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Modelling of heat transfer in biomechanics – a review Part II. Orthopaedics

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The aim of this paper consisting of three parts is to review available results pertaining to various heat transfer problems of biomechanics. The second part covers thermal problems specific to orthopaedics. Three classes of problems are investigated: exothermal bone cement polymerisation *in situ*, frictional heat generation during articulation of joint implants and frictional heat generation during bone cutting and drilling. The existing results pertaining to modelling and experimental measurements are reviewed. Thermal damage criterion is discussed and various possible means of minimising injuries to tissues are discussed.

The first class of problems studied includes also our own results.

Key words: heat transfer, bone cement polymerisation, frictional heat generation, articulation of joint implants

1. Introduction

In the first part of our paper, the issues related to the modelling of heat transfer in perfused tissues and the criteria of thermal damage to tissues have been examined. The foundations of the most popular bio-heat transfer equations have been discussed along with their inherent limitations. The results obtained with the use of these models have been compared with the selected experimental data. Also the models of thermal damage available in literature have been given, with a brief review of the experimental methods of identification of their parameters.

In the present, second, part thermal problems specific to orthopeadics are rewieved. They can be distinguished as the heat transfer phenomena occurring during and after orthopaedic operations in the direct vicinity of the operation site. One can divide them into three distinct categories: issues concerning heating up of the limb–implant system by the latent heat of the bone cement polymerisation (in cemented prostheses), problems of frictional heating occurring during normal functioning of the joint prosthesis and issues concerning rapid heating up of the bone due to sawing and drilling during orthopaedic operations. The characteristics of those three kinds of problems are outlined in table 1.

Table 1. Basic types of heat transfer related to joint replacements

	Bone cement polymerisation	Bone drilling and sawing	Frictional heating of the implant
Range of application	all cemented implants, hip endoprostheses (acetabular and femoral parts)	joint replacements of any kind; fixation of screws in bone	joint replacements of any kind
Time and duration of exposure to elevated temperatures	during and immediately after implantation, duration: several minutes	prior to implantation, duration: few minutes	during joint operation, temperature cycles repeated many times during the service life of artificial joint
Location	bone tissue–bone cement interface, bone cement domain	cut surface/drilled hole	articulating surfaces, interface (acetabular cup and prosthesis stem in hip implants)
Other physical phenomena affected	bone cement polymerisation (quantity of toxic monomer leftover), thermal bone necrosis, aseptic loosening of the prosthesis	thermal bone necrosis, surgical trauma	acetabular cup wear and material degradation
Materials	metals, bone cement mixtures (based on PMMA), bone and surrounding tissue	bone, metals	UHMWPE, ceramics metal alloys, bone and surrounding tissue
Max. temperatures	up to approx. 60 °C at the bone surface and over 100 °C within cement domain (dependent on the type of cement)	over 150 °C (sawing without coolant)	around 43 °C in the middle of the prosthesis head and less than 40 °C on the surface of bone

Another class of problems dealing with the influence of high temperature on hard tissue, not included in table 1, is related to various hyperthermia treatments of bony sarcoma or adjacent soft tissues. Laser techniques used in periodontology, in the treatment of oral diseases, are a good example, cf. [41]. The freezing and lyophilisation of the bone fragments before the storage and implantation in the case of heterogenous bone implants is not covered by the present part of our paper either. These techniques are described in the third part, which is concerned entirely with the issues of cryogenics.

From the physical point of view, the solution of the problems characterised in table 1 consists in finding the usually transient temperature distribution in highly heterogeneous tissue domain subject to heat fluxes of various origin. Formulations of the bioheat transfer equation presented in part I of our paper can be used to predict heat dissipation in the soft tissue surrounding the bone, joint and prosthesis. However, while the perfusion phenomena were of central importance for the modelling of soft tissue heat transfer, in the case of the orthopaedics several other phenomena – specific to the orthopaedics only – also need to be taken into consideration.

2. Heat generation during bone cement polymerisation: hip endoprosthesis

The physical situation is sketched in figure 1 (intramedullary component of cemented hip prosthesis) and in figure 2 (cemented acetabular part). The stem of the endoprosthesis is fixed in the medullar cavity of the femur with the aid of polymethylomethacrylate bone cement (PMMA) which is inserted by the surgeon in a "doughty" state and then polymerises *in situ*. The same pertains to the acetabular cup. Cemented implants are also used in the case of the knee joint reconstruction, cf. [33], [34]. The polymethylomethacrylate bone cement is also often used for filling the voids in bone and for fixation of other joints than hip joints, see [27] for an example of use of PMMA in stabilisation of fossa component of temporomandibular joint prosthesis – the proximity of brain and the dura mater to the site of the polymerisation is an additional unfavourable condition.

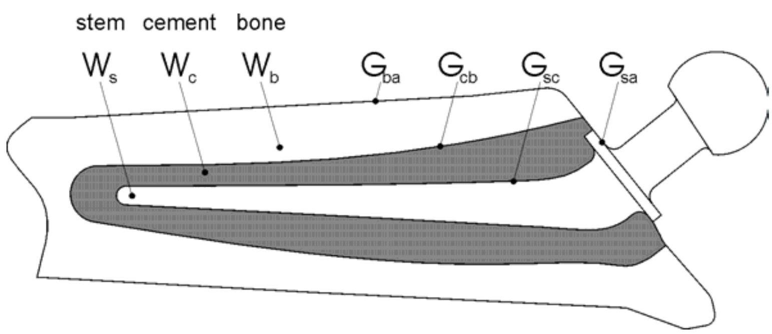

Fig. 1. Intramedullary cemented part of femoral prosthesis, the respective domains are labelled with Ω and domain boundaries with Γ supplemented with obvious subscripts

When modelling the heat transfer in the cemented bone–implant system the following simplifications are usually imposed:

1. Axial symmetry of the model along the axis of the stem, see [26], [42], [43]. In [20], a two-dimensional, axisymmetric model was considered.

2. Uniform heat generation by the bone cement with constant power, non-zero only in a specified period of time.

3. Uniform heat generation by the bone cement with varying power in a specified period of time, see [20] and [43].

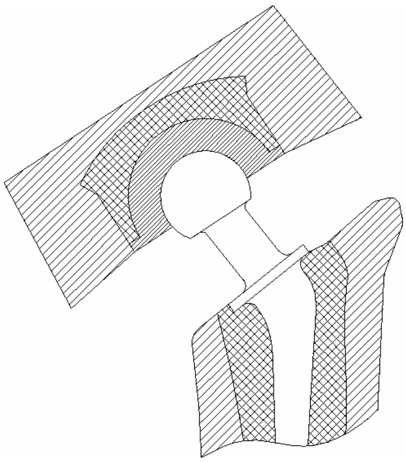

Fig. 2. Cemented acetabular part of femoral prosthesis

4. Perfect thermal contact between the materials [43]. In [26], several areas of null contact are considered on an otherwise perfect thermal interface.

5. Temperature-independent material moduli, see [20], [26], [42], [43].

6. Material isotropy [20], [26], [42], [43]. This assumption can be easily generalised, but the lack of reliable material coefficients is the major obstacle. There are
a few experimental measurements of anisotropic thermal properties of the tissue and these usually do not encompass all the components of the thermal conductivity (or diffusivity) tensor, cf. [11] and [30].

7. Constant boundary conditions independent of temperature [20], [26], [42], [43]. The temperature in the biological system is thus assumed to vary over a range small enough to allow the changes in the boundary convection to be disregarded. Otherwise the convection film coefficient should be reiterated for every computed boundary temperature. In view of the relatively small temperature variation, ommitting this fact should be regarded as a reasonable assumption.

8. No influence of the surrounding muscle tissue, see [20], [26], [42], [43].

These simplifications originate mainly from the lack of knowledge that would be required to deal with more sophisticated models. For example, the thermal conductivity of the bone tissue, as reported in numerous papers on the subject (see e.g. [5], [8], [20], [26]), varies from 0.26 to 0.60 [W/(mK)] and the specific heat varies from 1150 to 2370 [J/(kgK)], etc. Generally, the material parameters of bone tissue depend on multiple factors, mainly the bone composition (cortical/spongeous) and bone marrow and water content. These depend on the location in the bone investigated and a person examined.

One way to deal with this difficulty is to construct mathematically tractable model of heat transfer in bone tissue on the basis of anatomical observation, see [16] for the description of bone structure. While being more realistic this approach presents considerable difficulties arising from considerable scatter of measurements of bone properties and the fact that the very architecture of the bone changes with time [16].

Another possible way of approaching the problem is to use experimental results to obtain the distribution of values of average material properties needed to solve the problem within the framework of the classical heat conduction model, cf. [5]. While less prospective, this approach allows immediate construction of simple theoretical models. Such an approach was used in [20], [26], [42], [43].

Another issue is the model of cement polymerisation. At least three different approaches can be envisaged. The first approach is to refrain from modelling the process of polymerisation and to assume a constant power produced, within the cement domain, during the specified period of time. The advantage of this idealisation is that only the linear equation of heat conduction with constant source term has to be considered. The drawback is that the model is rather oversimplified in comparison with real situation. Furthermore, one can gain no information about the monomer leftover, within the bone, as no calculation of polymerisation is performed. Yet a more simplified version of this approach is to assume that the cement polymerisation is instantaneous and the cement mass heats up homogeneously by a certain temperature increase that can be obtained from the heat balance equation, namely:

$$\varDelta T_c = \frac{q}{c_c}, \tag{1}$$

where q denotes the overall heat released during the polymerisation of unit mass of cement [J/kg] and c_c is its specific heat [J/(kgK)]. Then, the following initial condition is imposed:

$$T(\mathbf{x}) = \begin{cases} T_0 & \text{if } \mathbf{x} \in \varOmega_s\,, \\ T_{\text{body}} & \text{if } \mathbf{x} \in \varOmega_b\,, \\ T_0 + \Delta T_c & \text{if } \mathbf{x} \in \varOmega_c\,, \end{cases} \tag{2}$$

where T_0 denotes the ambient (room) temperature, T_{body} is the normal body temperature and ΔT_c is obtained from equation (1). The domains considered are denoted by $\varOmega$ supplemented with the following subscripts: s for the prosthesis stem, b for the bone and c for the cement. Then the transient equation of heat conduction is solved to obtain the maximum temperature rise on the bone domain boundary. Such a model of the acetabular implant was constructed and solved by JEFFERISS et al. [21] for the one-dimensional case. Figure 3 presents the results. Discontinuous initial condition (2) is distinguished by the thick line. The problem of PMMA polymerisation is reduced merely to the choice of initial temperature of cement mass. As can be inferred from figure 3 the initial temperature of 70 °C produces temperatures of the order of 50 °C at the bone–cement interface in the first stage of the process of heat dissipation. JEFFERIS et al. [21] argue that the instantaneous polymerisation model will always yield higher tissue temperature than any other model, where the energy is released over the finite time period. Therefore the data presented in [21] are supposed always to reflect the "worst case" in terms of tissue damage. It should be observed, however, that this would only be the case if the tissue-damage criteria were based on the temperature only, not on the temperature history. The experimental data presented in part I of the present paper suggest that preheating lowers the ability of the tissue to withstand elevated temperatures. Therefore such simplified model of heat release during polymerisation might not be satisfactory.

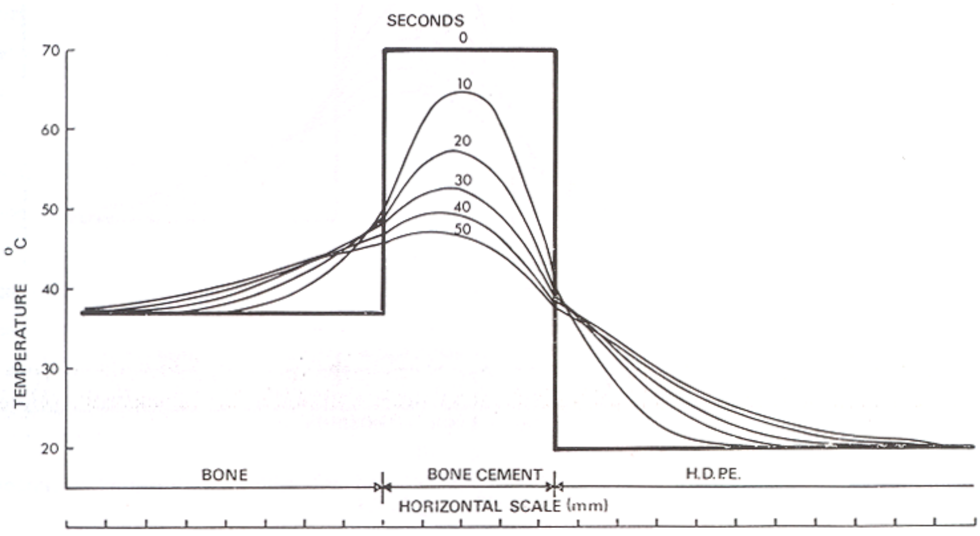

Fig. 3. Temperature field relaxation in one-dimensional model of cemented implant with instantaneous polymerisation. Thick line denotes discontinuous initial condition, after [21]; H.D.P.E. = high density polyethylene

The second approach is to assume uniform heat generation within the cement domain, with the rate varying with time. Such a procedure also gives no information about the distribution of final monomer leftover, but an attempt is undertaken to make the model to resemble the real situation more closely. The properties of cement such as the retardation time can be modelled here. Still only linear equation of heat conduction needs to be solved. Such an approach was followed by HUISKES [20] and by SWENSON et al. [43].

The third approach is to model the polymerisation process of bone cement by means of an appropriate kinetic equation. The polymerisation of monomethylometacrylate is a free-radical polymerisation [15], [32], so it is the first-order reaction with respect to monomer. The rate of polymerisation is linearly proportional to the concentration of monomer [32]. Additional effect that should be taken into account in constructing the kinetic equation is the glass transition of the cement. The kinetic equation is coupled with the heat conduction equation via the heat generation term, which makes the problem non-linear. More detailed description of this model will now be presented. Such an approach was primarily proposed by MAZZULLO et al. in [26] and has been developed by the present authors.

The available models exclude surrounding soft tissues from considerations. Extending them to cover the heat exchange in the muscles would be an important step in constructing the mathematical model of heat exchange in human limb or extremity, cf. [39], [52], [55].

In section 2.1, the model of heat transfer during cemented prosthesis fixation is presented and simplifications commonly made are discussed.

2.1. Formulation of the polymerisation problem

The temperature throughout the considered domain Ω is an unknown function. In the models that exclude surrounding soft tissue, Ω is essentially reduced to the sum of geometrical domains of the prosthesis stem, cement and bone domains, when intramedullar part is considered. Similarly, when the acetabular part is studied this domain consists of the prosthesis head, acetabular cup, cement and pelvis fragment domains.

The starting point in construction of a mathematical model is the energy balance equation, cf. [6], [51]

$$\rho\, c \frac{\partial T}{\partial \tau} + \nabla \mathbf{q} = q_v\,, \tag{3}$$

where τ denotes time, $\mathbf{q}$ is the heat flux vector, q_v is the volumetric heat production rate and the remaining symbols have been defined in the previous section. Taking into account the constitutive equation (Fourier's law):

$$\mathbf{q} = -\lambda \nabla T \tag{4}$$

we obtain the Fourier–Kirchhoff equation:

$$\rho\, c \frac{\partial T}{\partial \tau} = \nabla \cdot \left(\lambda \nabla T \right) + q_v\,. \tag{5}$$

The parabolic model of the heat conduction (Fourier's law) has been chosen here because the considered physical problem is assumed not to justify introduction of more sophisticated models. Hyperbolic models may be considered (e.g. the Vernotte law) when the rate of heat flux is very high which is not the case here.

Equation (5) is satisfied separately in stem, cement and bone domains (Ω_s, Ω_c and Ω_b):

$$\begin{aligned}
\rho_s c_s \frac{\partial T}{\partial \tau} &= \nabla \cdot (\lambda_s \nabla T), && \mathbf{x} \in \Omega_s, \\
\rho_b c_b \frac{\partial T}{\partial \tau} &= \nabla \cdot (\lambda_b \nabla T), && \mathbf{x} \in \Omega_b, \\
\rho_c c_c \frac{\partial T}{\partial \tau} &= \nabla \cdot (\lambda_c \nabla T) + q_v(T, w), && \mathbf{x} \in \Omega_c.
\end{aligned} \tag{6}$$

As we already know, the subscripts s, b and c label the prosthesis stem, bone and cement, respectively. In addition, the boundary conditions need to be imposed. The boundary of the considered domain is denoted by $\Gamma = \partial(\overline{\Omega}_s \cup \overline{\Omega}_c \cup \overline{\Omega}_b)$, where the bar denotes the closure of the domain. Only the Fourier boundary condition (also called the Newton boundary condition) is considered:

$$\lambda \frac{\partial T}{\partial \mathbf{n}} = \alpha (T_\infty - T), \quad \mathbf{x} \in \Gamma. \tag{7}$$

Here T_∞ and α are the functions prescribed on the boundary Γ, representing ambient temperature and convection film coefficient, respectively; $\mathbf{n}$ denotes the outward unit normal to Γ.

To complete the formulation of the problem, thermal contact conditions have to be specified. These are the continuity of the heat fluxes across the contact surfaces. The temperature is discontinuous across these surfaces with the jump proportional to the normal heat flux across the interface. The interfaces are denoted by $\Gamma_{bc} = \overline{\Omega}_b \cap \overline{\Omega}_c$ (the interface between the bone and cement domains) and $\Gamma_{cs} = \overline{\Omega}_c \cap \overline{\Omega}_s$ (the interface between the cement and stem domains). The constitutive equation on the bone–cement interface is given by:

$$\lambda_b \frac{\partial T_b}{\partial \mathbf{n}_b} = -\lambda_c \frac{\partial T_c}{\partial \mathbf{n}_c} = \beta_{bc} [[T]], \quad \mathbf{x} \in \Gamma_{bc}. \tag{8}$$

As usual, $[[f]]$ denotes the jump of f. Assuming perfect contact between the bone and the cement means that $\beta_{bc} \to \infty$ in this notation, i.e. that the temperature jump across the interface must be equal to zero when contact is perfect and nonzero heat flux across the interface can be maintained.

In table 2, the values of the material coefficients collected from the literature ([3], [5], [8], [20], [26], [43] and [54]) are provided. It should be noted that these values correspond to a certain type of prosthesis considered here (metal stem, polyethylene distal plug, cf. [26]). As is well known there is a wide variety of materials used for prosthetic heads, acetabula, stems, etc.

Table 2. Values of material properties relevant to modelling the heat transfer during PMMA polymerisation

	Prosthesis stem (metal)	Femur (bone)	Bone cement	Polyethylene
Thermal conductivity λ [W/(mK)]	14	0.26–0.60	0.17–0.21	0.29–0.45
Specific heat c [J/(kgK)]	460	1260–2370	1460–1700	2220
Density ρ [kg/m^3]	7800	1000–2900	1100	960

Table 3. Values of interface conductivities/convection film coefficients, the values are in [W/(m^{2}K)], after [20]

	Cement	Ambient
Metal stem	1 000–10 000	50–100
Bone	100–1 000	500–10 000

In table 3, possible values of interface conductivities are provided after [20] and [26]. Note that ambient temperatures corresponding to different interfaces may be different (the ambient–bone interface means the muscle–bone system, whereas the ambient–stem interface is the air–stem one). The values for thermal contact conductivity between polyethylene acetabular cup and cement and articulating surface of prosthesis head are estimated by HUISKES in [20] to be of the order of 500 [W/(m^{2}K)]. One may suspect higher values at the latter interface due to the presence of the synovial fluid. In [42], different values of the thermal contact conductivity for all interfaces were used, namely in the range of 10^2–10^6 [W/(m^{2}K)] and the study performed revealed no significant change in the process when the values exceeding 2000 [W/(m^{2}K)] were used. So, according to the estimations provided in table 3, the determination of the values of the thermal contact conductivity of the bone–cement interface may prove important, see [14]. Since experimental data on this topic are scarce (see [14] for a review of the recent developments) such research would be very helpful. Currently, we are planning to make measurements of the thermal contact conductivity of the interface *in vitro* in order to estimate the magnitude of temperature jump across the interface in the conditions experienced *in vivo*.

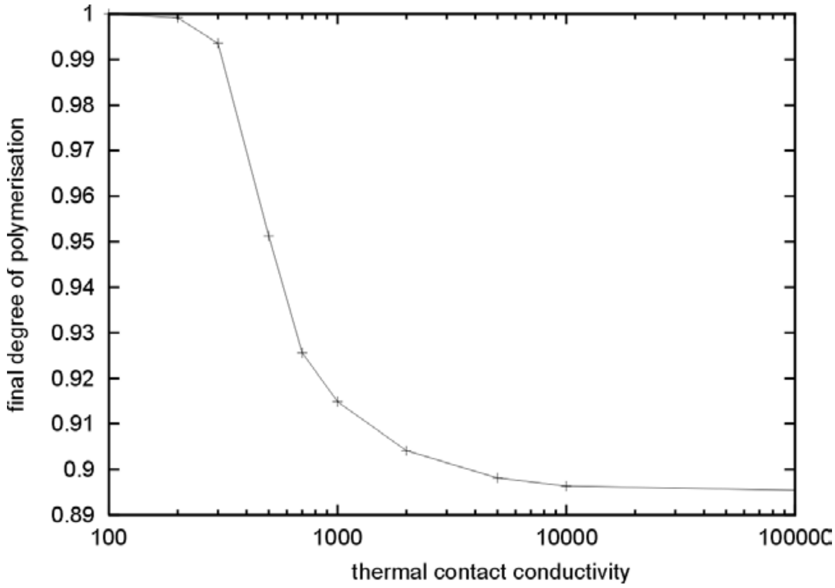

Fig. 4. Volume-averaged final degree of polymerisation versus thermal contact conductivity [W/(m^{2}K)] taken uniform and the same for all the interfaces, after [42]

The values of the interface conductivities have a significant influence on both the temperature distribution and the monomer leftover. In figure 4, the simulated relationship between the thermal conductivity and the overall volume-averaged final degree of polymerisation is given. Details of the numerical simulations are given in [42].

One lacking parameter in the set of equations (6) is the heat generation term in the domain Ω_c denoted by q_v . This term is directly proportional to the rate of polymerisation, which in turn is modelled via equation of polymerisation kinetics, cf. MAZULLO et al. [26],

$$q_v = Q \frac{dw}{d\tau}, \tag{9}$$

$$\frac{dw}{d\tau} = a \exp\left(-\frac{E_a}{BT}\right) P(T, w). \tag{10}$$

Here E_a, α are the experimental parameters, B is the universal gas constant and the function $P(T, w)$ is a factor that allows us to take into account the glass transition of the cement (phase change):

$$P(T,w)=\begin{cases}\dfrac{\alpha}{w^*(T)}w^{1-1/\alpha}\left(w^*(T)-w\right)^{1+1/\alpha}, & \text{if } w<w^*(T),\\ 0, & \text{if } w\geq w^*(T),\end{cases} \tag{11}$$

where α is an empirical parameter and $w^*(T)$ is the equilibrium degree of polymerisation at given temperature:

$$w^*(T)=\begin{cases}\dfrac{T}{T_g}, & \text{if } T<T_g,\\ 1, & \text{if } T\geq T_g.\end{cases} \tag{12}$$

Here T_g denotes the glass transition temperature of the cement. MAZZULLO et al. [26] completed their model by supplying the required parameters for the commercial cement (Howmedica SIMPLEX P), which are listed in table 4.

Table 4. Polymerisation constants used in equations (3.8)–(3.10), after MAZZULLO et al. [26]

Q [kJ/kg]	a [1/s]	E_a [J/mol]	α	T_g [K]	B [J/(mol K)]
193.0	2.6397×10^8	62 866.0	9.2	378.0	8.3143

The dependence $dw/d\tau = f(w, T)$ is shown in figure 5, where $f(w, T)$ is specified by the right-hand side of equation (10).

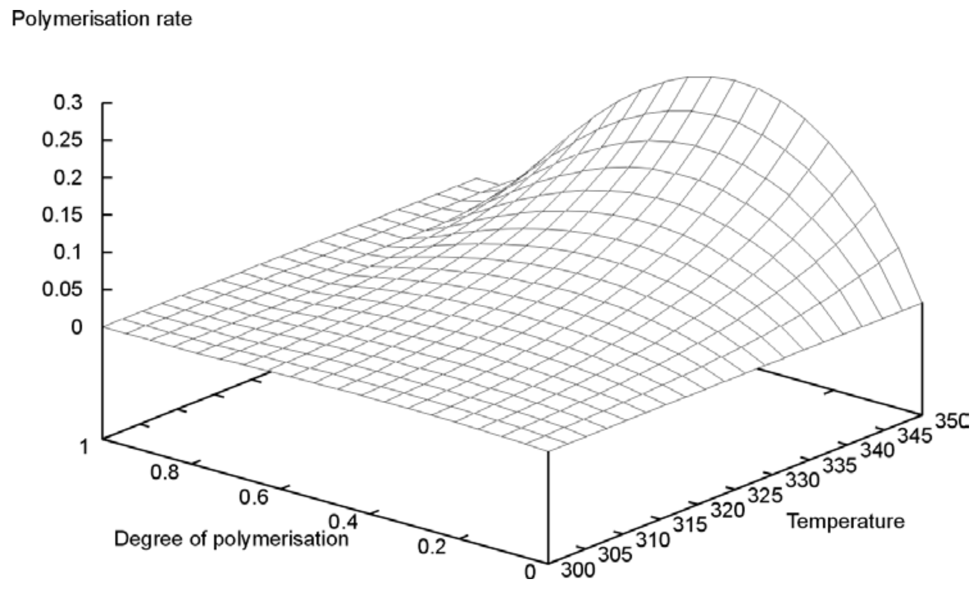

Fig. 5. Polymerisation rate model $dw/d\tau = f(w, T)$

This model allows one to obtain the polymerisation rate as a function of two parameters: the temperature and instantaneous degree of polymerisation (monomer conversion). The polymerisation rate-ratio curves calculated for different temperatures are shown in figure 6. These curves can be thought of as rates of isothermal polymerisation of cement for different temperatures. It should be noted, however, that given the same initial conditions the rate of isothermal polymerisation is always lower than that which would occur in any real situation, unless some enormously effective cooling measures are undertaken. Also the monomer leftover would be the *highest* in isothermal conditions. These conclusions are a straightforward consequence of the polymerisation kinetics equations (10)–(12).

The other extremal variant with the fastest possible polymerisation and the lowest monomer leftover is the model of adiabatic polymerisation. The polymerisation faster than adiabatic would require external heating-up of the cement (which is not considered here). Such a model is constructed by considering a simple point-mass described by two variables: the degree of polymerisation w and the temperature T. The heat exchange is null. The model consists of simplified equations for the temperature and the polymerisation:

$$
\begin{aligned}
c\frac{dT}{d\tau} &= Q\frac{dw}{d\tau}, \\
\frac{dw}{d\tau} &= a\exp\left(-\frac{E_a}{BT}\right)P(T,w).
\end{aligned}
\tag{13}
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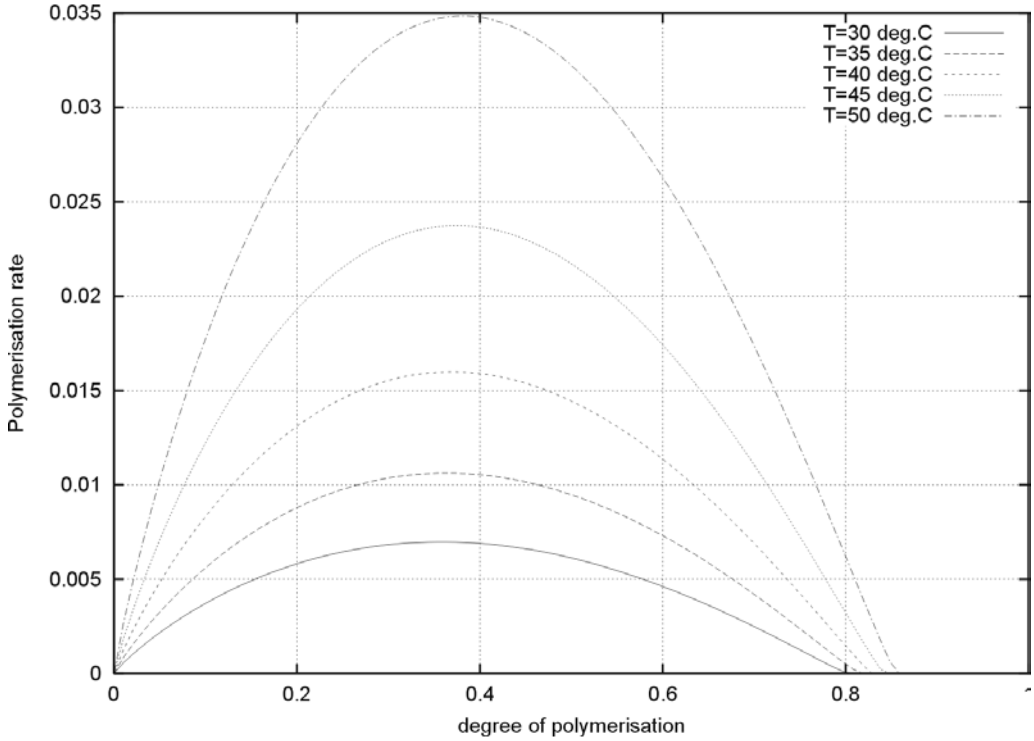

Fig. 6. Polymerisation rate versus polymerisation ratio
for different temperatures (isothermal process)

Solving equation (13) leads to the linear dependence of temperature on the degree of polymerisation *w*:

$$T = \frac{Q}{c}\left(w - w_0\right) + T_0 , \tag{14}$$

where T_0 is the initial temperature ($T_0 = T\ (\tau = 0)$), and the ordinary differential equation for the degree of polymerisation, provided that $w < w^*(T)$,

$$\frac{dw}{d\tau} = a \exp\left(- \frac{E_a}{B\left(\frac{Q}{c}\left(w - w_0\right) + T_0 \right)} \right)$$
$$\cdot \left(\frac{\alpha\, cT_g}{Q\left(w - w_0\right) + cT_0} \right) w^{(\alpha-1)/\alpha} \left(\frac{Q\left(w - w_0\right) + c\left(T_0 - wT_g\right)}{cT_g} \right)^{(\alpha+1)/\alpha} . \tag{15}$$

The numerical solution of the last equation is depicted in figure 7.

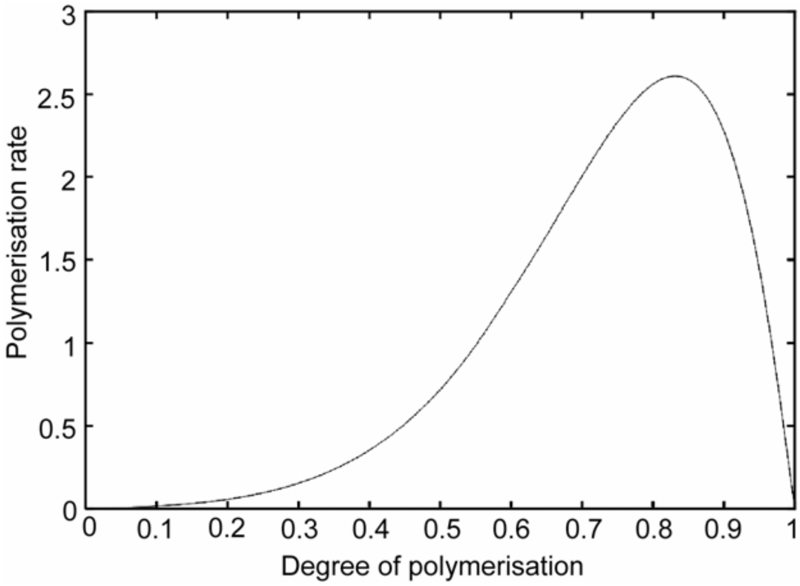

Fig. 7. Polymerisation rate versus polymerisation ratio (adiabatic process)

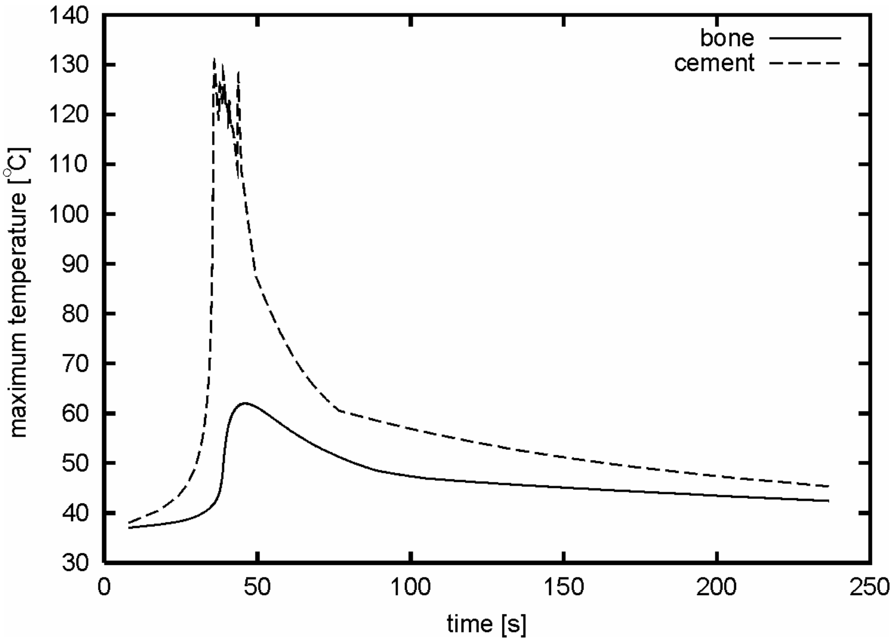

Fig. 8. Maximum temperatures in bone and cement domain, after [42]

Particularly, for the adiabatic polymerisation to be complete (w = 1) it is sufficient that initial condition satisfies:

$$Q(1-w_0)-c(T_g-T_0)=0\,. \tag{16}$$

If w_0 is higher than specified by equation (16) or T_0 is lower, the polymerisation will be incomplete.

In physical situations, heat exchange occurs and temperature gradients exist. The polymerisation time, the monomer leftover and attained temperatures will be therefore between those characteristic of the adiabatic and the isothermal processes.

The complete model of polymerisation allows us to find computational solutions of both coupled scalar fields of temperature and polymerisation ratio. The specific solution, namely maximum temperatures at appropriate domains plotted versus time, is depicted in figure 8.

It should be stressed that the two coupled field equations (5) and (10) are of different type, namely the Fourier–Kirchhoff equation is a partial differential equation for the unknown temperature T, whereas the polymerisation kinetics equation is an ordinary differential equation with respect to the function w. This difference stems from the fact that monomer diffusion phenomenon was tacitly omitted.

2.2. Exothermic polymerisation *in situ:* possible consequences and remedies

The ultimate goal of calculation of temperature/monomer leftover distribution throughout the implant is to acquire knowledge if the chosen prosthesis fixation technique presents considerable danger to the living tissue and to assess the probability of possible loosening of the prosthesis as an effect of resulting damage. To this end one needs criteria of thermal/chemical damage to the bone tissue (bone necrosis) that can be compared with computed values of the temperature and monomer leftover.

In general, the use of polymethylomethacrylate bone cement in the orthopaedic surgery may lead to threats. Theses can be classified as follows, cf. [17], [21],

1. Vascular disturbance at the site of implantation.
2. Disruption of the cortical and marrow circulation.
3. Temperature effects during polymerisation.
4. Chemical effects during and after polymerisation.

Only point 3 is considered here, although the formulations presented encompass also the phenomena of incomplete polymerisation and allow us, in principle, to predict chemical damage to tissue. For the experimental investigation of chemical trauma the reader is referred to LINDER [23], where cement dough was implanted into rabbit tibia and care was taken to minimalise the mechanical and thermal injury. The reported chemical trauma was very limited, giving rise to speculation that the primary injury mechanism during implantation is of mechanical nature, i.e. vascular injury, blocking of the Haversian canals by cement or intramedullary fat particles during drilling, etc.

The issue of thermal bone tissue necrosis has been investigated by many researchers. The results obtained suggest two possible basic mechanisms. One is the collagen protein denaturation. According to SWENSON et al. [43], it takes place at a temperature range of 56–70 °C. The second mechanism is caused by cellular death, which occurs at lower temperatures and is therefore more important. The results presented in [2], [3], [4], [20], [26], [43] and [45] point out the time–temperature dependence inherent in thermal necrosis criteria. For example, the temperature of 70 °C is believed to kill cells instantly, the temperature of 50 °C needs to be maintained for 30 seconds and that of 45 °C – for 5 hours. Higher temperatures (of the order of 70 °C) are needed to destroy the regenerative capacity of the bone tissue, cf. [20], [43].

From the available data the straightforward mathematical criteria based on the "additivity rule" have been constructed by MAZZULLO et al. in [26]. Let us assume that the time necessary to cause thermal bone necrosis at a given temperature T is

$$\tau_c = M \exp\left(\frac{\mu}{B(T - T_{\mathrm{ref}})}\right), \quad T > T_{\mathrm{ref}} . \tag{17}$$

Obviously, as $T \rightarrow T_{\mathrm{ref}}$ the time till necrosis τ_c becomes infinite and below the reference temperature T_{ref} no thermal damage processes take place. M and μ are model constants. Under given non-isothermal conditions the local measure of thermal bone tissue damage η can be constructed as an integral of fractions of exposure time at given temperatures over time

$$\eta(\mathbf{x}, \tau) = \int_0^{\tau} \frac{dt}{\tau_c(T(\mathbf{x}, \tau))} . \tag{18}$$

The values of η equal to or in excess of unity indicate local bone necrosis. This criterion is analogous to the Palmgren–Miner hypothesis of linear damage accumulation in fatigue mechanics and has similar drawbacks. For example, it does not take into account the succession of different stages of thermal load – the heating periods of various intensity will produce the same damage irrespective of their relative order.

The necessary constants for the model were obtained by MAZZULLO et al. in [26] by means of linear regression analysis and are given in table 5.

Table 5. Experimental constants for thermal bone necrosis criteria appearing in equation (3.15), after MAZZULLO et al. [26]

M [s]	μ [J/mol]	T_{ref} [K]
1/27.4	1000.0	310.0

While it seems to be a good measure for a single heat shock cellular damage, this criterion may be insufficient when dealing with repeated thermal bone loading as the living tissue can probably adapt to higher temperature by means of producing "heat shock" proteins. It is not known whether they also exist in normal human joint, but this is feasible as natural joints heat up by ca 2.5 °C during walking and probably more during more intensive activities, cf. [46].

Nowadays researchers tend to believe that heat-induced bone necrosis is the secondary mechanism of tissue damage during implantation of cemented endoprostheses. RECKLING and DILLON [35] concluded their measurements of temperature at the bone–cement interface in acetabular component with the statement that temperatures high enough to cause bone necrosis are not attained during the polymerisation of the cement. Conversely, SCHATZKER et al. [37] measured temperatures of the order of 75–95 °C at the bone–cement interface *in vivo*. This value is high enough to cause thermal bone necrosis.

The toxic leftovers of the polymerisation process are nowadays considered to be a more challenging problem. The negative influence of residual monomer, which is a powerful fat solvent, is substantially prolonged when compared with pure high-temperature damage. Also the free radicals released from the cement dough during the process of polymerisation are chemically highly reactive and thus likely to cause protein coagulation and other undesirable processes, cf. [21]. Willert et al. cited by SWENSON et al. [43] report 3 mm necrotic zone in 3-week postoperative specimens. Possible damage mechanisms were identified to be PMMA polymerisation heat, free radical release and vascular damage. GOODMAN et al. [17] investigated the influence of acrylic cement on proximal humeral and proximal tibial methaphysis of the dog in the minimally loaded state. The animals were killed at 2, 4 and 5 months after the operation and the histological sections were done. Apart from other observations made by GOODMAN et al. [17] it is worth noting that the small quantities of the cement were found in the marrow spaces and in the Haversian canals indicating finger-like penetration outside the cement domain. The cement plugging of the Haversian canals resulted in localised areas of bony necrosis. Measurements of the remodelling activity were also made and the bone immediately surrounding the cement was significantly less active. The study presented in [17] suggests that after implantation into the site characterised by low level of mechanical loading the cement is encapsulated by thin connective tissue membrane containing scattered histiocytes and giant cells. Inflammatory cells were seldom observed. Unlike in the case of weight-bearing implantations the synovial-like fluid and fibrocartillage were absent. Although marrow necrosis did occur in the areas surrounding the cement implant, viable marrow has been found also in the nearest vicinity of the implant. This effect suggests that the blocking of the Haversian canals and not, or at least not only, the toxic influence of monomer leftover is the factor that contributes to the bone tissue necrosis.

It seems therefore important to model the polymerisation process as well as temperature distribution and to develop criteria similar to equation (18) for assessing tissue damage caused by polymerisation leftovers.

The eventual effect of bone necrosis is bone resorption at the bone–cement interface. The mantle of fibrous connective tissue is developed and the mechanical load-carrying capacity of the interface is seriously compromised, which usually leads to implant loosening and the need for reoperation.

Different measures are taken or considered to remedy this problem. These are:

1. Implanting cementless prostheses.

2. Development of new low-temperature and bioactive bone cements, cf. [36], [38], [50] and the discussion below.

3. Water cooling (acetabular parts of total hip implants, cf. [53]).

4. Shielding layers, cf. [20], [26].

5. Lowering thermal contact between cement mantle and the bone which can also have a beneficial effect of lowering monomer leftover, cf. [42].

6. Pre-cooling or pre-heating the prosthesis stem (possibly the bone and cement mixture).

7. Using as little cement as possible.

The most straightforward method is to refrain from using cemented implants at all, in favour of cementless ones. There are, however, extensive clinical data from postoperative follow-ups indicating that percentage of failures marked by the necessity of revision is significantly lower in the case of cemented prostheses, see [31] and the references cited therein. Also, cementless implants are avoided in the case of patients exhibiting weaker bones, like rheumatoid patients.

The most promising method is to develop new cements that polymerise at low temperatures leaving no monomer leftover. Another advantage of lowering the maximum temperature in the system (not necessarily bone temperature) is that some heat-labile cement components (e.g. antibiotics) are not deactivated during the cement setting and also the potential problems of MMA boiling and cement porosity induced in this manner are avoided.

Lowering the peak temperatures during the polymerisation process reduces also the overall shrinkage of the cement during the cooling phase. As calculated by HOLM [19] the final cooling of the cement is responsible for 40–50% of the final contraction (the coefficient of linear expansion $\alpha = 2.27 \times 10^{-4}$ [1/K], see [19] and the references therein). It is well known that the PMMA cement shrinkage is an undesirable effect that contributes to the loosening of the prosthesis.

The task of lowering polymerisation temperature can be partially accomplished by means of adding "heat sink" additives to the PMMA powder and/or changing initial P/L ratio of cement mixture. Unfavourable outcome of these actions is the lower mechanical strength of the cement since its porosity increases. These issues are discussed in detail in [20].

Pre-cooling the prosthesis stem is proved to be ineffective, cf. [4], [43], [47]; furthermore it was shown by BISHOP et al. [4] that the low stem temperature substantially deteriorates the cement–stem interface quality and that pre-heating should be used instead of pre-cooling. Pre-cooling of the cement mixture also prolongs the setting period of the cement. In that period, any motion of the installed prosthesis usually causes fixation failure. Implant has to be extracted from the femur, the cement has to be removed and new implant installed. Water cooling is possible only in fixation of acetabular components and is reported to have a positive effect on the bone temperature [53].

The possibility of lowering the thermal contact between cement and bone was also investigated, cf. [20], [26]. MAZZULLO et al. [26] showed that introducing thin layer of rubber-like material would have a beneficial effect on both the conversion of the monomer and temperature of the bone. Such a barrier could also protect the tissue against the diffusion of toxic substances from the cement dough. Unfortunately, in view of the efforts to create the best possible environment for the bone tissue to grow into and "interlock" with the cement, such a solution should be considered impractical.

3. Bone drilling and sawing during orthopaedic operations

As outlined in table1, a temperature increase during orthopaedic operations is of short duration; however, considerable temperature values may be reached. Two processes may lead to frictional bone heat-up during operation: sawing (e.g. during bone preparation for endoprosthesis implantation) and drilling (e.g. during preparation for screw fixation). The resulting thermal bone necrosis at the screw site leads to instability and consequently invalidates any benefits from any stabilising devices fixed to the bone. The correct choice of drilling/sawing parameters is therefore of importance. The parameters under consideration are:

1. Rate of rotation of the drill, speed of the saw.
2. Force on the drill or saw in the direction of drilling/sawing.
3. The kind of tool, level of wear.
4. Predrilling.
5. Cooling (irrigation) parameters.

The effectiveness of the above measures was measured experimentally, see [22], [25], [48], [49]. The outcome can be summarised as follows:

- The rotational speed of the drill is reported to have no marked influence on the spatial temperature distribution in bone but it affects the duration of the exposure to high temperature – higher drill speeds produced high temperature for shorter periods [25]. Similar measurements made by KRAUSE et al. [22] for high-speed (20,000 rpm and 100,000 rpm) cutting burs confirmed that that there is no general correlation between rotational speed of the bur and bone temperature, tendencies for different kinds of burs being different.

• The increase in force measured in the direction of drilling results in a significant drop in temperature in the vicinity of the drill. MATTHEWS and HIRSCH [25] report approximately 15 °C drop in temperature (from ca 82 °C) in location 0.5 mm from the drill when force is changed from 2 to 6 kG and further 17 °C drop with force changed to 12 kG, see figure 9. Similar effect was reported by KRAUSE et al. [22] for high-speed cutting burs – higher feed rates, and therefore cutting forces, cause lower temperature elevations. This tendency varied with kind of the bur used. Such an effect is much more pronounced when the duration of exposure to temperatures over 50 °C is compared. For forces 2, 6 and 12 kG these durations are 35, 8 and ca 1 sec. respectively, at the mentioned location. This effect is reported not to occur for dental burs and smooth pins used as drilling tools (as opposed to twist drills). Those have no means of eliminating the bone debris and therefore milling occurs rather than cutting and furthermore the debris gets compacted between the tool and the hole walls, thus greatly increasing the friction.

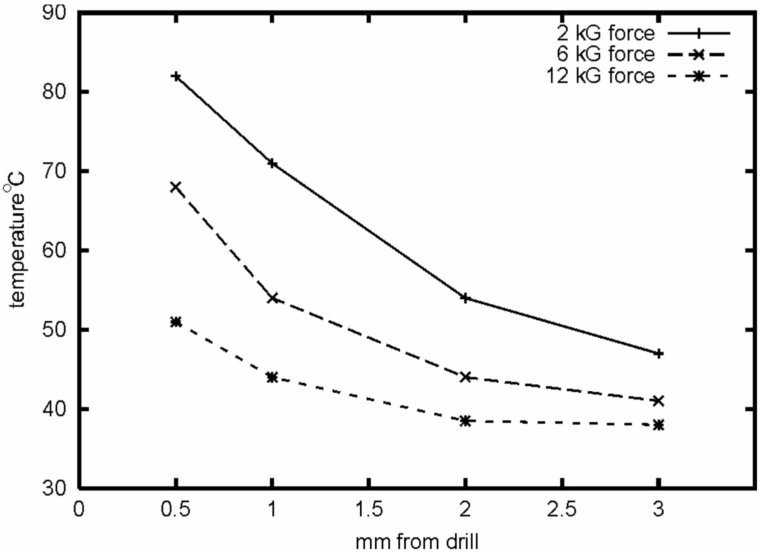

Fig. 9. Temperatures recorded in bone for different axial forces during drilling, after [25]

• Study of the influence of drill wear conducted in [25] showed that worn drills (used to drill 200 holes before) could produce temperatures over 20 K higher in the immediate neighbourhood than new ones. Also time of exposure was significantly prolonged. As can be inferred from this one and other experimental investigations the shape of the tool has a significant influence on bone temperature during drilling/cutting cf. [22], [48]. In [49], various geometries of saw-blade teeth were studied, without cooling. The resulting average temperature variation was 91–196 °C. Figure 10 presents a comparison of temperatures measured by KRAUSE et al. [22] for two different saw blades.

• The investigations of the effect of predrilling conducted by MATTHEWS and HIRSCH [25] showed temperatures during predrilling (drill diameter of 2.2 mm) and subsequent enlarging of the hole (diameter of 3.2 mm) to be virtually the same and, at the distance of 0.5 mm from the drill, by ca 60 K lower than that measured in control drilling (3.2 mm) (where temperatures exceeded 100 °C). This result can be viewed as the proof of beneficial effect of predrilling or as the rough measurement of the influence of drill diameter on maximum temperature attained, which turns out to be very high.

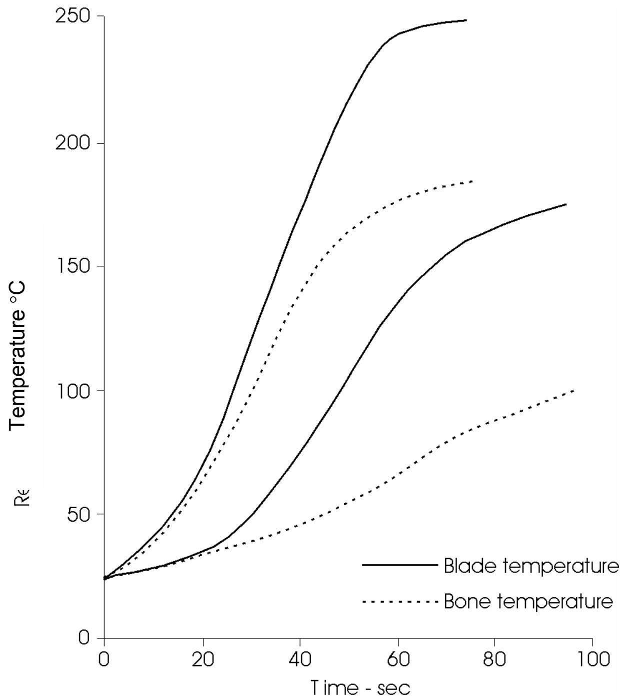

Fig. 10. Temperatures recorded for two saw blades, after [22]

• Cooling of the drill and surrounding bone by means of irrigation with water is reported to lower the temperature of the bone substantially. MATTHEWS and HIRSCH [25] found that for their experimental set-up and the diameter of drill they used (32 mm) no significant advantage was gained when raising the coolant flow above 500 cm^3/minute. The coolant they used was water at room temperature. Using pre-cooled water would probably allow less coolant to be used. The results of measurements conducted by KRAUSE et al. [22] for different kinds of burs and reciprocating saws confirmed conclusion that cooling may significantly reduce the bone and tool temperature. Studies by TOKSVIG-LARSEN et al. [48] with the prototype oscillating-blade saw showed that with an adequate coolant flow the temperature elevations are negligible and well below the values usually associated with bone necrosis.

As can be seen from the above considerations, appropriate choice of tools and cooling methods permits us to avoid the danger of thermal bone damage during preparation for an orthopaedic operation entirely.

3.1. Formulation of the problem

In the first approximation, the problem of bone heating-up during sawing can mathematically be stated as finding the temperature distribution in the infinite solid with moving line heat source. The geometry is depicted in figure 11. For steady-state conditions the solution is given in [6] in the form

$$T(x,z) = T_0 + \frac{q_l}{2\pi\,\lambda_b}\exp\left(\frac{Ux}{2a_b}\right)K_0\left(\frac{U\sqrt{x^2+z^2}}{2a_b}\right). \tag{19}$$

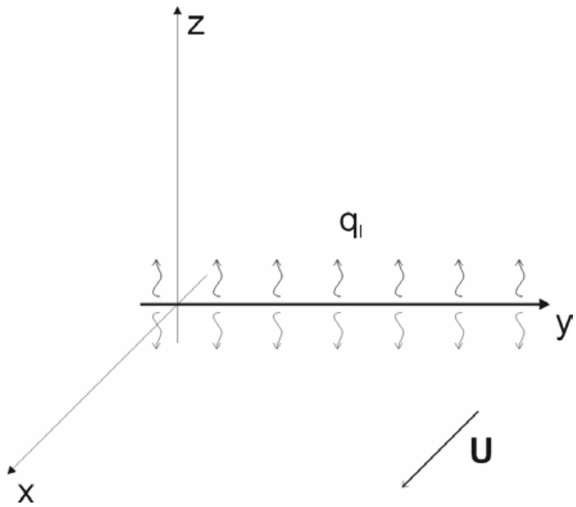

Fig. 11. Idealisation of the problem of bone cutting: line heat source parallel to the y-axis is embedded in the infinite medium moving in direction parallel to the x-axis. The heat generation rate is q_l [W/m]

It is assumed here that infinite line source located at the y-axis moves with the velocity U (sawing feed rate) in the direction of the x-axis; K_0 is the modified Bessel function of the second kind of order zero, whilst λ_b and a_b are the thermal conductivity and thermal diffusivity of the bone, respectively; q_l is the rate of heat generation expressed in [W/m]. The problem can be also visualised as the distribution of smoke in
a medium, which flows past the line emitting smoke, cf. [6].

The closed-form solution (19) can be obtained due to the simplicity of the model. The assumptions not reflecting the physical situation are:

• Treating the bone specimen as an infinite three-dimensional solid. Such an assumption should result in underestimating the temperature as the heat is allowed to escape the real geometrical domain of the bone without any boundary resistance. This is thought to have no significant influence as long as the sawing process is fast.

• Treating the saw as a one-dimensional entity. This is reasonable for steady-state situations (deep cut), when the temperature of the saw does not change anymore. The heat flow through blade in its direction and the blade heat capacity are thus neglected.

In such a formulation the crucial parameter q_l needs to be obtained experimentally by means of temperature or calorimetric measurements.

Alternatively to this formulation, Rosenthal (1946), cit. by KRAUSE et al. [22], proposed the solution for an infinite solid with moving plane heat source instead of line source.

The problem of bone drilling could be analogously modelled by infinite solid with moving point heat source. Again, the reported dependence of heat generated on feed rate (the velocity U) escapes modelling and needs to be supplied. Furthermore, the diameter of the drill is not a parameter either, whereas experimental data strongly suggest its importance. Such a model should be therefore considered as too simplified.

To summarise, we conclude that the problem of bone heat-up during drilling or sawing is nowadays not crucial since modern cutting tools are designed in such a way as to minimise or even completely eliminate the risk of thermal damage to tissue, cf. e.g. [48].

4. Frictional heat generation in joints

Friction occurs in all types of joints, both in natural and artificial. The heat generation and dissipation is therefore a process that takes place every time the joint is used. In normal human hip joints, the temperature elevation measured is of the order of +2.5 °C during walking and probably more during running, cf. [46].

The artificial joints are less efficient and one can expect that temperatures attained in such a joint can be higher. BERGMANN et al. [1] reported a 3.5 °C temperature rise in the case of titanium alloy hip prosthesis after 45 min of normal walking. The temperature was measured inside the neck of the prosthesis. The values at the bearing surface can be considerably greater. In [13] and [24], the finite element estimation of this temperature was performed based on the measured subsurface temperature for different articulating pairs. The zirconia, cobalt-chromium and alumina implant heads articulating on polyethylene cups were compared.

Investigation of the problem of the temperature distribution in an artificial joint when the joint is being used is performed for two specific reasons:

• To establish whether thermal damage to the tissue can take place. Such a damage may lead to prosthesis, usually hip acetabular cup or knee prosthesis, loosening via the mechanism outlined in section 3. The issue of criteria for thermal damage is more difficult here than in the case of bone cutting or cement polymerisation because the thermal loading is now cyclic and this may result in "thermal bone damage accumulation". On the other hand it is suggested that bone tissue can develop "heat shock" proteins and adapt to elevated temperatures, cf. [7], [28].

• To assess the working temperatures and their influence on wear and creep rate of materials commonly used for articulating surfaces. In the case of polyethylene (PE, UHMWPE) implant components, the three most important factors contributing to failure are: excessive stress (which may be magnified by the presence of residual stresses, cf. [29]), material oxidation due to γ-irradiation during sterilisation and thermal damage. The latter is vital to long-term prosthesis performance. According to YOUNG et al. [54], the Arrhenius-type relation can be applied to assess the time to failure t of polyethylene at elevated temperature T:

$$\log \frac{t}{t_R} = \frac{E_{\mathrm{act}}}{2.3B}\left(\frac{1}{T} - \frac{1}{T_R}\right).$$

Here the subscript R denotes design values, B is a universal gas constant and E_{act} is the material-dependent activation energy. When the values of constants appropriate for PE are assumed it appears that prosthesis service life is shortened by half when the temperature rises by 2 K over the design value.

Moreover, as pointed out by LU and MCKELLOP [24], at elevated temperatures the proteins present in the lubricant fluid in the experiments *in vitro* may precipitate forming a cushion shielding the bearing surfaces from wear and reducing the effect of the adhesive transfer of the polyethylene to metal surface. On the other hand, with the precipitation of soluble components, the lubricating quality of the fluid deteriorates.

The frictional heat generation has been investigated by various researchers in three distinct ways:

• *In vivo* measurements in patients. In [2], the temperature distribution and forces acting on the head of the implant were measured by means of instrumented endoprosthesis, see also [18] for technical details.

• *In vitro* measurements of frictional torque, temperatures and material wear on a laboratory set-up over a prescribed range of joint motions. Such measurements were made by DAVIDSON et al. [9], [10] and by LU and MCKELLOP [24]. Influence of the choice of materials for articulating surfaces and a number of other parameters were studied. Some of the results are reviewed below. In [9], the authors describe the so-called "simulated *in vivo*" test where movement of the prosthesis is replaced by resistance heater embedded in the prosthetic head and the system is assembled in such a way as to resemble the real situation as much as possible (the joint capsule is simulated by pieces of bovine muscle tissue), therefore reproducing occurring *in vivo* mechanisms of heat transport. However, the effect of vascularity and blood flow is not reproduced in such a set-up.

• Computer simulation. Unlike in the case of bone cement polymerisation, the stationary temperature field is usually considered, see [3]. This corresponds to the thermal equilibrium of the system when the heat generation rate by friction equals the heat dissipation rate. This happens approximately after 1 hour of continuous walking (the half of total temperature rise taking place in first 6 min.) according to *in vivo* measurements by BERGMANN et al. [2]. DAVIDSON et al. [10] reported half of this equilibration time for tests *in vitro*. LU and MCKELLOP [24] indicated much longer times, of the order of several hours, which is probably caused by different experimental protocol (much larger quantities of lubricant and lower forces in the joint). The greatest difficulty of computer simulations seems to lay in the fact that one needs to choose the parameters for the model correctly and to do this one needs to resort to some experimental data. The heat generation rate at the surface of contact of articulating components and thermal properties of tissues and synovial fluid must be obtained from experiments and, as was already shown, experimental results show considerable scatter. Furthermore, if only the Fourier–Kirchhoff equation is used, the effect of vascularity is neglected or can be only roughly approximated.

Below some of the important results reported in the literature are reviewed.

4.1. Factors influencing frictional heat generation

4.1.1. Materials

Material of articulating surfaces has a marked influence on the heat production and rate of wear. In [9] and [10], the authors provide an extensive study of these issues. Three kinds of articulating pairs have been investigated (mentioned here in stem-acetabular cup order):

1. Co-Cr-Mo steel on UHMWPE (Ultra High Molecular Weight Polyethylene).
2. Alumina (Al_2O_3) on UHMWPE.
3. Alumina on alumina.

Some of the results of *in vitro* tests are displayed in figure 13, after [9]. Specialised test set-up was used to produce rocking motion with variable hip loading. The profile of the loading during a single cycle is depicted in figure 12. The load–time history was selected to reflect natural hip loading during walking. The results cover experiments conducted with two values of maximum force applied to femoral head, namely 2500 N and 5000 N.

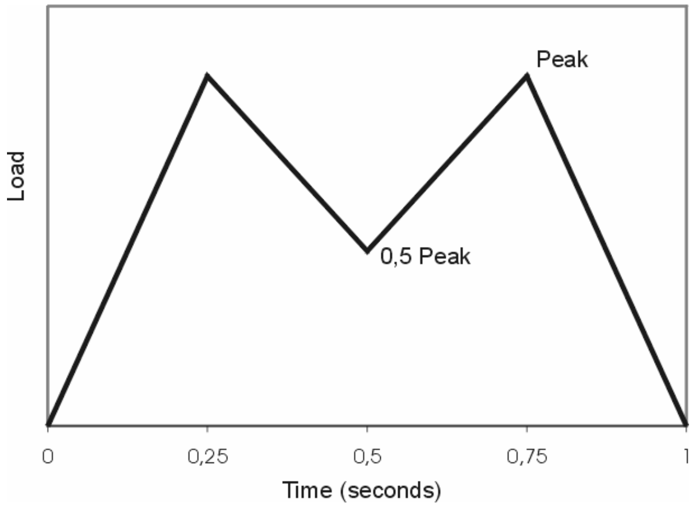

Fig. 12. Walking load history used in *in vitro* frictional heating tests.
Peak value was set to 2500 N or 5000 N, after [9]

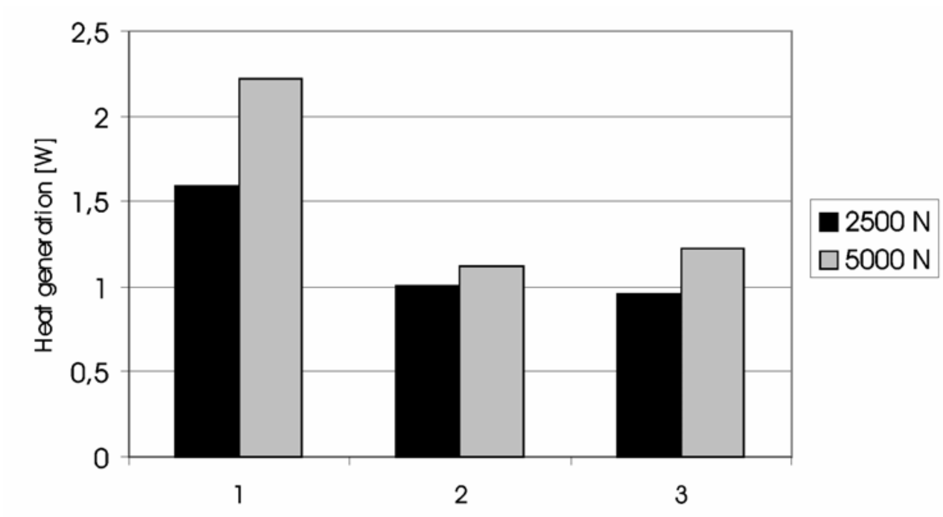

Fig. 13. Heat generation for different articulating surfaces: 1 – Cr-Co-Mo on UHMWPE,
2 – alumina on UHMWPE, 3 – alumina on alumina, after [9]

The results presented in [24] show superior performance of alumina-UHMWPE pair when compared to metal-UHMWPE pair *in vitro*. BERGMANN et al. [2] performed experiments which showed the superior performance of alumina-alumina articulation *in vivo*.

The differences in heat production for various pairs come from different friction coefficients and can greatly influence wear performance of artificial joint. More detailed information about the rate of wear for different kinds of surfaces can be found in [10] and [13].

4.1.2. Lubrication

Another factor important for frictional heat generation is the lubrication. In natural joints, it is accomplished by means of the *synovial fluid*, probably liquid-crystalline, biological substance, see e.g. [44].

The available volume of synovial fluid in the joint capsule is of order of 2 cm^3, depending on the type of joint. It is a yellow, clear and highly viscous liquid. It forms a film layer of thickness depending on the type of joint and location within it, in range from 6 μm to 1 mm. The synovial fluid in a healthy joint consist of water (to 94%) and hyaluronic acid (2–3% by weight). Moreover, the synovial fluid contains some macromolecular components like glycoproteins, phospholipids and low-molecular compounds, e.g. liquid crystalline cholesterol esters and small ions [44].

The main purpose of the synovial fluid is the lubrication of the joint. Furthermore, it provides the necessary nutrients for the cartilage and protects it from enzyme activity. The properties of the synovial fluid are affected by pathological processes. The shear viscosity coefficient is smaller for the synovial fluid from degenerated joints.

In the *in vitro* tests [10], the lubrication was attained by means of water or hyaluronic acid in different concentrations. The hyaluronic acid was chosen since it is the primary lubricant component of the synovial fluid. Additionally, friction in dry medium as well as in the presence of 2 mg bone cement powder was investigated. No significant difference between water and hyaluronic acid lubrication was reported, while friction in dry medium was, as expected, substantially greater. In figure 14, illustrative experimental relation is depicted between frictional torque in joint and lubrication conditions for steel-UHMWPE articulating pair [10]. To obtain these results the hip simulator was used to create a rocking motion over a 46° range. The axial load variation was chosen to reflect the walking loads history, see [10] for details.

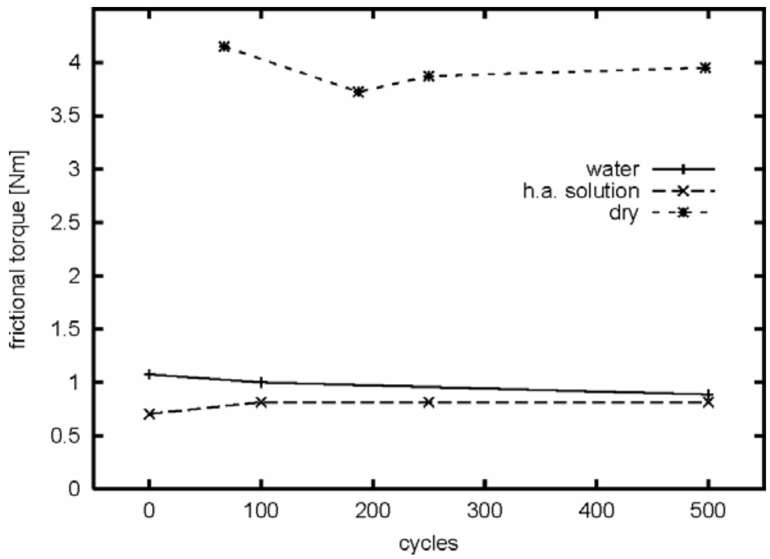

Fig. 14. Frictional torque in lubricated and dry media. Peak hip load = 5000 N, after [10]

The frictional torque rises by an order of magnitude when bone cement is present in the joint, even in a small quantity. As results presented in [10] indicate, there is an almost linear relation between frictional torque and equilibrium temperature rise.

A number of other factors influence the temperature distribution in joint. These are:

• Stride length and step frequency and consequently the flexion–extension angle and angular velocity, cf. [40].

• Adaptation. BERGMANN et al. [2] reported that two patients with low temperature in measured implanted joints had high body weight but were very active. This effect is assumed to have been caused by physiological adaptation of vascularity to higher temperature (higher perfusion rates). It is, however, not always present.

• Possible head–acetabular cup separation during the joint movement. This effect is not present *in vivo*, where various supporting structures exist to restrain the femur head (fibrous capsule, acetabular labrum, ligament of the head of the femur and the iliofemoral, ischiofemoral, pubofemoral and transverse acetabular ligaments). During the total hip arthroplasty some of those structures may get surgically removed or resected to facilitate surgical exposure. The kinematics of the artificial joint is therefore different from the natural one. Measurements performed by DENNIS et al. [12] prove that articulating surfaces separation occurs *in vivo*. The influence of this effect on joint temperatures is not known but it is suspected to be beneficial [2]. The gap that opens during the separation would be filled with synovial fluid, which would cool the articulating surfaces and the lubricating film would be renewed.

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