 Satisfactory radiological results at 20 years using third generation of cementing technique

of the bone-cement interface in a canine total hip arthroplasty model after using two different cementing techniques [11]. He noticed a marked increase in the radiographic appearance of the amount of cement influx into the cancellous bone when canal plugging, lavage, and pressurization of the cement were used. Similarly the histological examination of their specimen showed that the cement had reached far into the endosteal cortex and that the cancellous bone had remained viable when the above mentioned technique was used. The Swedish Hip Arthroplasty Register has shown a survivorship of 95 % at 10 years using this modern cementing technique [12, 13].

Mixing the cement has been the subject of considerable controversy. It seems that the mixing technique plays a role regarding the formation of voids inside the cement. These voids can adversely affect the mechanical behavior of the cement. Macaulay W et al. studied three mixing techniques, vacuum mixing, centrifugation, and hand mixing, and concluded that the best result with the least number of voids was in the method of vacuum mixing [14]. Mau et al. reached similar conclusions in their study of various vacuum mixing systems with different brands of cement regarding porosity, reliability, and bending strength [15]. Contrary to these findings, the Swedish Arthroplasty Register 2000 report noticed that at 5-year follow-up there was a higher risk of revision after vacuum mixing as

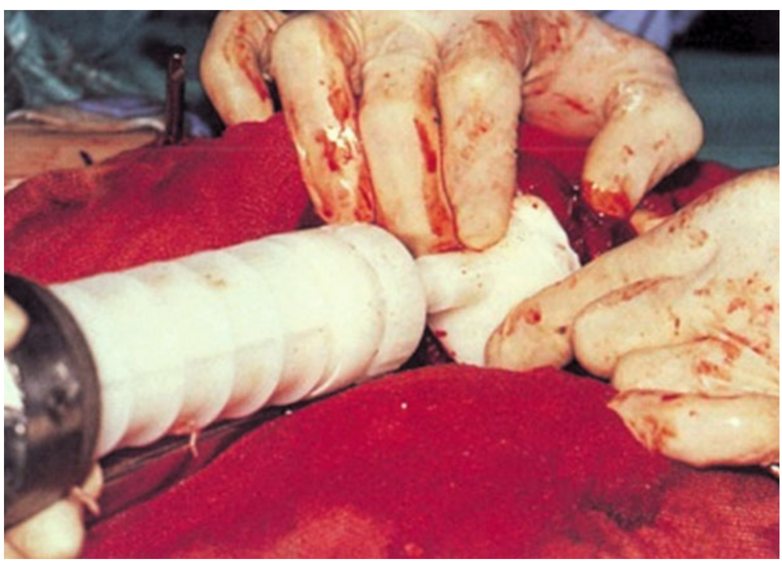

Fig. 3.4 Maintenance of cement pressurization using a proximal seal

opposed to manual mixing of the cement [12]. However, after 5 years the risk of revision after vacuum mixing became considerably less. It seems that a consensus exists that mixing the cement in vacuum produces a far more homogeneous dough with extremely few voids.

Jafri et al. described an experimental model to investigate the effect of preheating the femoral component on the porosity of the cement at the bone-cement and cement-implant interface [16]. They concluded that heating the femoral stem to 40 °C before insertion reduces the porosity of the cement significantly. Similarly, Baleani et al. showed experimentally that both vacuum mixing and preheating the stem increased the static mechanical strength of bone cement and additionally improved its fatigue life [17]. This is based on the theory that the curing of the cement is initiated at the bone-cement interface because this area is warmer. As a result of curing, shrinkage away from the cement-stem interface will follow causing this interface to weaken. Preheating the stem could reverse the direction of polymerization and hence protect the integrity of both interfaces. Curing of the cement takes place through an exothermic reaction during which polymerization is completed in temperatures ranging from 66 to 82 °C [18]. These temperatures can be detrimental for the integrity of collagen in the surrounding tissues, because collagen cannot withstand temperatures in excess of 56 °C. In the clinical situation, however, the curing temperature rarely rises over 48–56 °C, because the local blood circulation, the metallic surface of the implants, and the large surface area of the bone-cement interface dissipate the heat and enhance the cooling of the area, so that intraoperatively the temperature of the cement is 32.3 °C. The addition of antibiotics in the cement has been a major advantage in the attempts to provide antibiotic prophylaxis around the implant and, thus, to decrease the rate of infection [19, 20]. However, the amount of antibiotic which is impregnated in the cement can equally affect its mechanical properties. The flexural strength of antibiotic-loaded cement is inferior to that of cement without antibiotics. In addition the cement toughness decreases with excessive amounts of mixed antibiotics. It has been shown that the maximum of 2 g of antibiotic can be safely added to 40 g of cement powder without detrimental effects on its biomechanical properties [21, 22]. However, the addition of antibiotics to the cement and their slow elution in the surrounding tissues in very small quantities can raise other issues, such as toxicity, the development of resistance by the microorganisms, and the development of allergic reactions to the antibiotic, which may be manifested in the form of loosening, in cases of revision with cement loading with the same antibiotics. Furthermore, the choice of antibiotics that can be loaded in the cement is limited since they should not be affected by the heat of polymerization; they should be water soluble and heat stable.

Thickness of the Cement Mantle

The thickness of the cement mantle has been traditionally accepted to be approximately 2–5 mm, especially in the proximal and medial area of the femur and at the tip of the distal end of the implant areas in which the cement is prone to damage after initial loading [23]. This amount of thickness of the cement mantle assures a very satisfactory result both clinically and biomechanically [24]. On the other hand ,certain investigators have shown that the “French paradox,” according to which the thickness of the cement mantle can be as small as 1 mm, could give equally good results [25] (Fig. 3.5). Skinner et al. compared the clinical and radiological 10-year survival of two groups of patients with cemented total hip prostheses. One group had the femoral canal over reamed by 2 mm and the other group had their canal reamed to the same size as the prosthesis. The survival was slightly better in the group of line-to-line reaming. There were significantly more lytic lesions and radiolucent lines in the group of 2 mm-thick cement mantle [26]. In such cases, these canal filling stems, being polished and either taper shaped or rectangular, transfer their loads directly to bone through close cortical contact. Obviously, they are not meant to subside but they offer certain theoretical advantages. By removing most of the weak cancellous

Fig. 3.5 Satisfactory radiological result at 25 years with the "French paradox" principle

bone, they transfer the loads almost directly to the much stronger cortical bone, thus improving the stability of the implant. During insertion of these stems, the orientation and the insertion depth are more accurately obtained. Additionally, these canal filling stems can produce a high intramedullary pressure during insertion, a fact that increases the amount of interdigitation of the cement into the bone, providing a high-quality bone-cement interface [27]. Although certain retrieval studies have suggested that thin mantles are more susceptible to the production of cement cracks, biomechanical studies have shown that the rate of the propagation of fatigue cracks in the cement are independent of the thickness of the cement mantle [28, 29]. These findings are of considerable clinical significance because, through these cracks, wear particles may transverse the interface and enter the bone, initiating the development of osteolysis. Interestingly, Ramaniraka et al. in a study of the fixation of cemented femoral components showed that considerably thicker cement mantles of 5–10 mm could increase micromovement and have a detrimental effect on the implant survival [30].

Surface Texture of the Femoral Component

The influence of the surface texture of the femoral component on the stability of the bone-cement interface cannot be better illustrated than in the case of the Exeter hip arthroplasty. In an attempt to improve the rate of survival of the femoral components, the designers changed the surface texture from polished to matt. However, in a midterm follow-up period, they noticed that, contrary to their expectations, the rate of loosening increased [31]. Having to revert to the original design, they explained that the failure of the matt surfaced implant was due to the fact that the matt surface can wear more easily through abrasion and lead to the development of defects in the cement mantle through which joint fluid with wear particles can lead to destabilization of the cement-implant bond. Massin et al. in a finite element analysis has shown that stresses in a strong implant-cement bond, such as in the cases of femoral stems with a rough surface finish, are predominantly tensile and shear and less of a compression type [32]. Bone cement is tolerant of compression loads but not in tension and shear [33]. Consequently, these types of stresses will, in time, result in damage to the cement-stem interface. Once the cement-implant bond has been destabilized, loosening of the bone-cement interface will follow [34]. Waandres et al. in finite element interface models showed that the majority of plastic displacement was caused by fatigue damage and that this fatigue damage considerably increases the stress levels in the bone [35]. Della Valle et al. has reported that such a rough surface finish adversely affects the survivorship of cemented implants because of loosening and metallic shedding in the bone-cement interface

[36, 37]. On the other hand, in the case of polished texture of the femoral component, the initial micromovement and subsidence of approximately 2 mm takes place gradually over the first 2 years after implantation at the cement-implant interface, finally reaching a stable position, thus protecting the bone-cement interface and avoiding loosening [38]. Numerous attempts to improve the stability of the cement-stem bond have been made by roughening the surface of the implant or by precoating it. A rough surface finish of a femoral stem has consistently produced inferior results. Due to the differences in elasticity between metal, cement, and bone, the repetitive loads which are applied to this construct by the patient's body weight and the contraction of the muscles of the proximal femur make the chances of absolute stability improbable. RSA studies both in vitro and in vivo have shown that perfect stability of the stem does not exist [39].

Shape of the Femoral Component and Metallurgy

The shape of the femoral stem plays an equally important role in the long-term survival of the bone-cement interface. Ideally, a femoral stem should be able to transmit all type of stresses to the surrounding cement and bone, without creating peak forces and excessive micromovement. In the cases of the double taper (Exeter) or triple taper (C stem) collarless design, the axial loading of the implant will convert the axial forces into radial compressive forces at the bone-cement interface. This shape of stem, if combined with a smooth polished surface, will allow for a gradual subsidence and consequent stabilization over the first 2 years after implantation. In a radiostereometric analysis Alfaro-Adrian and Stefansdottir showed that these stems can subside axially from 0.9 to 1.4 mm and into retroversion from 0.4 to 0.5 mm in the first year, followed by stability for the next years [40, 41]. This migration seems to be independent of the thickness of the cement mantle and of the viscosity and type of cement used [42, 43]. There are, in addition, femoral stems designed in such a way that are not intended to subside and, consequently, are extremely dependent on a perfect cementing technique which should provide a cement mantle with no voids (composite beam concept). Alfaro-Adrian and Catani et al. [44] have used radiostereometric methods to study the rate of migration of these stems and concluded that the longitudinal migration is less than in the taper design, ranging from 0.1 to 0.5 mm during the first year, but their movement into retroversion is considerably higher, ranging from 1.0 mm to even 2 mm. These stems initially provide good stability, but their tolerance to long-term migration is not known [45]. Certain non-taper-designed femoral stems are provided with a collar. The collar could be useful in transferring loads from the implant to the femoral calcar and the medial cement mantle, in addition to reducing tensile stresses to the stem and preventing migration [46]. The disadvantage of the collar, however, is exactly this prevention of migration and the settling of the femoral stem in a final stable position. Additionally, in the long term it does not seem to prevent absorption of the calcar. The anatomical shaped stems are designed to fit the overall shape of the femur in a better way, thus allowing for a better centralization of the stem and providing a more symmetrical thickness of the cement mantle. Their anatomical shape and the presence of a collar prevent the subsidence of these stems, but numerous reports, as well as the Swedish Arthroplasty Registry, have shown that excellent and long lasting clinical results can be obtained [47]. Thien and Karholm in an analysis of three different cemented stems have suggested that in cases of femoral stems with rough surface finish, a small-size stem could be a risk factor for debonding and loosening of the bone-cement interface. Similarly, an increased offset and long femoral neck would have the same deleterious effect [48].

The choice of metallurgical construction of the femoral components is equally important for the long-term survival of the bone-cement interface. Titanium alloys, being less tolerant to wear, should not be used for stems with rough surface finish. As described, these stems are prone to creating tensile stresses that can readily lead to wear, of the abrasion type, and the production of wear particles. This is illustrated by a number of reports of the inferior performance of unpolished cemented titanium stems [49, 50].

Cemented Fixation of the Acetabulum

Cemented fixation of the acetabular component in total hip arthroplasty is a widely accepted method. The principles of correct cementation technique apply for the acetabulum as well as for the femoral stem. Despite the increased tendency over the last few years to prefer cementless fixation of acetabular components, recent reports from the Swedish National Hip Arthroplasty Register and the National Joint Registry for England and Wales showed that cemented fixation produces better and longer lasting survival with intact bone-cement interfaces compared to cementless fixation [7, 50].

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