Finite element computer methods for design and fixation problems of orthopaedic implants

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Aseptic loosening of artificial joints fixed with acrylic bone cement is a major complication in orthopaedic implant surgery (Slooff et al., 1976). Stresses in the cement and on the cement–bone interface may give rise to fractionating of the cement or interface loosening. These stresses depend on loading and material and geometrical properties. For optimal design and use of implants it is important to evaluate the relationships between these properties and the stresses. The application of finite element methods (FEM) seems to be well suited for stress analyses of these irregular structures (Brekelmans, Poort, and Slooff, 1972). Three-dimensional FEM models of femoral hip joint implants in situ were reported by Bartel and Ulsoy (1975) and also by Hampton et al. (1976). If models like these are refined enough to be accurate, both computer time and data handling are limitations for extensive use. A two-dimensional model of femoral implants was studied by Andriacchi et al. (1976); this model was not realistic. McNeice, Eng, and Amstutz (1974) used the theory of composite materials in their two-dimensional model, to take the three-dimensional geometry into account; this method leads to correct displacements but to unrealistic stress fields. A better method was followed by Hampton et al. (1976), who used a spanning element to model three-dimensional geometry.

The stress analyses mentioned were aimed at the study of the mechanical behavior of one or more specific prostheses in situ. The results of these studies cannot be generalized as a rule. It is the object of this paper to show that by using simplified FEM models, the influences of various important parameters on the stresses in bone, cement, and implant can be studied with sensitivity analyses. In doing so, criteria for optimal design and use of implants can be developed.

METHODS

The cemented femoral component of a total hip prosthesis is modeled as an axisymmetrical rod, fixed with cement into a bony cylinder (Figure 1); materials are assumed to be linear elastic. A physiological joint load (F) is assumed to consist of a compression force (Z), a bending force (X), and a bending couple (M), all with respect to the axis. A comparable model was studied by Anand et al. (1976) in relation to bone ingrowth in porous material; they could, however, not take three-dimensional loading into account. Coordinates (x, y, z and r, ϕ, z), displacements (u, v, w), and stresses (σ_i and τ_{ij}) are defined in Figure 1. The stress tensor is symmetrical, so τ_{ij} is equal to τ_{ji}.

Displacements, resulting from the three loading cases, were calculated using a three-dimensional isoparametric element (Clough, 1969) and also using an axisymmetrical ring element that can take non-axisymmetrical loading into account by expanding loads, displacements,

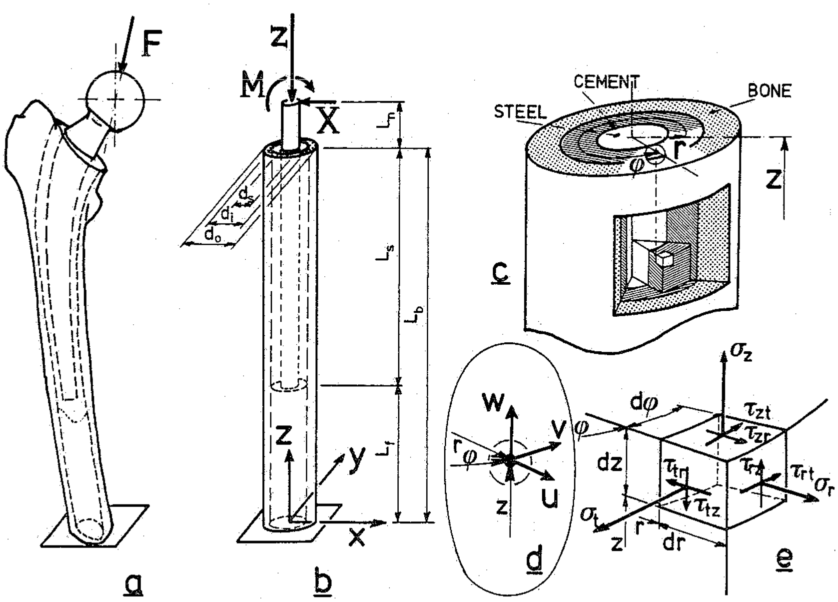

Figure 1. Model (b) of a femoral hip prosthesis in situ (a). Geometrical and loading parameters (b), coordinates (b, c), displacements (d), and stresses (e) are defined.

strains, and stresses into Fourier series (Wilson, 1965; Huiskes, 1974) (see Figure 2, models 1 and 2, respectively). The results were compared with regard to computer time, accuracy, and data handling. In a finer mesh the same "Fourier" element was used for model 3. In this model stresses were also calculated while varying the loading and geometrical and material properties.

RESULTS

In Figure 3 displacements calculated with models 1 and 2 are compared. Computer time for model 1 was ca. 20 times longer than for model 2, apparently with comparable accuracy.

Some results of the stress calculations with model 3 for loading case X are shown in Figure 4. Because of the "action-reaction" law, σ_r, τ_{rt}, and τ_{rz} have to be continuous across the interfaces. In Figure 4 the three continuous and three discontinuous stresses are drawn in different graphs.

Figure 5 gives the shear stresses at the bone-cement interface for the three loading cases, as well as the axial stresses in the stem and the bone, both at the interfaces. The axial stresses in the bone are compared with "natural" stresses—i.e., axial stresses present if the same loads were applied to the empty cylinder. For these calculations Poisson's ratio of the materials was assumed to be 0.33; Young's modulus for bone: 20,000 (N/mm^2), for cement: 2,000 (N/mm^2), and for steel: 200,000 (N/mm^2). Z = 1,000 (N), X = 100 (N), and M = 10,000 (Nmm). d_o = 30 (mm), d_i = 20 (mm), and d_s = 10 (mm). L_b = 205 (mm), L_s = 130 (mm), and L_n = 25 (mm).

To illustrate the possibilities of sensitivity analyses with the model, Figure 6 compares different stress values at the interfaces, as function of the z coordinate, calculated for different stem lengths for loading case Z.

CONCLUSIONS

Although the use of the "Fourier" element is confined to axisymmetrical constructions (and not only cylindrical constructions; a curved contour is possible), it can conveniently be used in simplified models of implanted orthopaedic prostheses to study the influences of important design parameters on the stresses. The stress distribution in the structure depends on magnitude but also very much on direction and point of application of the applied load. Stress concentrations in prosthesis, cement, or bone will likely occur near the tip and even more so near the collar of the stem. Optimization should be directed toward these two regions. These stress concentrations become even more pronounced if the cement material is made stiffer. The shear stress τ_{rz} dominates τ_{rt} at the inter-

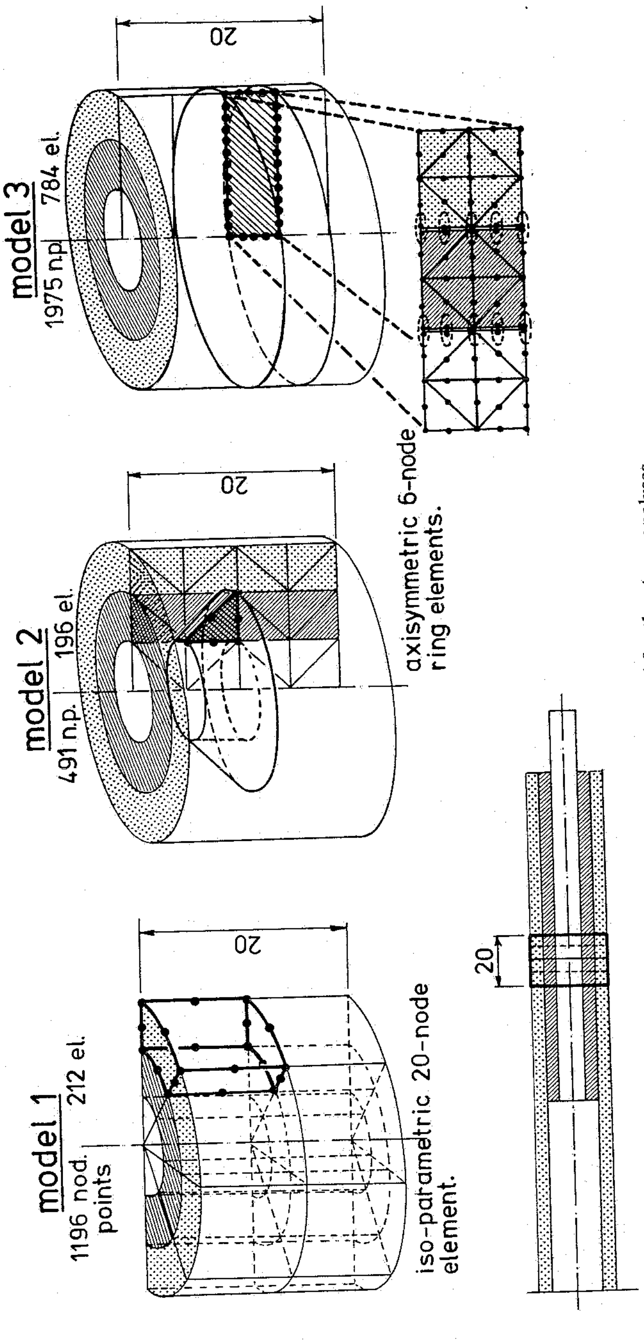

Figure 2. Different elements and more-or-less refined meshes were used for the stress analyses.

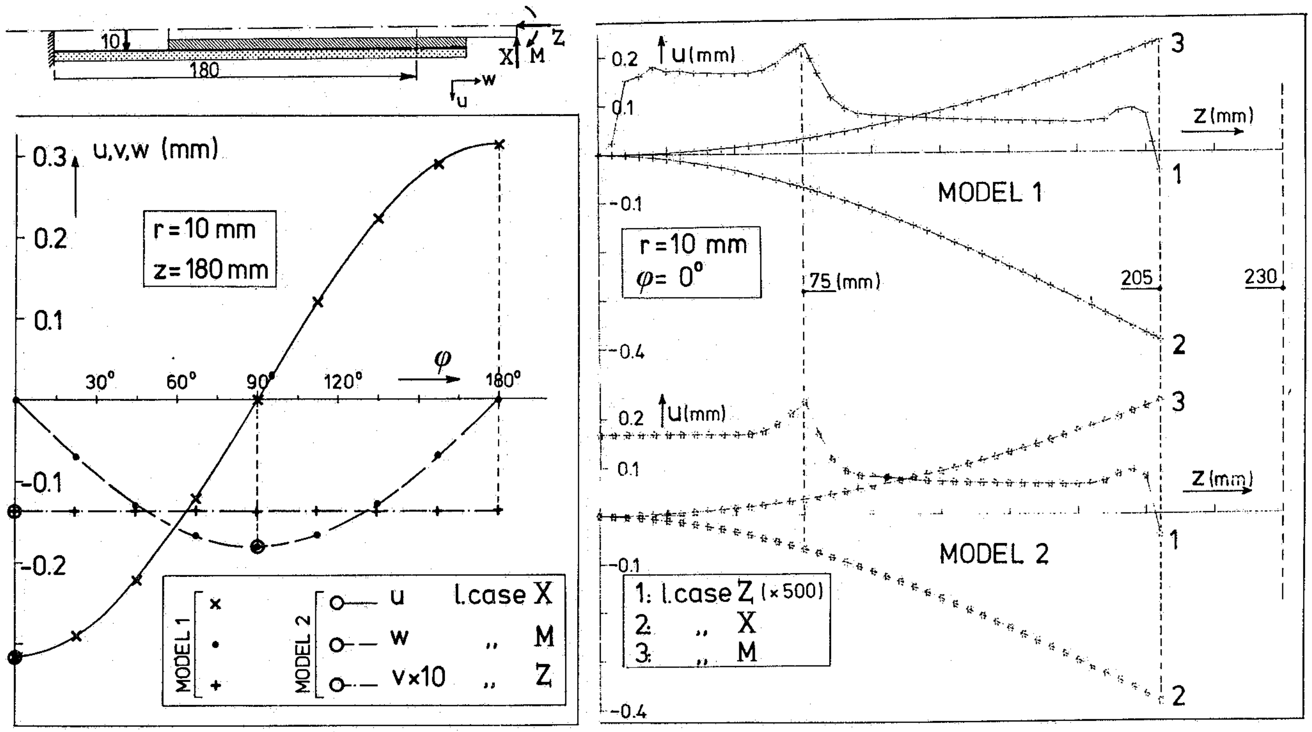

Figure 3. Comparison of results as calculated by models 1 and 2. Radial (u), tangential (v) and axial (w) displacements for the three loading cases, respectively, (*left*) on a circumferential line at the cement-bone interface (displacements are symmetrical with respect to $\phi = 180°$) and radial displacements (u) for three loading cases (*right*) of an inner line on the bone contour. (Scales of curves are not identical.)

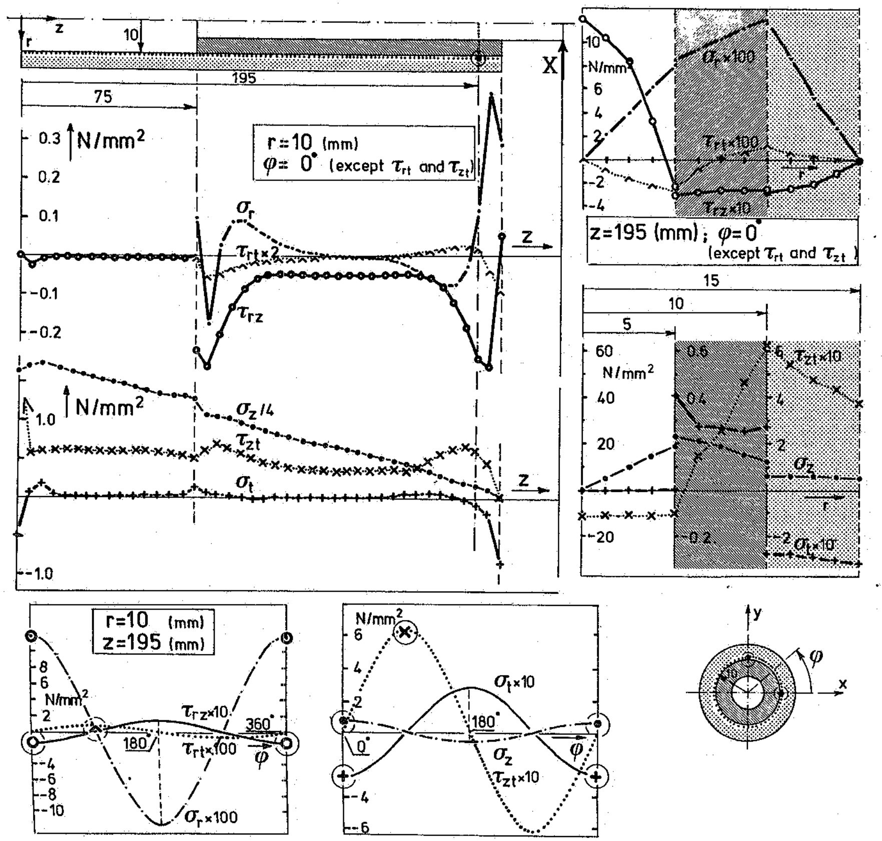

Figure 4. Six independent stress parameters on three different lines for loading case X. Different graphs are drawn for continuous and discontinuous stresses (discontinuous stresses as present in the bone material). τ_{rt} and τ_{zt} are zero for $\phi = 0°$; *top graphs* show the values at $\phi = 90°$.

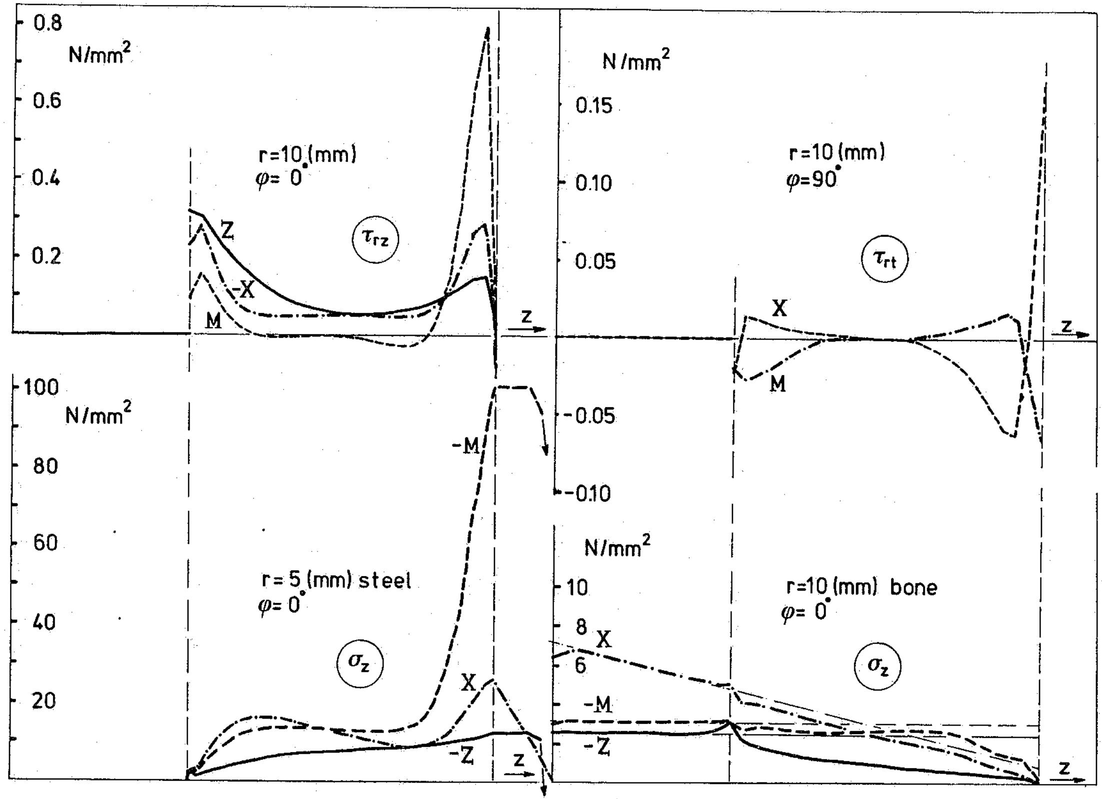

Figure 5. Shear stresses at the bone-cement interface (*top*), axial stresses in the stem (*left bottom*), and axial stresses in the bone (*right bottom*) for three loading cases. *Thin lines* in the *right bottom graph* indicate "natural" stresses. Some curves (as indicated) have been inverted in sign.

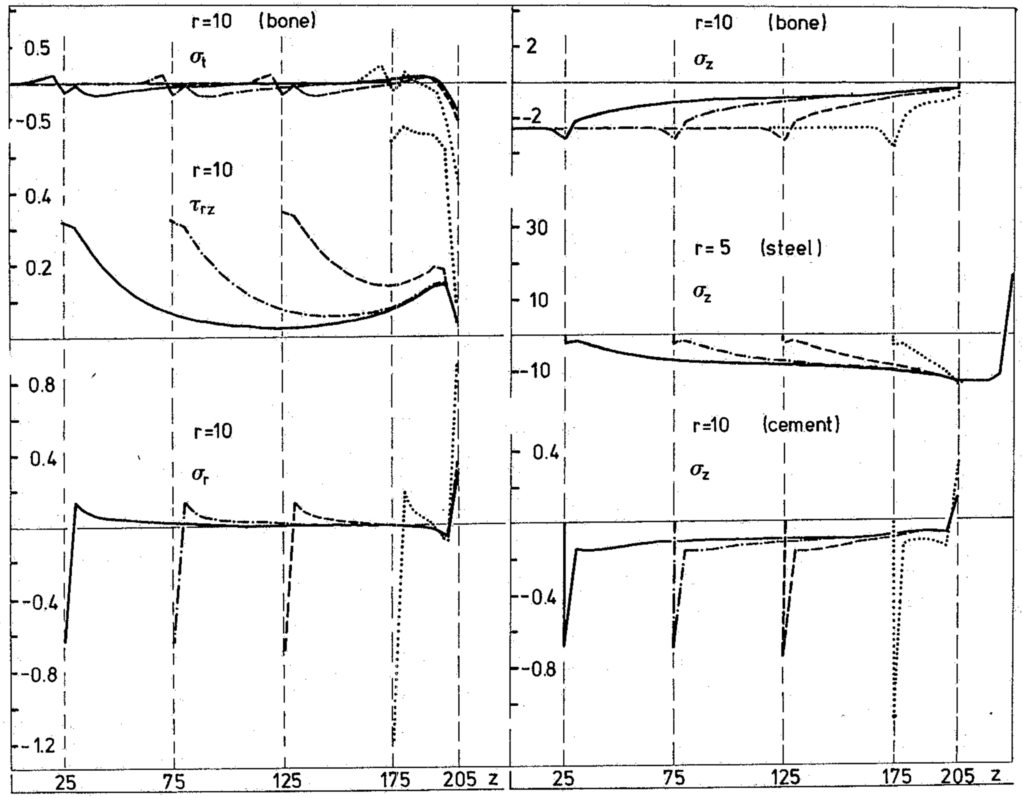

Figure 6. Stresses as calculated with loading case Z for four different stem lengths. Values are independent of ϕ. Dimensions are in mm and N/mm^2. Stresses are continuous if no material is indicated.

faces; it could give rise to interface loosening at the tip or at the collar.

The introduction of bending forces and moments to the bone across the cement-bone interface by σ_r and τ_{rt} takes place in the proximal part of the bone. The axial stress σ_z dominates other stresses in the stem and in the bone; it equals by approximation the equivalent stress. This means that the stress normal to the cement-bone interface (σ_r) is, as far as the bone is concerned, of insignificant magnitude. This is not true for the cement; here the stress situation is more complex—i.e., several stresses are of comparable magnitude although τ_{rz} has the most significant influence.

Flexibility of the structure in bending is mainly determined by the properties of the bone. This means that less stiffness of the stem and greater stiffness of the bone will lower the stresses in the stem. There exists an optimal stem length for a medullary prosthesis; a longer stem can even give rise to higher stress concentrations. This optimal length probably depends on other geometrical and material properties. In general a stem should not be very long and not very stiff.

To obtain lower stresses in the connecting layer and distribute them better over the interfaces, it is advantageous to use a material for this layer with low values of Young's modulus and Poisson's ratio.

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