A method for bone-cement interface thermometry

An in vitro comparison between low temperature curing cement Palavit® and Surgical Simplex® P

Søren Harving, Kjeld Søballe and Cody Bünger

Previous temperature measurements at the bone-cement interface have all shown large variations. To evaluate more precisely the temperature profiles during cement curing, a new experimental model was developed. Eight thermocouple electrodes at the bone-cement interface for each test specimen were used for continuous temperature recordings. Temperature profiles of Palavit® was compared with those of Surgical Simplex® P in an in vitro model using isolated pig femurs. Defects of 12 × 17 mm in the femoral metaphysis were filled with cement. In six tests with each type of cement, Palavit® peaked at a temperature of 50 ± 0.5 °C, whereas Surgical Simplex® P peaked at 60 ± 0.7 °C. Core temperatures reached peak values of 70 ± 0.8 °C and 95 ± 2.2 °C for Palavit® and Surgical Simplex® P, respectively.

Biomechanics Laboratory, Orthopedic Hospital, Århus, Denmark
Correspondence: Dr. Søren Harving, Volshøjvej 8, DK-8240 Risskov, Denmark. Tel. +45-86 212827
Submitted 91-01-12. Accepted 91-06-27

Temperatures at the bone-cement interface have been measured to be in the range of 37° to 120 °C during polymerization of the cement (Sloff 1971, Meyer et al. 1973, Debrunner 1974, Reckling and Dillon 1977, Larsen and Ryd 1989). These measurements have been performed with only one probe for each measurement, which might explain the great variations in results observed within each study.

The aim of the present study was to develop a technique for exact bone-cement interface temperature recording in a standardized in vitro model and to compare temperatures at the bone-cement interface during polymerization of a new low temperature curing bone cement with ordinary cement.

Materials and methods

The two cement types used were Palavit® (Schering-Plough) and Surgical Simplex® P (Howmedica). Both cement types were stored at the same and constant room temperature before use. To minimize alterations in heat production due to disproportion in monomer/polymer content, only whole standard portions of cement were used for each experiment. The two cement types have equal proportions of monomer/polymer content while Palavit® has a lower viscosity profile than Surgical Simplex® P.

Twelve isolated pig femurs were heated in a water bath with a constant temperature of 38 °C. In the medial femoral condyle, 8 mm proximal to the distal limit of the epiphysis, a 1.6-mm guide wire was inserted. A cannulated drill expanded the hole to 12 mm in diameter and 17 mm in depth. The direction of the drill was perpendicular to the femoral shaft and parallel to the weight-bearing surface of the femoral condyle. To prevent uncontrolled heat production, which might result in differences in bone temperature, the drilling was performed manually with a maximum of 30 rotations/min. The volume of the defect defined the amount of cement used in each experiment. The drill hole was cleansed with 20-mL saline solution and dried with a suction catheter. Through each of two separate 1.5-mm canals, four copper-constantan thermocouple electrodes (thickness 0.01 mm, nylon-coated) were introduced into the bore hole, where approximately 4 mm of the tip was bent to an acute angle and visually retracted and placed in contact with the bone. To prevent dislocation, the thermocouples were fixed at the exit of the two accessory canals with small wooden wedges. An additional two electrodes were looped around the guide pin to measure core temperature in the cement. The temperatures were recorded using a 10-canal microprocessor thermometer (accuracy ± 0.1 °C; Comark 6600, Comark Electronics Ltd, West Sussex, U.K.) connected to an IBM personal computer. Signals from the temperature measurements were recorded from each canal every 10 seconds. The cement was manually mixed in a bowl according to the manufacturer's instructions, and was injected into the hole after 1.5 minutes. Using a special tool, a compression of an arbitrary 25 N was applied. A Teflon piston was used to minimize heat absorption from the cement. The temperature recording was continued for 15 minutes. Correct positioning of the thermocouple electrodes was radiographically verified after each experiment. Radiographs were obtained both parallel to the long axis of the cement plug and perpendicular to the same axis. The room temperature was measured at the beginning of every experiment. The experiments were performed alternating between the two types of cement to reduce any systematic errors (e.g., a shift in room temperature during the experiments, which could result in a different room temperature when testing the two cement types). Following temperature measurements, the cement plugs were sawed in quarters to evaluate macroscopic air entrapment in the cement.

The Student's *t*-test was used for the statistical calculations.

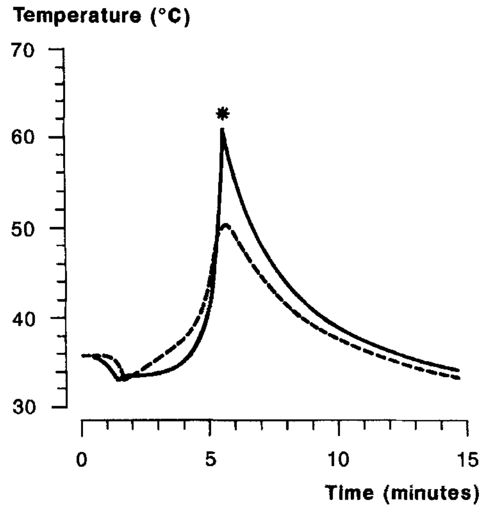

Figure 1. Mean values from bone-cement interface temperature measurements during the curing of the cement. Correction has been made to simulate identical peak time for all the measurements for each type of cement.
——— Surgical Simplex® P bone cement.
------ Palavit® bone cement, *$P < 0.01$.

Results

After removal from the water bath, the bones were prepared at room temperature, which resulted in a bone temperature at the beginning of the cement mixing of 36.0 °C ± 0.1 °C (mean ± SEM) for both cement types (Figure 1). The drop in temperature seen after 1:30 minutes was caused by the introduction of the colder cement into the hole. Palavit® immediately developed heat and reached a maximum temperature of 50.0 °C ± 0.5 °C. Surgical Simplex® P initially gave off no heat, but reached an average peak value of 60.2 °C ± 0.7 °C ($P < 0.01$). The temperature that developed with Surgical Simplex® P cement was above 50 °C for 1:25 minutes, whereas Palavit® only reached this temperature for a few seconds.

The temperature peaked respectively after 5:30 ± 0:10 minutes and 5:40 ± 0:10 for Palavit® and Surgical Simplex® P . Within the same cement plug, the maximum range in peak time was 0:40 minutes for Surgical Simplex® P and 0:30 minutes for Palavit®.

Due to technical problems, there were two recordings—one for Surgical Simplex® P and one for Palavit®—where only five probes at the bone cement interface were activated.

The core temperatures were similar to interface temperatures for both cement types during the first 3 minutes. After this period, the core temperatures rose more pronouncedly, and reached peak temperatures of 69.5 °C ± 0.8 °C for Palavit® and 95.2 °C ± 2.2 °C for Surgical Simplex® P ($P < 0.01$). The room temperature during the experiments was 26.9 ± 0.3 °C. The radiographic examination did not reveal any misplaced thermocouple electrodes or cement penetration into bone. No air entrapment was observed radiographically or by macroscopic examination after dividing the cement plugs.

Discussion

The theory that bone necrosis is induced by thermal injury during polymerization of bone cement is based on previous experimental and clinical investigations. By histochemical studies, Lundskog (1972) found bone cell necrosis after exposure of 50 °C for 30 seconds. In accord with these findings, Eriksson and Albrektsson (1983) found that bone tissue heated to 50° for 1 minute will not remain as functioning bone, but will be resorbed and replaced by fat cells. Mjöberg (1986) has demonstrated a reduced tendency for hip prosthetic components to migrate, as measured by roentgen stereophotogrammetric analysis, when using less exothermic bone cement.

The maximum temperature of bone-acrylic cement interface is a function of many factors, but primarily of the amount of monomer that actually polymerizes, which is in turn dependent on the chemical formulation of the cement and the volume (or thickness) of the cement used (DiPisa et al. 1976). The content of chemical accelerator and deaccelerator systems added to the monomer or polymer has a marked impact on the rate of heat production from curing cement.

Heat generation from curing bone cement has in previous measurements caused bone-cement interface temperatures ranging from 37 °C to 120 °C (Sloff 1971, Meyer et al. 1973, Debrunner 1974, Reckling and Dillon 1977, Larsen and Ryd 1989). Previous studies have all shown large variations in temperatures measured. Due to large temperature gradients in the area, the exact location of the probe at the bone cement interface is crucial. However, because cement is pressurized into the bone tissue during insertion of prosthetic components (Noble and Swarts 1983; microinterlock), the location of the bone cement interface is no longer predictable, and thus temperature recording can be within the cement rather than at the bone-cement interface, and thereby causing inaccuracy in measurements. Our model was designed to prevent this problem, and we have not seen any microinterlock, as judged by radiography and inspection of divided cement plugs, in any of the experiments. Another explanation of the great variations in temperature measurements is the manifest difficulties in standardizing the thickness of the cement in the area of temperature measurements in the clinical situation. In the design of the present study, we tended to minimize this problem by using a standardized amount of cement with a well-defined shape and standardized pressure on the cement plug in each experiment. Local variations in temperature at the bone-cement interface due to differences in heat conductivity in mineralized bone and soft tissue surrounding trabecular bone is another parameter responsible for variations in the temperature measurements reported in the literature. To minimize this problem, we have used eight temperature probes at the bone-cement interface, and thereby have obtained a more reliable mean value with SEM values less than 0.7 °C.

Blood flow in bone tissue might cool curing cement in the in vivo situation. On the other hand, it has been demonstrated that blood flow generally ceased in a rabbit model upon the insertion of cement (Albrektsson and Linder 1984). The vascular standstill covered distances of 0.5 mm and 2 mm from the cement, which will markedly reduce the cooling effect of blood flow in the in vivo situation.

Another problem when measuring temperatures is the heat capacity of the probes, especially when using several probes. We have used size 0.01-mm electrodes; and although vulnerable, they are so thin that heat capacity is of no practical importance.

Palavit® in our study developed a significantly lower peak temperature than Surgical Simplex® P cement, which was probably due to the content of terpinolene. The temperature developed with Surgical Simplex® P cement is above the crucial temperature for bone tissue, i.e., 50 °C for 1:25 min. whereas Palavit® only reached this temperature for a few seconds. According to the above-cited literature, this difference in temperature is so pronounced that it presumably will influence the bone-repair process around the cement.

These in vitro studies have to be supported by in vivo evaluation of bone repair following cement application because other features, such as the cooling effect of blood circulation, and also acute chemical trauma of bone cement, might be altered when changing the chemical composition of the cement. A higher chemical toxicity of a new cement could eliminate the beneficial effect of a lower thermic toxicity. The conclusions of this experiment only account for the thermic impact of the two cement types. The model seems to provide reliable and continuous interface temperature recordings between bone and cement.

Acknowledgements

The thermal recording equipment and bone cement were kindly supplied by Essex Chemie A/G.

References

Albrektsson T, Linder L. Bone injury caused by curing bone cement. A vital microscopic study in the rabbit tibia. *Clin Orthop* 1984; 183: 280–7.

Debrunner H U. Die Erwarmung von Knochenzement bei der Polymerisation. *Arch Orthop Unfallchir* 1974; 78 (4): 309–18.

DiPisa J A, Sih G S, Berman A T. The temperature problem at the bone-acrylic cement interface of the total hip replacement. *Clin Orthop* 1976; 121: 95–8.

Eriksson R A, Albrektsson T. Temperature threshold levels for heat induced bone tissue injury: a vital microscopic study in the rabbit. *J Prostht Dent* 1983; 50: 101–7.

Larsen S T, Ryd L. Temperature elevation during knee arthroplasty. *Acta Orthop Scand* 1989; 60 (4): 439–42.

Lundskog J. Heat and bone tissue. An experimental investigation of the thermal properties of bone and threshold levels for thermal injury. *Scand J Plast Reconstr Surg* 1972; 9: 1–80.

Meyer P R Jr, Lautenschlager E P, Moore B K. On the setting properties of acrylic bone cement. *J Bone Joint Surg* (Am) 1973; 55 (1): 149–56.

Mjöberg B. Loosening of the cemented hip prosthesis. The importance of heat injury. Thesis. *Acta Orthop Scand* (Suppl 221) 1986: 1–40.

Noble P C, Swarts E. Penetration of acrylic bone cements into cancellous bone. *Acta Orthop Scand* 1983; 54 (4): 566–73.

Reckling F W, Dillon W L. The bone cement interface temperature during total joint replacement. *J Bone Joint Surg* (Am) 1977; 59 (1): 80–2.

Slooff T J. The influence of acrylic cement. *Acta Orthop Scand* 1971; 42 (6): 465–81.