Stress analyses of implanted orthopaedic joint prostheses for optimal design and fixation
Huiskes, H.W.J.

Published in:
Acta Orthopaedica Belgica

Published: 01/01/1980

Document Version
Publisher’s PDF, also known as Version of Record (includes final page, issue and volume numbers)

Please check the document version of this publication:

• A submitted manuscript is the author's version of the article upon submission and before peer-review. There can be important differences between the submitted version and the official published version of record. People interested in the research are advised to contact the author for the final version of the publication, or visit the DOI to the publisher's website.
• The final author version and the galley proof are versions of the publication after peer review.
• The final published version features the final layout of the paper including the volume, issue and page numbers.

Link to publication

Citation for published version (APA):
Stress analyses of implanted orthopaedic joint prostheses for optimal design and fixation

by R. HUISKES

Laboratory for Experimental Orthopaedics
Department of Orthopaedic Surgery, University of Nijmegen
Department of Applied Mechanics, Eindhoven
University of Technology, The Netherlands

Introduction

Aseptic loosening of orthopaedic joint prostheses is still a major complication in orthopaedic surgery and will probably become even more problematic in the future, since the ages of patients for whom an arthroplasty is indicated, tend to decrease.

There can be little doubt, that loosening is a direct result of high stresses in the connecting layer between prosthesis and bone (e.g. bone cement, PMMA) or at the interfaces. Also prosthesis fracture, often reported in the American literature, is stress related.

Stresses in the materials result from the physiological joint loading and depend also on the mechanical interaction between the different structures in the system. This interaction depends on the stiffness characteristics and hence on the geometrical and mechanical material properties of the prosthesis, cement layer and the bone.

Given the loading characteristics of the joint and given the bone in which the prosthesis has to be implanted, stress values may be limited by using an optimal prosthesis design, a prosthesis and interconnecting layer with optimal material properties and an optimal implantation procedure as to its mechanical consequences. It has been recognized recently in orthopaedic research literature that the optimal design and material parameters can only be evaluated using stress analyses, either experimental or theoretical.

It is the object of this paper to show that experimental stress analyses of intramedullary fixation systems give limited information, but are useful if the results are interpreted correctly; furthermore to show that useful information can be acquired using simplified theoretical simulation.
models, but only if one has a sound understanding of the influences of the assumptions on which the simplifications are based.

Some results of experimental and theoretical analyses are presented.

Experimental stress analysis

Since in photoelastic models the different material properties in the structure cannot be simulated, the brittle coating and strain gauge techniques appear to be the only ones suited for experimental analyses of the prosthesis-bone system. Of these techniques, strain gauges have by far the most general application possibilities (Durelli, 1977).

Extensive strain gauge measurements on a loaded cadaveric femur, intact as well as provided with prostheses, were performed (v Heugten, 1975; Huiskes et al., 1976). On the femoral shaft and the collum, 112 rosette strain gauges were attached. Forces in three perpendicular directions \( F_x, F_y, F_z \) and pure couples in three planes \( M_x, M_y, M_z \) were applied on the head. From the three strain values, registered for each strain gauge, the magnitudes and orientations of the principle strains were calculated. Assuming Young's modulus of the bone as 20,000 N/mm² and Poisson's ratio as 0.37, principle and equivalent stresses were calculated from the strains, using Hooke's law.

The measurements were repeated after providing the femur with different kinds of intramedullary nails, bone plates and non-cemented as well as cemented hip prostheses, among others a short stem and a long stem Müller prosthesis. The experiment is illustrated in figure 1.

It is not the object of this paper to discuss the results in general. Figure 2 shows an example of results: strains as function of the longitudinal distance along a medial line on the shaft, upon loading of the head with a vertical force \( F_y \), for the intact femur as well as the femur with the above mentioned Müller prosthesis. In theory the stress values should be identical on the distal femur in all three cases, since this part « feels » only the overall load and is not directly influenced by the presence of the stem. The fact that this proves to be untrue, is caused by two factors (fig. 2 and 3): because the positions of the femoral head and the prostheses heads differ, an extra moment is introduced; this changes the overall load. This also means that differences in strains measured on the proximal femur cannot be accounted to the influence of the stem. Secondly, the femur proved to show a considerable amount of geometrical non-linearity (fig. 3), meaning that an extra moment is introduced through displacement of the head upon loading with a force. Application of the same force.
magnitudes in positive and negative direction, which would result in identical absolute strain values if the system behaved linearly, resulted in strains that were, in absolute value, about 50% as high for the downward direction. The influence of this phenomenon is dependent on

the stiffness of the structure, hence also on the presence of a prosthetic stem.

These two factors make it quite difficult, not to say impossible, 
interpret the differences in strain values in terms of stem influence. Moreover, one cannot expect to be able to evaluate the mechanical influences of different prostheses by comparing strain values in a few single strain gauges, as was done by Oh and Harris (1978); even after
FIG. 2. — Strains on a medial line on the femoral shaft measured while a force was applied on the head.

FIG. 3. — Upon load with a force, strain values are not comparable because of geometrical non-linearities.
FIG. 4. — Principal stresses on the femoral surface, as calculated from strain values, measured with a couple applied on the head. Intact and provided with two prostheses with different lengths.
correction of the force direction, as they did, considerable inaccuracies will remain.

These disadvantages disappear, if a pure couple is used for loading, because the influence of a pure couple is indifferent to a change in its location. Although this type of loading is not physiological, it gives optimal possibilities for a comparison of stem influences.

Figure 4 gives an example of resulting stresses upon loading of the head with couples in two planes. The stress values on the distal side are practically identical for the three cases and the differences in stress values on the proximal side are purely due to the stem influence. As was also established, a couple of negative orientation gave, in absolute sense, identical strain values.

From the information in figure 4 and similar results, it can be concluded that, because of the proximal reduction of the bone stresses, the prosthesis stem is taking a part of the load. It can also be established, that the reduction in bone stress is higher if a stiffer stem is used. Nothing, however, can be concluded with respect to stresses in the stem, in the cement, in the bone and at the interfaces. An experimental analysis of this kind gives only limited information and theoretical analyses are needed to supply the knowledge necessary to evaluate optimal design and material parameters.

**Theoretical stress analyses**

For theoretical stress analyses of the intramedullary bone prosthesis structure only beam theories and finite element methods (FEM) can be used. Because the applicability of beam theories is quite limited, FEM are usually preferred.

In using the FEM, a model of the structure has to be developed. This model is based on a mathematical description of the geometrical, material and loading properties of the real structure. Because of the complexity of these properties, the descriptions used are approximations, based on more or less far-fetched assumptions.

The material properties are usually assumed to be linear elastic, isotropic and homogeneous, although certainly for bone and probably also for bone cement (PMMA) this is a very rough approximation. Although the physiological loading of the hip varies to a great extend, usually a few static loading types are taken into consideration; mostly those related to specific functions like standing on one or two legs. In most cases only loading in one plane is taken into account.
The geometry of the structure can either be modelled as two-dimensional or as three-dimensional. Only with a three-dimensional model the actual three-dimensional state of stress can be calculated.

A complicating factor is presented by the connections of the materials in the structure, which will be loose on tension and under certain conditions also on shear. With one exception (Svensson et al., 1977) the interfaces are assumed to be fixed.

For a two-dimensional model there are several options (fig. 5).

1. The structure may simply be approximated according to its longitudinal cross-section. This results in a type of sandwich construction, with no connection between the medial and lateral cortex. Andriacchi et al. (1976) analysed a model of this kind.

2. The connection can be simulated using a «spanning element», as was done by Hampton et al. (1976) and Svensson et al. (1977); the stiffness of this spanning element has to be evaluated, using experiments or beam theories.

3. Another possibility is the use of composite materials theory, as was applied by McNeice et al. (1976). The modulus of elasticity of an element in the two-dimensional model is then derived from the geometrical and material properties of the bone, cement and stem fractions.
in the anteroposterior region that the element represents. Although such a two-dimensional model shall be accurate in its flexibility characteristics in the frontal plane, it is not in the calculated stresses.

For an, at least fairly, numerical accurate three-dimensional model, the element mesh should not be too rough, resulting in a system of very many degrees of freedom. Consequently the use of such a model is restricted by computer cost and data handling tasks.

The three-dimensional model reported by Hampton et al. (1976) is probably too rough to offer results that have fair accuracy. Very few results of this model have been published; the same is true, at least up to now, for the three-dimensional model claimed by Vichnin et al. (1977).

Although the reader is often impressed by the geometrical perfection, in one plane or in three dimensions, of the above mentioned FEM models, there is considerable uncertainty about the influences of all the simplifying assumptions with respect to the material properties, as there is still much uncertainty about physiological joint loading. Moreover, two-dimensional models give only plane stress components, while the effects of out of plane loading remain uncovered. Because of the uncertainties, the results of these analyses cannot be interpreted in absolute and detailed terms.

With theoretical simulation models, however, parameters can easily be varied, so that their influences on the mechanical behaviour of the structure can be evaluated. In this way the effects of certain assumptions can also be studied.

For an extensive and systematic parameter analysis, two conditions have to be met. The model should be simple and cheap enough to be used for many calculations and the amount of parameters should be restricted. On the other hand, of course, the model should be realistic enough to allow for valid results and to guarantee this, the relation between the model and the structure should be well understood.

The models published in the literature often do not meet these requirements. Indeed, very often detailed conclusions are derived from results with models that only in their geometrical aspects have some relation with reality, and in which the influences of parameters are only occasionally analysed.

A simplified model of intramedullary implant fixation

To be able to analyse the influences of the most important parameters on the three-dimensional stress distribution extensively and also to be
able to study the possibilities and limitations of different analysis methods, a simplified model of the intramedullary fixation system was used (fig. 6). The model is axisymmetric and was analysed with FEM, using ring elements that can take non-axisymmetric loading into account by expansion of loads, displacements, stresses and strains into Fourier series. A discussion of this method and a few results were published previously (Huiskes et al., 1978; Huiskes and Sloof, 1978). A comparable model was analysed by Bartel (1977). The parameters, taken into account, are summed up in table I.

For each case the three-dimensional stress components and the equivalent stresses, based on the criterion of maximum deformational energy (fig. 7), were calculated.

In figure 8 the equivalent stresses on a longitudinal line on the outside surface of the bone cylinder, upon loading with a bending moment, are shown for the bone without implanted rod and for the bone with two rods of different length. By comparing these results with those in figure 4, it can be seen that the results match the experiments in a qualitative way. The model, however, gives the opportunity to investigate the relation of these outside surface stresses with the internal stress distribution.

As was already shown for the experiments, the stress distribution in the model proved to be very sensitive for the direction of an applied
TABLE I

Parameters of the model

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Symbol</th>
<th>Dimension</th>
<th>Values and variations</th>
</tr>
</thead>
<tbody>
<tr>
<td>height</td>
<td>( \ell_s )</td>
<td>mm</td>
<td>80, 180, 130, 200</td>
</tr>
<tr>
<td>breadth</td>
<td>( d_s )</td>
<td>mm</td>
<td>10, 15, 20</td>
</tr>
<tr>
<td>Young's modulus bone</td>
<td>( E_b )</td>
<td>N/mm²</td>
<td>( 2 \times 10^9, 1 \times 10^9 )</td>
</tr>
<tr>
<td>Young's modulus cement</td>
<td>( E_c )</td>
<td>N/mm²</td>
<td>( 2 \times 10^9, 1 \times 10^9 )</td>
</tr>
<tr>
<td>Poisson's ratio bone and stem</td>
<td>( v )</td>
<td>—</td>
<td>0.33, 0.4, 0.2</td>
</tr>
<tr>
<td><strong>load</strong></td>
<td>( P )</td>
<td>N</td>
<td>1,000</td>
</tr>
<tr>
<td><strong>force</strong></td>
<td>( X )</td>
<td>N</td>
<td>100</td>
</tr>
<tr>
<td><strong>moment</strong></td>
<td>( M )</td>
<td>N·mm</td>
<td>10,000</td>
</tr>
</tbody>
</table>

- \( L_s \) and \( d_s \): Bone inner diameter \( d_i = 20 \text{ mm} \); outer diameter \( d_o = 30 \text{ mm} \).
- Poisson's ratio bone and stem: 0.33.

![3-D FEM model](image)

**FIG. 7.** Equivalent stresses in stem, cement and bone.
force, as is illustrated in figure 9. This means that it is quite unrealistic to analyse a model for a specific physiological load, if general conclusions have to be derived.

It can be established that, with exception of the most proximal region of the bone, the stem and the bone behave approximately according to beam theory, which implies that the axial stress component \( \sigma_z \) in these materials is by far the most significant. The stress state inside the cement, however, is truly three-dimensional. In two-dimensional analyses the stress situation in the cement will therefore probably be underestimated.

Certain aspects of the mechanical behaviour, however, especially stresses in the stem, can conveniently be analysed using beam theories, as for instance is being done by Calderale and Gola et al. (1977).

Figure 10 shows equivalent stresses in the stem upon transverse loading for three different thicknesses. Obviously the often expressed opinion

Acta Orthopaedica Belgica, Tome 46, Fasc. 6, 1980
that a stem should be thick in order to prevent fracture is not quite correct. As can be seen in figure 10, showing the part of the load that is being taken by the stem, a thicker (stiffer) stem takes more load, which means that there exists an optimal value for the flexi-

![Diagram](image)

**FIG. 10.**—Equivalent stress in the stem (left) and the part of the bending moment that is taken by the stem (right). Transverse loading case; three different stem diameters.

![Diagram](image)

**FIG. 11.**—Equivalent stresses in the stem (left) and the cement (right) on longitudinal lines near the stem-cement interface, as calculated for the model. Transverse loading; three different ratios of stem and bone moduli of elasticity.

A stiff stem will also cause high concentrations in the cement, especially near the distal tip. Moreover, a flexible stem will result in a more natural stress distribution in the bone, especially when it is combined with a calcar collar.

The stress distribution is greatly dependent on the flexibility ratio between stem and bone, which is illustrated in figure 11; obviously a thick
(stiff) stem should not be implanted into a flexible (osteoporotic) bone.

A thin stem, however, has a bad influence on the normal stress on the stem-cement interface ($\sigma_n$). So altogether it would be advantageous to design a stem that is thick, although flexible by means of a low modulus of elasticity.

Controversial statements are found in the literature with respect to the significance of the properties of the interconnecting layer. It could be established that both modulus of elasticity and Possion's ratio (compressibility) have a marked influence on all stress components in the cement and at the interfaces. Lower values for both these material constants smooth the stress distribution, so the use of a flexible and compressible material would be advantageous. These specifications do not fit easily on well known materials. Silastics, for instance, although

Acta Orthopaedica Belgica, Tome 46, Fasc. 6, 1980
flexible, are quite incompressible. Porous acrylics would be both flexible and compressible, although perhaps not strong enough.

To be able to judge the possibilities and limitations of different kind of models, the axisymmetric structure was analysed using several different methods (fig. 12). Model 3 being the model discussed here, model 1 (Huiskes et al., 1978) refers to the same model analysed with three-dimensional isoparametric elements. Computer costs and data handling tasks for model 1 proved to be enormous, which makes it unsuitable for parameter sensitivity analyses. Model 2 is no longer of any significance.

Model 4 is a two-dimensional FEM approximation of the axisymmetric structure, using a spanning element to represent the in-plane stiffness of the bone cylinder. Model 5 is identical to model 4, but allows the stem-cement connection to loosen upon tensile stress, by using an iterative process with the FE calculations; also sliding is possible between the two materials. Model 6 is an analytical model based on the theory of beams on elastic foundation; stem and bone are assumed to be each others elastic foundation, separated by an elastic layer.

Using models 1 or 3, the three-dimensional stress components can be calculated. Models 4 and 5 give only plane stress components. Model 6 gives axial stresses in stem and bone for all loading cases, interface shear stresses for axial loading (6a) and interface normal stresses for bending and transverse loading (6b). For all the models bending and

![Image](https://example.com/image.png)

**FIG. 13.** A comparison of results from models 3, 4 and 6a. Shear stress at the bone-cement interface upon axial loading.
FIG. 14. — A comparison of results from models 4 (two-dimensional) and 5 (two-dimensional with loose stem); normal stress at the bone-cement interface on two longitudinal lines; loading with a bending moment of 10,000 Nmm.

FIG. 15. — A comparison of stem equivalent stresses, as calculated for a fixed stem (model 4) and a loose stem (model 5); transverse loading.
axial compression stiffness of stem and bone are identical. The axial loading case was not investigated with model 5.

The results of models 3, 4 and 6 for the axial loading case, proved to be comparable and show reasonable agreement (fig. 13). For the bending and transverse loading cases fair agreement is achieved between the results of models 4 and 6. Although the results of models 3 and 4 show comparable tendencies for these loading cases they are difficult to interrelate. Generally speaking the stresses in stem and bone show reasonable agreement, while the stress state inside the cement and at the interfaces is somewhat underestimated in the two-dimensional model.

A comparison of results from models 4 and 5, shows that a loose interface between stem and bone has a marked influence on the stress distribution, especially in the cement and at the interfaces. While stresses in the bone remain practically unchanged, the stress in the stem is higher for the loose case (fig. 15). Stress concentrations in the cement (equivalent stresses) are also higher for this case; shear stress, being zero at the stem-cement interface, is small at the bone-cement interface, while practically no tensile stress occurs (fig. 14).

Conclusions

Strain gauge experiments for evaluation purposes of hip-prostheses designs and materials, give limited information, but are, if interpreted correctly, useful in combination with theoretical analyses. Theoretical simulation models have very limited value if the influences of their underlying assumptions are not investigated, even when they show perfect geometrical conformity with reality. By using simplified models, extensive parameter analyses are possible, so that a sound general understanding of the relation between the parameters and the stress behaviour can be developed.

Axisymmetric, two-dimensional and beam models may be used to analyse the intramedullary fixation system, each having its own possibilities and limitations, some of which were demonstrated here.

Acknowledgement

The experiments were carried out by J. IJzermans and P.C.M. van Heugten at the dept. of Applied Mechanics, Eindhoven University of Technology (Chairman: J.D. Janssen).

The experimental femur was operated by T.J.J.H. Sloof (dept. of Orthopaedic Surgery, University of Nijmegen). For the finite element calculations a computer system called FEMSYS, developed by J.P.A. Banens (Computer

Acta Orthopaedica Belgica, Tome 46, Fasc. 6, 1980
Centre, Eindhoven University of Technology), was used. J. van Heck (Eindhoven University of Technology) assisted with data handling for the two-dimensional models.

This work was presented in May 1978, at the 2nd Meeting of the European Soc. of Biomechanics.

BIBLIOGRAPHY


R. HUISKES
Dept. Orthopaedics,
University of Nijmegen,
6500 HB Nijmegen, The Netherlands.