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Conduction Analysis of Cement Interface Temperature in Total Knee Arthroplasty

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We applied an axisymmetric model of the tibia to a finite element method and analyzed the heat conduction from bone cement in total knee arthroplasty using numerical simulation with the finite element analysis software, ABACUS. We hypothesized the thermal necrotic map of bone. Moreover, we suggested a method for preventing thermal necrosis of bone using this simulation.

We adopted an initial temperature of 32ºC and a cement layer of 3mm to our simulation and analyzed heat conduction. The maximum temperature into the cement layer was 65ºC and 56ºC at the bone-cement interface 200 sec after the start of heat generation. Bone necrosis was observed approximately 2mm from the bone-cement interface.

To thin the cement layer, the maximum temperature and bone necrotic area reduced. At a cement layer of less than 1mm no bone necrotic area was observed. To lower the initial temperature of the bone surface, the maximum temperature and bone necrotic area was also reduced. At an initial temperature of less than 28ºC we did not find any bone necrosis. Even if we hypothesized pouring cold water around the prosthesis, the maximum temperature did not reduce at any time, neither did the necrotic area reduce.

We should make the cement layer as thin as possible and cool the bone before cementing to prevent thermal bone necrosis.

Bone cement is often used in orthopedic surgery, such as arthroplasty. However, in histology, bone necrosis is often found at the bone-cement interface [2,7,9,18,25]. Bone necrosis may cause loosening to the prosthesis of an arthroplasty [1,3]. Thermal injury by bone cementing is one of the most important causes of the bone necrosis [7,14,19,25].

We applied the finite element method, and analyzed heat conducted from the bone cement in total knee arthroplasty (TKA) using a numerical simulation. We hypothesized the thermal necrotic map of the bone. Moreover, we suggested a method for preventing thermal necrosis of the bone in TKA using this simulation.

MATERIALS AND METHODS

Conduction analysis

An axisymmetric finite element model of the tibia was used for the analysis (Fig. 1). We adapted various formulae for heat conduction and thermal properties (Table Ⅰ)[10,23] to the finite element analysis software, ABACUS (Hibbit, Karlsson & Sorensen) and analyzed the heat conduction from bone cement along the tibia component in TKA. The tibia component

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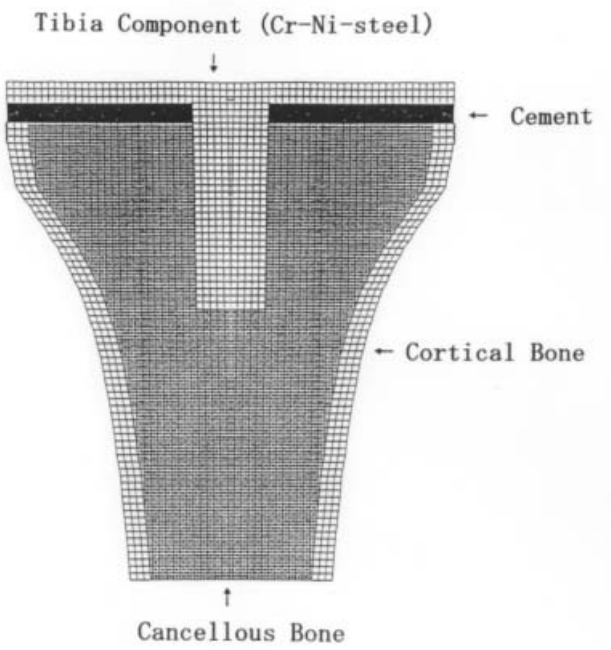

Fig. 1. Axisymmetric model of the tibia applied for analysis.

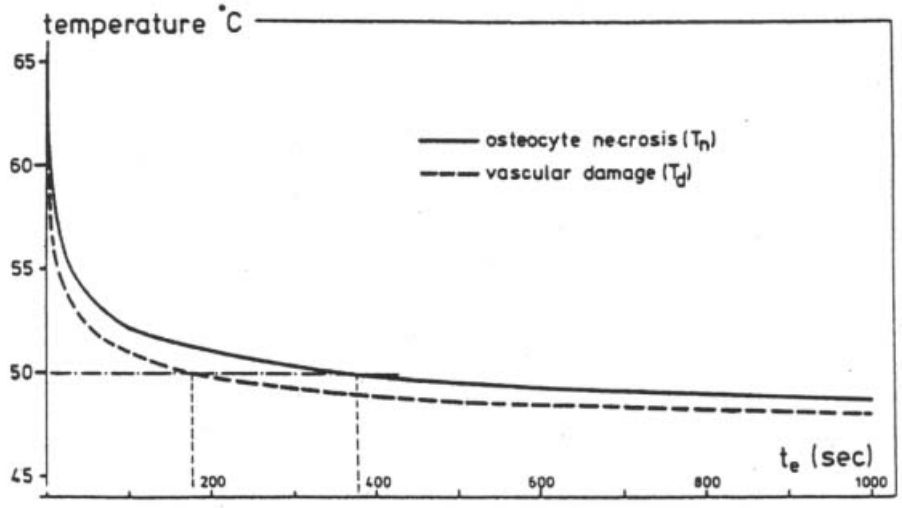

Fig. 2. Time-dependent threshold level for cell necrosis.
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was made of stainless-steel (Cr-Ni-steel) and its tibia plate was 3mm in width. Both initial and ambient temperature were assumed to be 32°C.

Details of the calculation method were described in our precious publication, and validity of the simulation has been confirmed in the animal experiment [8].

We measured the temperature of the bone surface in 10 TKA operations using a Digital Thermometer CTM-205 (TERUMO) and measured the width of the cement layers in 30 TKA X-rays.

Estimation of the thermal necrotic area

Temperature threshold levels for heat induced tissue injury is time-dependent. Moritz et al. reported the relationship between thermal injury for porcine and human skin and duration of time [20]. According to their data, Huiskes et al. plotted a temperature threshold level curve for bone necrosis (Fig. 2)[10].

In our study we contrived a damage function D(T) which showed the percentage of damage of a bone cell when that cell received heat, T°C, for one second. For example, a bone cell necrotizes, i.e., damage reaches 100% when it receives 50°C heat for 360 sec. Then, the amount of damage caused by one-second exposure to 50°C heat was caluculated and expressed as follows.

$$D(50) = 1/360 \times 100 = 0.28\ \%$$

Thus we found the relationship between temperature and percentage of bone necrosis (Fig. 3), and approximated this function to the formula using the least squares method.

$$D(T) = 0.0127272T^3 - 1.92419T^2 + 97.20029T - 1640.088$$

When we integrate D(T) to 100%, the bone is considered to be necrotic.
So, we supposed the function $F(\tau)$.

$$F(\tau) = \int_0^{\tau} D(T(t))dt$$

t: time after the start of heat generation
T(t): the variance for time t, at T°C

We calculated the area $F(\tau) \geqq 100$ in our simulation. This area represents the bone necrotic area.

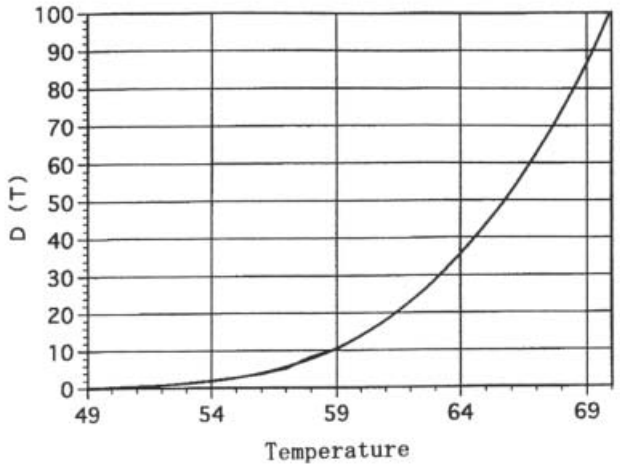

Fig. 3. Damage function.
The value of D(T) is the percentage of bone necrosis.

The prevention of thermal necrosis

We considered the possibility of thermal necrosis prevention by our simulation as follows:

1) Thin the cement layer
2) Lower the initial temperature of the bone surface
3) Pour cold water around the prosthesis

When we pour a large amount of cold water, the water temperature is considered to be the ambient temperature in our simulation. We adapted various ambient temperatures and various pouring times to our simulation.

RESULTS

Conduction analysis and thermal necrotic area

The mean temperature of the bone surfaces in 10 TKA operations was 32.1±1.0°C (30.4~33.4°C). The mean width of the cement layers in 30 TKA X-rays was 2.8±0.6mm.

We adopted an initial temperature of 32°C and cement layer of 3mm and analyzed the heat conduction (Fig. 4). The maximum temperature was 65°C into the cement layer and 56°C at the bone-cement interface 200 sec after the start of heat generation.

In this condition, we calculated the area $F(\tau) \geqq 100$. Bone necrosis was observed approximately 2 mm from the bone-cement interface (Fig. 5).

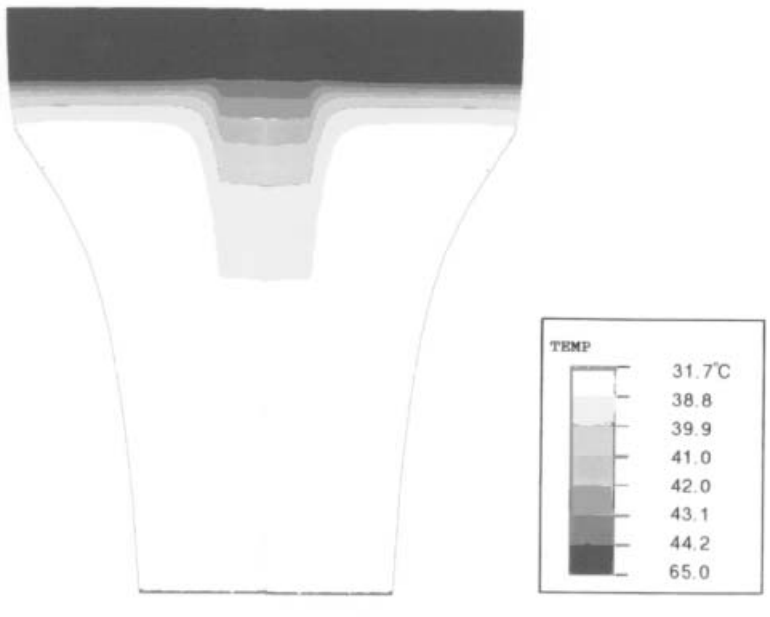

Fig. 4. Temperature profile from the simulation.
(initial temperature 32°C, cement layer 3 mm, 200 sec after the start of polymerization)

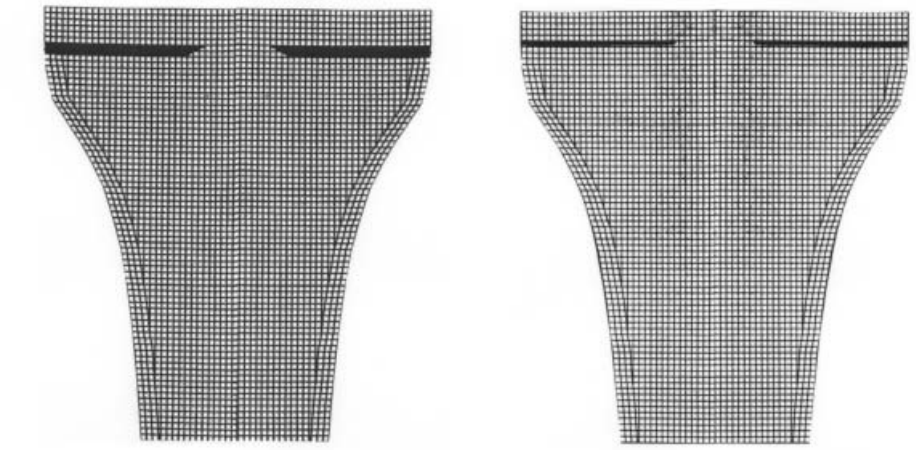

Fig. 5. (left) Necrosis map from the simulation. (initial temperature 32°C, cement layer 3 mm)
Fig. 6. (right) Necrosis map from the simulation. (initial temperature 32°C, cement layer 2 mm)

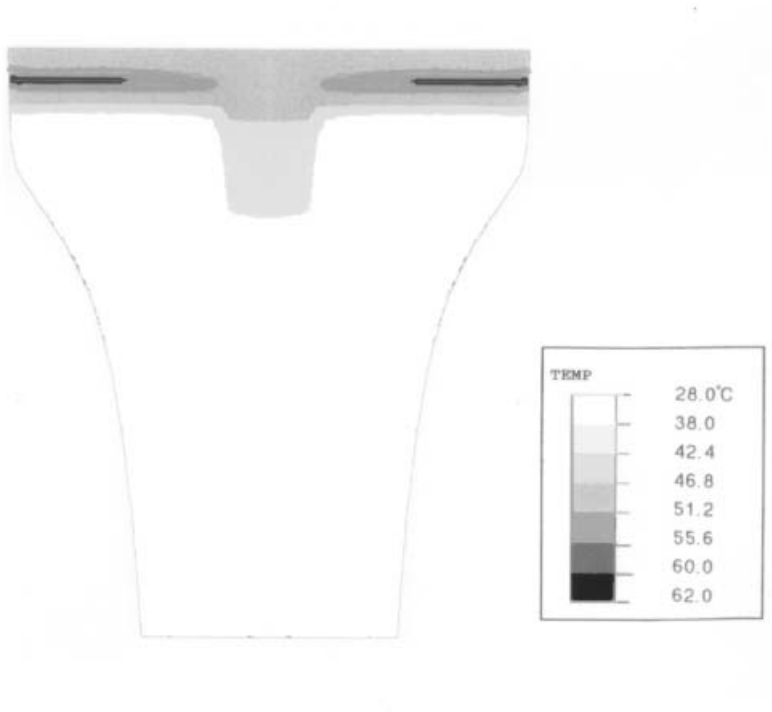

Fig. 7. Temperature profile from the simulation.
(initial temperature 28°C, cement layer 3 mm, 260 sec after the start of polymerization)

Table Ⅰ. Thermal properties of materials.

	Density [10^3 kg/m^3]	Thermal conductivity [J/ms °C]	Specific heat [10^3 J/kg °C]
Component	7.8	14	0.46
Cancellous bone	2.3	0.4	1.3
Cortical bone	2.1	0.38	1.26
Bone cement	1.19	0.17	1.6

The prevention of thermal necrosis

We changed various parameters of our simulation to prevent thermal bone necrosis.

1) Thin the cement layer by 1mm (initial temperature 32°C)
   a) Cement layer 2mm
      The maximum temperature was 56°C into the cement layer and 53°C at the bone-cement interface 200 sec after the start of heat generation. The bone necrotic area reduced to 1mm from the bone-cement interface (Fig. 6).
   b) Cement layer 1mm
      The maximum temperature was 46°C into the cement layer and 43°C at the bone-cement interface 210 sec after the start of heat generation. No bone necrotic area was observed at a cement layer of less than 1mm.
2) Lower the initial temperature of the bone surface by 1°C (cement layer 3mm)
   At 28°C the maximum temperature was 56°C into the cement layer and 51°C at the bone-cement interface 260 sec after the start of heat generation (Fig. 7). No bone necrotic area was observed at an initial temperature of less than 28°C.
3) Pour cold water around the prosthesis (cement layer 3mm)
   The maximum temperature both into the cement layer and at the bone-cement interface did not reduce at any pouring time with water at 10°C and 20°C (Tables Ⅱ, Ⅲ). Neither did the necrotic area reduce.

Table Ⅱ. Maximum temperature after pouring 20°C water around the prosthesis.

Time of pouring	Cement layer (°C)	Bone-cement interface (°C)
0 min after the start of polymerization	63.6	55.0
1 min	64.0	55.4
2 min	64.5	55.6
3 min	64.9	55.9
4 min	65.0	56.0
5 min	65.0	56.0

DISCUSSION

Bone cement has been used in orthopedic surgery such as arthroplasty. And various systemic or local complications of patients have been reported. Especially, one of the most clinically important complications is thermal necrosis around the cement layer. Some

Table Ⅲ. Maximum temperature after pouring 10°C water around the prosthesis.

Time of pouring	Cement layer (°C)	Bone-cement interface (°C)
0 min after the start of polymerization	63.0	55.7
1 min	63.6	55.1
2 min	64.2	55.5
3 min	64.8	55.9
4 min	65.0	56.0
5 min	65.0	56.0

reasons for thermal necrosis have been discussed in the literature, such as a local cytotoxic effect of the cement monomer, heat generation at polymerization, mechanical irritation of bone cells, etc. [1,3,7,9,14,18,19,25]. Among these, the effect of heat generation has been traditionally noted [7,14,19,25].

The effect of temperature on biological tissue was first reported by Cornheim in 1873, who observed tissue necrosis by supplying a temperature of 52°C for 6 or 7 minutes in rabbit ears [4].

The effect of temperature on bone tissue was first reported by Lundskog in 1972. This author reported that a temperature of 50°C induced cellular necrosis at an exposure time of 30 sec in rabbit bone [15]. Eriksson et al. showed that bone tissue heated to 50°C for 1 min, or 47°C for 5min, exhibited the finding of injury in Belgian hares and lop-eared rabbits [6]. This proved that thermal necrosis is time-dependent.

Bone cement generates heat during polymerization. Swenson et al. recorded a temperature of 124°C at the center of a sphere of cement [26]. Noble reported that the maximum temperature attained during polymerization ranged from 60°C to 70°C for 7 cements tested [21].

Homsy et al. recorded a peak temperature between 60°C and 70°C at the bone-cement interface in canine bone [9]. DiPisa et al. reported a peak temperature of 70.4°C at the bone-cement interface during surgical operation of total hip arthroplasty (THA) [5].

The simulation carried out in the present study showed a temperature of 65°C at the cement layer and 56°C at the bone-cement interface. These temperatures are thought to be sufficient for bone necrosis [6,15].

For the initial temperature of the bone surface, Larsen et al. recorded a mean of 32±2°C (27~35°C) [13], and Wykman recorded 33°C (31~36°C) in THA [28]. We recorded a similar mean temperature of 32.1±1.0°C (30.4~33.4°C) in TKA. Our present simulation shows that bone necrosis can be observed above 28°C initial temperature. So initial temperatures of Larsen, Wykman and ours are sufficiently high for bone necrosis.

There have been some numerical simulations of thermal bone necrosis. Swenson et al. showed that the area of bone necrosis was due to the width of the cement layer by a simulation with 5- and 10 mm width cement layer in THA [26]. Huiskes constructed a necrosis map of THA, which showed that the temperature of the cement layer around a cup was more than 100°C, and there was about a 2 mm width of bone necrosis area and about 2 mm width of vascular damage area external to the necrosis [10]. Mazzullo et al. constructed a THA model with a 3 mm wide cement layer around the stem and showed a 1.8 mm width of bone necrosis area [17]. However most simulation models have been for THA, and TKA models are rarely reported. Therefore we constructed a TKA model with 3 mm width cement

layer and showed a 2 mm wide area of bone necrosis. Generally, the cement layer of TKA is thinner than that of THA. However, even in TKA the possibility of bone necrosis by polymerization of the cement was suggested by our simulation. We reconfirmed that temperature is very important for bone necrosis.

Mjöberg summarized the methods to reduce heat injury to the bone during polymerization as follows [19]: (ⅰ) Reduce the amount of cement, (ⅱ) Use metal implants, (ⅲ) Precool the implant, the cement, or the bone, (ⅳ) Increase the heat capacity of the cement by adding heavy metals or melting crystalline monomers, (ⅴ) Reduce the polymerization rate, (ⅵ) Reduce the proportion of monomer by increasing the powder-to-liquid ratio, by adding aqueous gel, or adding granulated polymer.

Our simulation showed a 2, 1, 0 mm width of bone necrosis with a 3, 2, 1 mm thick cement layer in the TKA model. The thinner the cement layer, the smaller the bone necrotic area. Indeed, a thinner cement layer may be better for thermal bone necrosis, but correspondingly worse for the strength of the cement fixation. Huiskes et al. recommended a cement layer less than 2.5 mm width to prevent thermal necrosis [11]. However, Krause et al. reported that maximum strengths were achieved for human cancellous bone by facilitating penetration of the cement for distances of 5 to 10 mm into the bone [12].

Some reports have been published that water irrigation diminishes temperature elevation. Wykman reported that the median maximum temperature during cement curing with no Ringer irrigation was 48.8ºC (40.5~67.3ºC) and with continuous fluid irrigation was 40.9ºC (36.5~47.5ºC) [27,28]. In our simulation however, we could not find any relationship between water irrigation and temperature elevation. On pouring 10ºC cold water, the maximum temperature at the cement layer reduced from 65ºC to only 63ºC and at the bone-cement interface, the temperature reduced from 56ºC to 54ºC. Moreover the necrotic area did not reduce (initial temperature 32ºC, cement layer 3mm).

Some authors recommend the precooling of prosthesis [5,10,16,22]. DiPisa et al. reported that by precooling the acetabular component of THA to –84ºC before inserting it into the cement, the temperature during polymerization at the bone-cement interface could be reduced from 70ºC to 49ºC [5]. In our simulation, we found that precooling to less than 28ºC of bone surface prevents bone necrosis. Precooling of bone surface to this temperature can be accomplished by the use of the cooled saline before the cementing procedure.

Presently, bone cement consists mainly of PMMA (polymethylmethacrylate), however some new cements are being developed. Yamamoto et al. have been clinically testing bioactive bone cement which consists of CaO-SiO_2-P_2O_5-MgO-CaF_2, and that exhibits a much lower curing temperature than PMMA cement. The highest surface temperature for PMMA cement composite was 60ºC, while that for bioactive bone cement was 35ºC [29]. Peter et al. investigated biodegradable bone cement which consisted of PPF (polypropylene fumarate). Its maximum crosslinking temperature was between 38ºC and 48ºC, which was much lower than the crosslinking temperature of 94ºC for PMMA cement [24]. We expect that these new cements will have practical use in preventing thermal bone necrosis as a result of polymerization.

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