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Summary

Acrylic cement fixation of artificial joints has proven to be a successful concept. Potential advantages of cementless fixation notwithstanding, cement fixation can still, at this time, guarantee the best predictable and most dependable results for a period of at least 10-15 years, as witnessed by long-term follow-up studies in large centres. Nonetheless, acrylic cement is still the 'weak link' in the joint replacement structure, due to a number of adverse physical properties and flaws in the fixation technique.

It is the objective of this article to propagandize optimization of the cement fixation technique as a route to improvement of clinical results. Whereas several authors have suggested ways to improve the cement and interface strength, and to improve component design to reduce the stresses, this article is aimed in particular at reducing the surgical trauma to bone during the operation, as caused by the heat of cement polymerizing.

Temperatures in cement and bone are quantified based on thermal finite element analysis of several cases, including intramedullary fixation, acetabular cup fixation and cement pressurization into trabecular bone. The influences of geometrical and material parameters on the temperature patterns are evaluated. Preventive measures that can be taken to reduce the peak temperatures in bone and to reduce the bone regions liable to direct and indirect thermal necrosis are quantified as to their effects.

Introduction

There is little dispute that acrylic cement (PMMA) has triggered the success of total hip replacement when introduced by Charnley [1] more than 20 years ago as a prosthetic fixation medium. Whereas more recently developed fixation materials have potential advantages and deserve further experimentation, it is evident from recent long-term follow-up studies in large centres that at this time acrylic cement fixation can still guarantee the best predictable and most dependable results for a period of at least 10-15 years postoperatively [2].

Nonetheless, PMMA is not an ideal material for implantation and load-bearing. Its physical properties are such that certain problems may arise in the structure of the replacement, postoperatively in the course of time. A relatively small percentage of prostheses loosens preliminary, while the total expected life-span of the total hip replacement normally limits its application to older patients.

Acrylic cement is usually considered as the 'weak link' of the replacement structure, in view of its assumed role in the loosening mechanisms. This has motivated some to return to cementless fixation, while others experiment with alternative fixation materials such as porous coatings. It must be appreciated, however, that any fixation material, by the nature of its function, will constitute a 'weak link'. Another route for improving upon the present results, and probably more rewarding at the present time, is optimization of the cement fixation concept. This alternative requires an assessment of failure mechanisms in relation to acrylic cement properties.

The acrylic cement mixture is snap-curing and void-filling, thus allowing the surgeon to mold a well-fitting fixation bed for the prosthesis during the operation. The (cured) cement mantle is relatively soft, so it enables a smooth load transmission from the prothesis to the bone [3]. These are the favourable physical properties on which its success as a fixation material is based. On the other hand, however, PMMA is a rather weak material, while the strength of cement can even become less by a great number of surgical variables [4] and by gradual degradation in the body [5]. An important factor is the homogeneity of the cement mass, frequently compromised by a faulty cementing technique. As a consequence, the strength of the cement mantle is not always adequate to withstand the stresses generated by the joint loading without (local) fracture and bone-interface loosening.

Other adverse effects are caused by dimensional changes of the curing mass [6], leakage of residual, cell-toxic monomer into the bone [7] and finally the chances of thermal bone necrosis due to the heat generated in the curing process [8].

Precisely how these adverse properties of acrylic cement compromise the integrity of the joint-replacement structure in the course of time is not well known. Cement failure is thought to contribute to aseptic loosening [9], while fractured cement particles in the presence of micromotion have been shown to initiate a process of bone resorption at the interface [10]. Inadequate cementing and cement failure in any case increase the chances of component failure [11].

The initial integrity of the structure is compromised by bone necrosis, occurring in a narrow zone close to the cement-bone interface during or directly after fixation, caused by vascular damage, monomer leakage and thermal effects. Although this dead bone layer is replaced, at least to some degree, after resorption and revascularization have taken place, it appears evident that the amount of bone damaged significantly influences the postoperative strength of the structure. According to Feith [8] the heat of polymerizing is the most destructive of the bone-damaging factors.

In view of these effects, it appears that optimization of the cement fixation concept, at least where the cement itself is concerned, can follow three different routes: first, the cement and interface strength can be increased, second, the cement and interface stresses can be reduced and third, the amount of bone necrotized in the fixation procedure can be reduced. The first objective has been substantiated by the introduction of improved cementing techniques, as for instance water-peg cleaning of the bone bed and pressurization of cement, using a medullary canal plug and a cement syringe [12-15]. These methods have been shown to improve the cement and interface quality (strength), and consequently the clinical results, considerably [15]. The second objective can be reached by applying methods of stress analysis to generate mechanical criteria for design and fixation (placement) of the prostheses [3, 16, 17]. The third objective can be met by methods to reduce temperatures in bone during prosthesis fixation, requiring an assessment of the heat generation and conduction process [3, 18], which is the subject of the remainder of this article.

Heat generation in acrylic cement

The acrylic cement dough cures as a result of the polymerization of the monomer fraction in the cement mixture. In the polymerization process heat is generated to a total amount of $q = 508$ Joules per cm^3 of monomer [19]. This heat is generated gradually, in amounts proportional to the polymerization rate. It is temporarily stored in the acrylic cement mass and gradually released by heat flow through the mass and the surrounding structures.

Let us assume for the moment that a mass of cement cures isolated from its environment, without any release of heat. According to the law of energy conservation, all the heat generated is stored within the mass in this case. If the monomer volume fraction in the mixture is denoted by v_m, then the total amount of heat generated within the mixture follows from $Q_g = v_m q$ (J/cm^3). The heat stored within the mixture (Q_s) depends on its heat capacity (C) and the total temperature increase (ΔT): $Q_s = C\Delta T$ (J/cm^3). Since $Q_s = Q_g$ in this case of ideal isolation, it follows directly that $\Delta T = v_m q/C$ (°C). Hence the temperature increase of the isolated cement mixture depends on its heat capacity (C) and the monomer fraction by volume (v_m), since the total amount of heat generated per unit volume of monomer (q) is a fixed value.

Using the appropriate heat capacity values [3], it can easily be established that the temperature *increase* in a mixture consisting of monomer only ($v_m = 1$) would be about 339°C. In a 'normal' cement mixture ($v_m \approx 0.35$) the increase will be about 96°C. If the cement mixture is pressurized into trabecular bone, a composite of bone, polymer and monomer is obtained. If the pore volume fraction of the bone is 0.65 and the cement is of 'normal' composition, then the monomer volume fraction in the composite is $v_m \approx 0.23$. Using again the appropriate heat capacity value of this composite [3], it follows that the temperature *increase* in the composite is about 55°C. So in normal cement and in a pressurized cement-bone composite, assuming an initial temperature of 37°C, the temperatures may maximally rise to 133°C and 92°C, respectively.

In reality, the heat generation occurs gradually, while at the same time heat flows from the cement mass to the surrounding structures. This process is illustrated in Figure 1, where temperatures as functions of time are shown in the middle and at the surface of a curing ball of cement. The centre-point temperature development is approximately equal to that in the ideally isolated case, because the heat generation occurs too fast to allow for a significant amount of heat to flow from this region before the polymerization process has ended. In points closer to the surface, how-

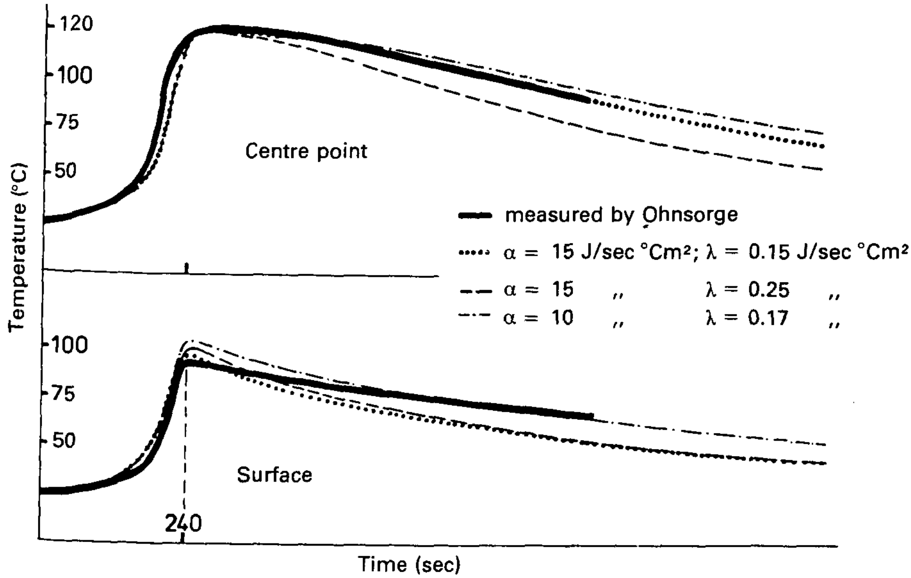

Fig. 1: Temperatures as functions of time in the centre and at the surface of a curing ball of acrylic cement [3]. The heat is generated during the process of polymerization and stored in the cement mass. The peak surface temperature is lower than in the centre, because heat is already released from the superficial part of the ball during the polymerization. The different curves are the results of measurements and of analytical studies, in the latter case assuming several values for the surface heat conductivity (α) and the cement conductivity (λ).

ever, the heat flow to the environment reduces the heat to be stored and thus reduces the peak temperatures reached.

In this example, the temperatures are not distributed homogeneously throughout the structure. However, in this case too, the local temperatures are governed by the law of energy conservation. In a small piece of the material, in a given small period of time, it must hold that

heat stored = heat generated + heat flow in – heat flow out

Thus, the temperature development in a specific point of the cement mass is governed by a balance between separate phenomena and, through the heat flow, also influences its neighbouring points. This balance is affected by a large number of parameters that can be categorized in geometrical ones (e.g. dimensions), material ones (e.g. monomer fraction, heat capacity, conductivity) and boundary conditions (e.g. surface conductivity, environmental temperature). Hence, the actual temperatures occurring in cement and bone during prosthesis fixation greatly depend on the circumstances.

It is the objective of the present studies to assess the influences of these circumstances and to evaluate the effects of measures that can be taken to reduce the temperatures. Use is made of an analytical computer technique, the finite element method. In using this technique the structure to be analysed is mathematically divided into small parts, elements, and for each element the heat balance is solved, based on mathematical descriptions of the relevant parameters mentioned before.

The details and experimental verification of this technique have been treated elsewhere [3].

It is important to realize that due to the schematic mathematical descriptions applied, the results are estimates of the real temperatures, by a certain degree of approximation (cf. Fig. 1). A versatile feature of this analytical method, however, is the ease with which conditions can be changed to evaluate their effects.

Bone temperatures during implant fixation

The first model regarded concerns intramedullary fixation in general. The characteristics of the model are shown in Figure 2. Evidently, the case in which the whole medullary canal is filled with cement presents a worst-case situation as far as the temperatures are concerned. Figure 3 shows temperatures as functions of time for this case at several points in the cement and the bone. The peak cement temperature (point 1) T_c is 121°C, the peak bone temperature (point 8) T_b is 59°C. These values are dramatically reduced when a metal stem is inserted, because less cement is present and the stem functions as a 'heat sink'. The consequences of this for two stem sizes are given in Table 1, showing peak cement and bone temperatures and the thickness of the bone region that reaches a temperature of 50°C or higher.

Other measures that reduce the temperatures, all evaluated in Table 1, are reducing the monomer fraction by increasing the powder-to-liquid ratio (which reduces the amount of heat generated per unit volume of cement), adding an aqueous gel [8, 20] to the cement (which also reduces the heat generated per unit

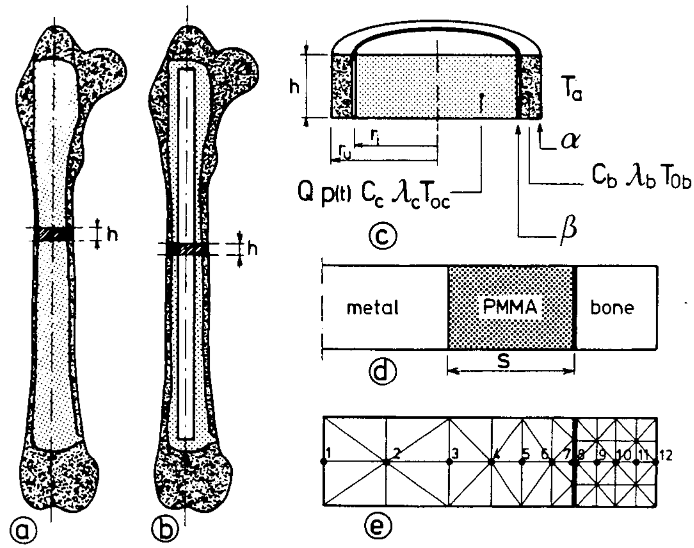

Fig. 2: A model for thermal analyses of acrylic cement during intramedullary prosthetic fixation [3]. Cases in which the whole bone cavity is assumed to be filled with cement (a), and in which implants of different sizes are inserted (b), were simulated. The local (axisymmetric) model with the various parameters is shown (c and d), as well as the mesh used for the finite element analysis (e).

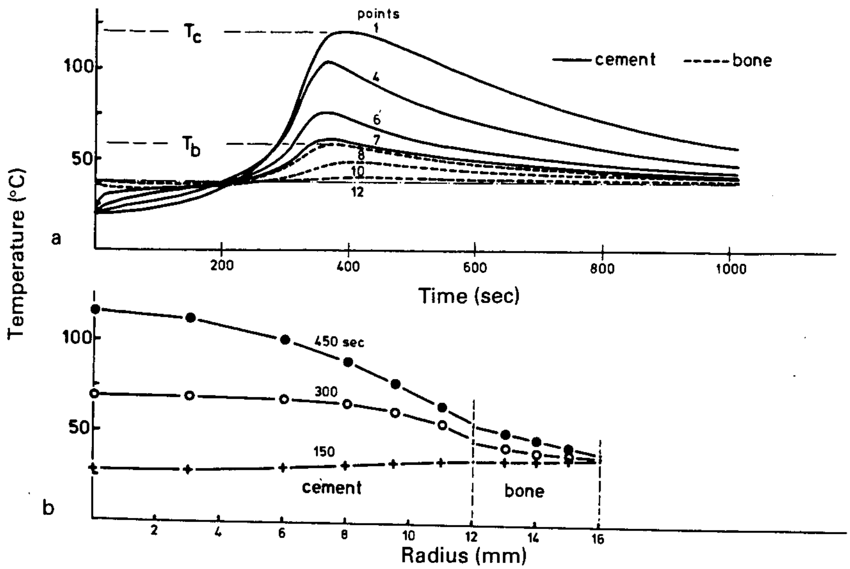

Fig. 3: Results of the analysis simulating the case in which the whole bone cavity is filled with cement (the reference case, cf. Table I). The upper graph (a) shows temperatures as functions of time at different locations in the cement and the bone; the numbers refer to the positions shown in Figure 2e. The peak temperatures occurring in cement (T_c) and bone (T_b) are indicated. The lower graph (b) shows temperatures at different times as functions of the radius, as distributed throughout the cement and the bone [3].

volume of cement and increases the heat capacity of the mixture), and chilling the bone prior to cement insertion.

Feasible measures not shown are refrigerating the prosthesis, which is not very

Table I: Peak temperature values occurring in the cement (T_c) and the bone (T_b) as evaluated in the model shown in Figure 2 in different circumstances. Also shown is the penetration depth of the 50°C isotherm (the thickness of the bone region that reaches a temperature of 50°C or higher). Changes of the values are given as well, relative to the reference case.

	Peak temperatures (°C)				Penetration depth	
	T_c	% change	T_b	% change	mm	% change
1. Reference case* (Fig. 3)	121	0	59	0	1.6	0
2. Metal implant of 12 mm diameter	96	–21	54	– 9	1.3	– 20
3. Metal implant of 20 mm diameter	50	–58	44	–26	0	–100
4. Powder-to-liquid ratio increased from 2 to 2.5	110	– 9	56	– 5	1.3	– 20
5. 30% aqueous gel added	81	–33	50	–15	0.3	– 80
6. Bone chilled from 37°C to 25°C	116	– 4	48	–19	0	–100
7. Dog femur (15/20 mm diameters)	–	–	54	– 9	–	–
8. Rabbit femur (7.5/10 mm diameters)	–	–	48	–19	0	–100

* Initial bone and environment temperatures 37°C, initial cement temperature 20°C; outer bone diameter 32 mm, inner bone diameter 24 mm; other parameter values as in [3].

effective in view of the distance between prosthesis and bone [3], and chilling the cement mixture prior to insertion, which potentially compromises the cement strength.

Another important effect indicated in Table I is that of size (dog and rabbit). Scaling down the dimension of the structure significantly alters the temperature values. Hence (thermal) experiments with acrylic cement in small animals are not representative of the human situation.

Thermal damage of bone

Although the bone temperatures in the previous analysis reach alarming values (up to almost 60°C) only in the most unfavourable circumstances, while the layer of bone that is subject to temperatures over 50°C is only small, the question remains what temperatures are just allowable before necrosis sets in.

In addressing this question a differentiation must be made with respect to direct thermal damage of collagen, of bone cells and of the vascularity, damage to the regenerative capacity of the bone and indirect thermal damage.

It appears intuitively obvious that not only the temperature, but also the exposure time must play a significant role in direct thermal necrosis of bone, which was indeed confirmed in a number of studies. Moritz and Henriques [21] established that the temperature threshold level for epithelial cell necrosis is dependent on exposure time. Their findings suggest that, for instance, necrosis occurs almost instantly at temperatures of 70°C and over, 30 seconds after exposure to a temperature of 55°C, but only five hours after exposure to 45°C. Lundskog [22] roughly confirmed these trends for bone cells, although finding a slightly lower threshold level (for example: bone cell necrosis after 30 seconds at 50°C). He also established that the regenerative capacity of the bone tissue is only damaged after exposure to temperatures of 70°C and over. Denaturation of collagen (after which regeneration of bone is improbable) has been reported at temperature threshold levels between 56°C and 70°C [3]. Not much is known about threshold levels for vascular damage. Moritz and Henriques [21] report that in the skin tissue vascular damage occurs after exposure times that are approximately 50% of those for cell necrosis.

When comparing the results of the previous analyses (Table I) with the threshold of Moritz and Henriques [21], it follows that in the worst-case situation (the whole medular canal filled with cement and no implant present) a bone layer of about 0.5 mm would be liable to cell necrosis and of 1 mm to vascular damage. In other cases no damage of any kind would result. According to the threshold levels proposed by Lundskog [22], a bone layer of about 1.5 mm would become necrotic in the worst-case situation and a layer of 1.1 mm when a thin metal implant is present in the bone. Damage to the regenerative capacity of the bone would not occur in any case (according to Lundskog's threshold), while only in the worst-case situation could collagen denaturize, in a zone of about 0.5 mm bone (if this occurs at a temperature of 56°C).

Although, as follows from this discussion, no exact criteria can be set, it appears reasonable to use the penetration depth of the 50°C isotherm in the bone (Table I) as an indication for the amount of bone liable to necrosis. In the case of intramedullary fixation this amount is only small (Table I) or non-existent when the cement layer is kept relatively thin. In any case, relatively simple measures can be taken to diminish the chances of direct thermal damage altogether. These measures are the more important where fixation of other than intramedullary prostheses are concerned, when indirect thermal damage is considered and for cases in which cement is pressurized into bone.

Due to the specific geometry of the cement layer and the inferior heat capacity and conductivity properties of polyethylene when compared to metal, the temperatures feasible during acetabular cup fixation are higher. The temperature patterns obtained in a thermal analysis of this structure were directly translated to the threshold criteria previously discussed and the results are shown in Figure 4. The bone region liable to necrosis may extend to a thickness of about 6 mm in this case, while maximum bone temperatures of about 65°C were found at specific locations. Also indicated in this Figure is the region of cement that reaches a temperature higher than 100°C, where local evaporation of monomer may occur, resulting in small gas entrapments that decrease the cement strenght. Of course, these results represent only a specific and schematized configuration and should not be interpreted in too absolute a sense. But when compared with the previous results on a relative basis, it is evident that direct thermal necrosis is more likely to occur on the acetabular side of a total hip replacement.

The hypothesis that indirect thermal effects may play a role in bone necrosis during implant fixation stems from a comparison between the results obtained by Feith [8] and those of thermal analyses simulating his experiments [3]. Feith inserted different kinds of cement into rabbit femurs and evaluated the necrotic zones histologically after sacrifice. His results are summarized in Figure 5. In all five groups shown the medullary canal was reamed and suctioned (R + S). In group 1 no cement was inserted, in group 2 a prepolymerized rod (residual monomer and no heat) was put into the bone, in group 3 'normal' cement was used (residual monomer and heat); cement without a catalyst was used in group 4 (overdose monomer and no heat), while in group 5 an aqueous gel cement [20] was put into the bones (about an equal amount of residual monomer but significantly less heat as compared to normal cement). Feith [8] concluded that the heat of polymerizing is the principal causative factor for endosteal bone necrosis (comparing groups 2, 3 and 5), although monomer, when administered in an overdose (group 4), has a high potential for extensive necrotic effects.

However, from the thermal analysis in which these experiments were simulated it follows that in the case of the rabbit femur filled with 'normal' cement (Table I) the temperature patterns are such that direct thermal necrosis is highly unlikely. This analytical case can be considered as an additional 'experimental' group (Fig. 5, group PR, polymerization heat but no monomer). When both the experimental and the analytical findings are accepted as true, the only possible explanation for the combined results in the six groups illustrated in Figure 5 is that the bone (and/or cement) temperature has an indirect influence on bone necrosis by enhancing the cell-toxic effects of the monomer.

So according to this hypothesis the temperatures may, apart from having a direct effect on thermal bone damage if maintained long enough at specific high values, also have an indirect effect via the cell-toxic monomer, even at much lower values. Although it must be admitted that the evidence for this hypothesis is 'circumstantial' and not abundant, it strongly suggests that any measure taken to reduce the temperatures during implant fixation would help the bone to recover rapidly from the trauma administered.

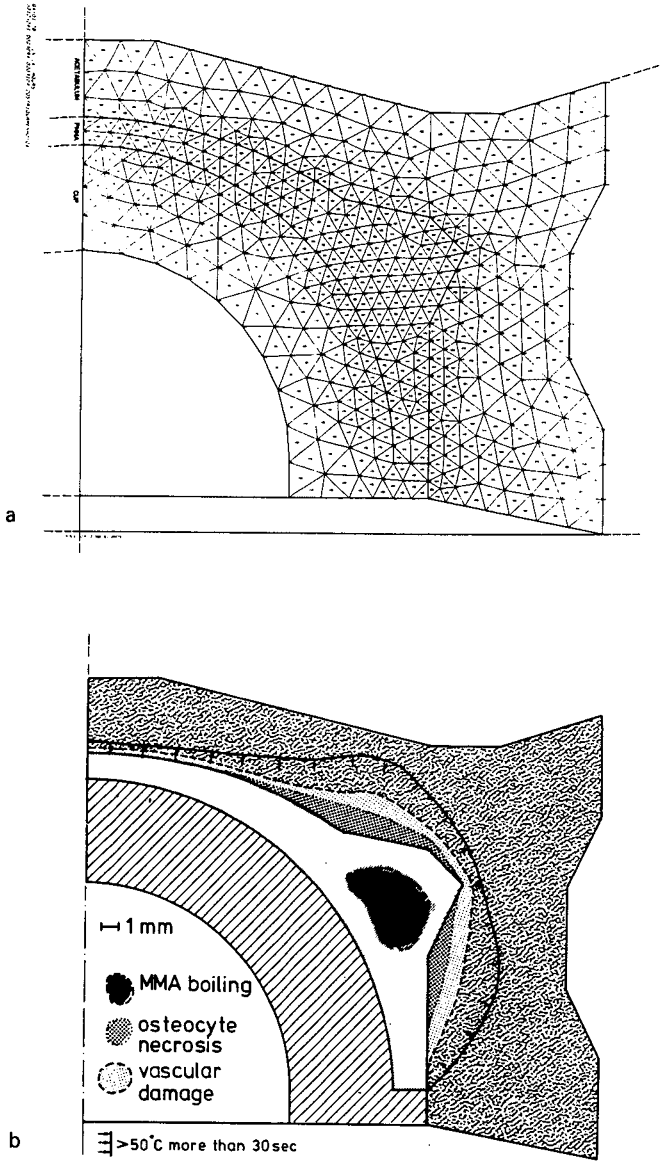

Fig. 4: Finite element mesh (a) representing a section of an (axisymmetric) model to analyse heat generation and conduction during acetabular cup fixation [3]. *The results of the analysis (b) are interpreted according to threshold criteria for thermal damage. In the cement the region reaching temperatures higher than 100°C (MMA evaporation) is indicated. In the bone the zones that would be liable to osteocyte necrosis and vascular damage according to Moritz and Henriques* [21] *are indicated, as well as the (larger) zone that would be liable to necrosis according to Lundskog* [22].

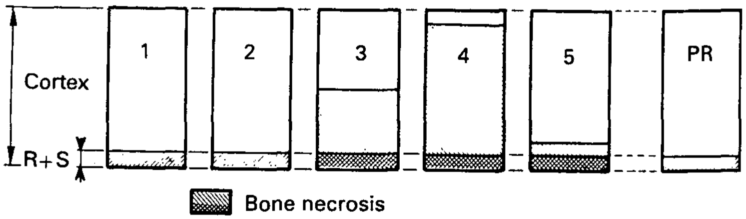

Fig. 5: Schematic representation of results reported by Feith [8], groups 1-5 and an additional group (PR) representing the present analytical results. The shaded areas (schematically) represent the extent of the endosteal necrosis in the cortex. Variables are: reaming and suction (RS), residual monomer (M) and heat of polymerizing (H). The groups represent: 1) RS, 2) RS + M, 3) RS + M + H, 4) RS + M (overdose monomer), 5) RS + M + H (little heat), and the group PR simulates: RS + H.

Cement pressurization into trabecular bone

In order to enhance the strength of the cement-bone interface it has been suggested to pressurize the cement into trabecular bone in a viscous stage [14]. As discussed previously in the introduction, a composite of bone, monomer and polymer is then obtained, in which case the maximum temperature *increase* (no heat flow) could be about 55°C (assuming the pore volume fraction of the bone to be 0.65; if the pore volume is 0.50, for instance, the maximum temperature increase could be about 48°C). Given an initial temperature of 37°C, a maximum temperature of 92°C could result; if the initial temperature were 20°C, the maximum temperature could be 75°C. If these maximum temperature values occurred, irreversible bone death would be instantaneous. However, whether these maximum values are actually reached again depends on the energy balance between heat generation, storage and flow, thus on the many parameters by which these phenomena are determined.

For this case too, a thermal analysis was carried out [18], simulating fixation of a tibial knee prosthesis (Fig. 6a). An example of results is shown in Figure 6b, giving temperature *increases* (over the initial temperature) as functions of time at specific locations in the solid cement layer, the penetrated cement-bone composite and the non-penetrated trabecular bone (see also Fig. 7). The pore volume fraction of the bone is assumed to be 0.65 in this case. If we assume the initial temperature of the bone to be 20°C (room temperature), the hottest point in the bone reaches a temperature of 74°C, while a zone of about 4.0 mm (almost the full penetrated region) reaches temperatures of 50°C or higher (Table II).

As indicated in Table II, these values increase to 80°C and 8.5 mm if the penetration depth is 10 mm, while a maximum bone temperature of 42°C results if it is not penetrated at all.

Again these results should not be regarded in too absolute a sense, but be interpreted on a relative basis, in view of the schematized character of the model configurations applied. The greatest asset of these methods is the ease with which parameters can be changed to evaluate their influences. As in the previous analysis, Table II shows the effects of measures that can be taken to reduce the temperatures, such as reducing the solid cement layer from 5 to 2.5 mm, chilling the bone from

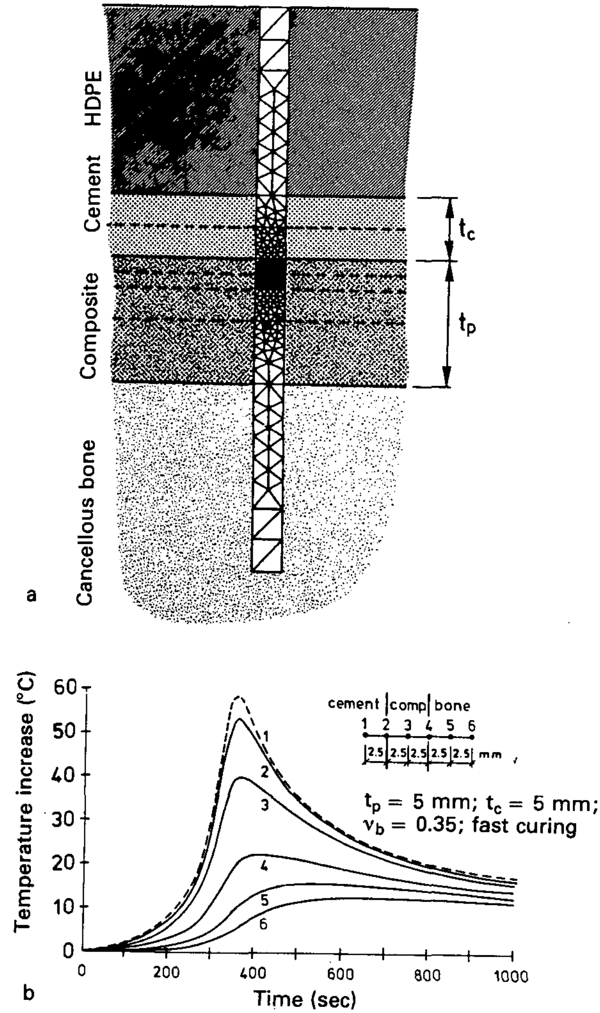

Fig. 6: A model to analyse heat generation and conduction during implantation of a tibial knee component, simulating cement pressed into trabecular bone (a). The finite element mesh is shown, the solid cement layer thickness (t_c) and the penetrated bone region (t_p) are indicated; HDPE = high density polyethylene. Temperature increases as functions of time (over the initial temperature) are shown (b) as evaluated in this model at different locations in the cement, the cement-bone composite and the unpenetrated bone. The parameter values are those of the reference case (cf. Table II).

20°C to 15°C and to 10°C prior to fixation, respectively, and reducing the autoacceleration rate in the polymerization process [3]. Although these measures are all effective in reducing the peak temperatures and the extent of the bone region liable to necrosis, none of them completely abolishes the chances of direct thermal damage, let alone those of indirect thermal damage. However, if a few of these measures are combined, the temperatures will be reduced dramatically, as indicated in Table II and illustrated in Figure 7. In this case, direct thermal damage is highly unlikely.

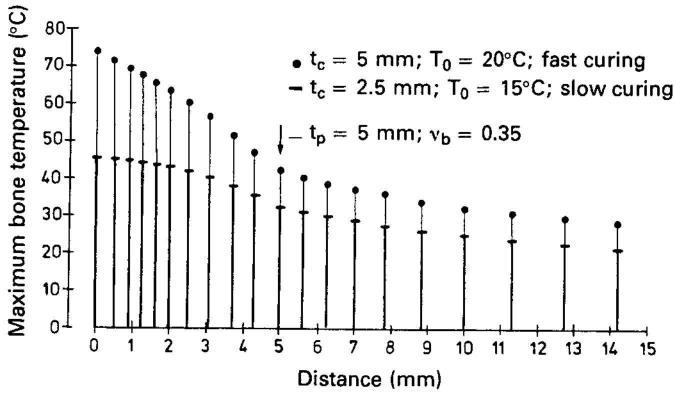

Fig. 7: Peak temperatures reached at several locations in bone as evaluated for the reference case (•) and the case in which three temperature-reducing measures (case 8, cf. Table II) are combined (–). The numbers on the horizontal axis represent the distance in bone from the solid cement/penetrated cement interface. A dramatic reduction of the temperatures can be obtained in this way, such that direct thermal necrosis becomes highly unlikely.

Table II: Peak temperature values occurring in the bone as evaluated in the model of Figure 6a in different circumstances. Also shown is the penetration depth of the 50°C isotherm. Changes of the values are given as well, relative to the reference case. All measures are single ones, except nr. 8, which is a combination of nrs. 4, 5 and 7.

	Peak bone temperature		Penetration depth	
	°C	% change	mm	% change
1. Reference case* (Figs. 6b and 7)	74	0	4.0	0
2. 10 mm penetration depth (t_p)	80	+ 8	8.5	+112
3. No penetration	42	−43	0	−100
4. Solid cement layer (t_c) reduced from 5 to 2.5 mm	62	−16	3.5	− 12
5. Bone and cement chilled from 20 to 15°C	69	− 7	3.2	− 20
6. Bone and cement chilled from 20 to 10°C	64	−13	2.5	− 37
7. Slow polymerization (auto-acceleration reduced)	62	−16	3.2	− 20
8. 4, 5 and 7 combined (Fig. 7)	46	−38	0	−100

* Initial cement and bone temperatures 20°C, solid cement layer (t_c) 5 mm, penetration depth (t_p) 5 mm, bone pore volume fraction 0.65; other parameters see [3, 18].

Discussion

Clinical results have shown that acrylic cement fixation of artificial joints is a viable concept [2]. It appears reasonable to assume that optimization of this concept, concerning component designs and fixation techniques, will even improve the results and increase the average expected life-span of the replacement structure. This assumption is substantiated by clinical evidence [15].

Optimization of the cement fixation concept should not only include the augmentation of cement and interface strength [12-14] and the reduction of stresses [16, 17], but should also be directed towards reducing the surgical trauma to the bone. There is abundant evidence that monomer leakage and the heat of cement polymerizing are important factors in bone necrosis at the cement-bone interface [7, 8]. In addition, as discussed here, there are strong indications that both factors have an interrelated effect. Hence, reducing the bone temperatures occurring during implant fixation appears to be an important step towards reducing the trauma and thus towards optimization of the structure.

The temperatures occurring in bone depend on a great number of parameters that govern the heat generation and conduction processes. In addition, the temperatures vary strongly from place to place in the structure, both of which aspects explain the great variety in experimental temperature measurement results reported in the literature [3]. Analyses based on finite element methods provide for approximative temperature data in schematized models, but, contrary to direct experiments, are rather instrumental in differentiating the circumstances and in evaluating the relative effects of the essential parameters and temperature reducing measures.

The results of these analyses should be interpreted in a relative rather than an absolute sense, in view of the simplifying assumptions applied in the models. Nonetheless, verification of the model predictions by experiments was satisfactory [3].

The analytical results indicate, qualitatively speaking, that chances of direct thermal necrosis during intramedullary stem fixation are slight when the cement layer thickness is kept small, that it is more probable in acetabular cup fixation and that it appears to be unavoidable after cement pressurization into trabecular bone, unless rigorous preventive measures are taken. Several possible measures were evaluated as to their effects on the temperatures and it was found that chilling of bone prior to fixation is quite effective, a measure that can be simply realized with the water peg during cleaning of the bone bed. A combination of measures (chilling the bone, limiting the solid cement mass, limiting the penetration depth and reducing the auto-acceleration rate in the polymerization process) diminishes the chances of direct thermal damage completely, even after cement pressurization. If indirect thermal damage plays a role as well (and this is strongly suggested) then these combined preventive measures would be of importance in all cases of cement fixation.

References

1. Charnley, J. (1970): *Acrylic Cement in Orthopaedic Surgery,* E. and S. Livingstone, Edinburgh.
2. Consensus Development Panel (1982): NIH-consensus paper: total hip joint replacement. *J. Am. Med. Assoc. 248,* 1817.
3. Huiskes, R. (1979): Some fundamental aspects of human joint replacement. *Acta Orthop. Scand. 185 (Suppl.).*
4. Lee, A. J. C., Ling, R. S. M. and Vangala, S. S. (1978): Some clinically relevant variables affecting the mechanical behavior of bone cement. *Arch. Orthop. Traum. Surg. 92,* 1.
5. Kusy, R. P. (1978): Characterization of self-curing acrylic bone cements. *J. Biomed. Mater. Res. 12,* 171.

6. de Wijn, J. R., Driessens, F. C. M. and Slooff, T. J. (1975): Dimensional behaviour of curing bone-cement masses. *J. Biomed. Mater. Res. 6,* 99.
7. Willert, H. G., Ludwig, J. and Semlitsch, M. (1974): Reaction of bone to methacrylate after hip arthroplasty. *J. Bone Jt. Surg. (Boston) Ser. A 56,* 1368.
8. Feith, R. (1975): Side-effects of acrylic cement, implanted into bone. *Acta Orthop. Scand. 161 (Suppl.).*
9. Stauffer, R. N. (1982): Ten-year follow-up study of total hip replacement. *J. Bone Jt. Surg. (Boston) Ser. A 64,* 983.
10. Willert, H. G. and Semlitsch, M. (1976): Problems associated with the cement anchorage of artificial joints. In: *Advances in Artificial Hip and Knee-Joint Technology,* p. 325. Editors: N. Schaldach and D. Hohmann. Springer, Berlin.
11. Chao, E. Y. S. and Coventry, M. B. (1981): Fracture of the femoral component after total hip replacement. *J. Bone Jt. Surg. (Boston) Ser. A 63,* 1078.
12. Slooff, T. J. J. H. (1969): A cement syringe. *Acta Orthop. Belg. 35,* 1012.
13. Halawa, M., Lee, A. J. C., Ling, R. S. M. and Vangala, S. S. (1978): The shear strength of trabecular bone from the femur and some factors affecting the shear strength of the cement-bone interface. *Arch. Orthop. Traum. Surg. 92,* 19.
14. Miller, J., Krause, W. R., Krug, W. H. et al. (1981): Low viscosity cement. *Orthop. Trans. J. Bone Jt. Surg. 5,* 352.
15. Harris, W. H., McCarty, J. C. and O'Neill, D. A. (1982): Femoral component loosening using contemporary techniques of femoral cement fixation. *J. Bone Jt. Surg. (Boston) Ser. A 64,* 1063.
16. Crowninshield, R. D., Brand, R. A., Johnston, R. C. and Milroy, J. C. (1980): An analysis of femoral component stem design in total hip arthroplasty. *J. Bone Jt. Surg. (Boston) Ser. A 62,* 68.
17. Huiskes, R. (1983): Design, fixation and stress analysis of permanent orthopedic implants: the hip joint. In: *Interaction Between Musculoskeletal Tissues and Orthopedic Devices,* Chapter 11. Editors: G. Hastings and P. Ducheyne. CRC Press, Boca Raton. [In press].
18. Huiskes, R. and Slooff, T. J. (1981): Thermal injury of cancellous bone, following pressurized penetration of acrylic cement. *Orthop. Trans. J. Bone Jt. Surg. 5,* 277.
19. Trommsdorf, E. (1963): Polymerisate der Acrylsäure, ihrer Homologe und Derivate. In: *Chemie und Technologie der Kunststoffe.* Vol. II, p. 541. Editors: R. Houwelink and A. J. Staverman. Akademische Verlagsgesellschaft, Leipzig.
20. de Wijn, J. R. (1982): *Porous Polymethylmethacrylate Cement,* Medical dissertation, University of Nijmegen.
21. Moritz, A. R. and Henriques Jr., F. C. (1947): Studies of thermal injury II. *Am. J. Pathol. 23,* 695.
22. Lundskog, J. (1972): Heat and bone tissue. *Scand. J. Plast. Reconstr. Surg. 9 (Suppl.).*