

Interface stresses in the resurfaced hip

Finite element analysis of load transmission in the femoral head

The load transmission and interface stresses in the Wagner resurfaced femoral head were evaluated for the purpose of studying possible failure mechanisms. We found that unnatural stress patterns occur in the head and at the implant-bone interfaces, in addition to regions of stress protection in the bone, possibly enhancing interface failure and bone remodeling. However, these stresses are not higher than those reported for other kinds of prostheses, e.g. acetabular cup, tibial plateau. From these findings, together with clinical observations, it is hypothesized that the femoral surface cup is more sensitive to local loosening than other prostheses. This hypothesis would indicate that prosthetic designs should be analysed relative to their potential to provoke failure propagation, rather than only initiation of mechanical failure and loosening.

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Wagner (1978) suggested that after surface replacement the stress distribution in the proximal femur would be more physiological than conventional intramedullary fixation. In addition, it seemed attractive that the load transmission across the implant-bone connection takes place in compression.

The purpose of our study was to evaluate the 3-dimensional load-transfer mechanism and the stress distribution in a proximal femoral structure with a Wagner surface replacement. The results were used as a base for comparisons with clinical and animal experimental findings, and to study the nature of the stress transfer concept in more detail.

Material and methods

The Finite Element Method (FEM) is suited to analyse stresses in complex biological structures (Zienkiewicz 1977, Huiskes & Chao 1983, Huiskes 1984). The application of this method implies the development of a model which mimics the real structure in essential features. These features can be categorized into: (a) geometry of the structure, (b) properties of the materials involved, (c) properties of the

connections between the materials and the boundaries of the model, and (d) the loading conditions. After delineation of the model as a geometrical entity with appropriate boundary conditions, the model is mathematically divided into little blocks or elements, connected in nodal points. In the computer simulation model, the separate elements are assigned material properties (elastic constants), and boundary nodes are provided with the appropriate loads. The computer program then calculates the strains and stresses in the nodal points by solving a set of equations.

A sagittal section of the three-dimensional model utilized in the present investigation is shown in Figure 1. The coordinate-angle Θ (Figure 1 and 3) denotes the circumferential location of a longitudinal plane; $\Theta = 0^\circ/180^\circ$ comprises the sagittal plane, and $\Theta = 90^\circ/270^\circ$ the coronal plane. The load F_1 represents the one-legged stance by approximation. F_1 is the resultant force of a distributed load over a circular area of the cup, ellipsoidal in contour (Figure 1a), with a unit force magnitude (1 Newton).

Our model comprised the head-neck region only (Figure 1). The rationale for such a limitation is twofold. In the first place it is obvious from mechanical considerations that the stress-distribution away from the head-neck region is not affected by the presence of the cup, which was actually once more proven experimentally by Oh et al. (1979) and Shybut et al. (1980). In the second place this limitation allows for an assumption of axisymmetric geometry without too great a loss in accuracy. In this case, ring-elements can be utilized (Figure 1b), and the three-di-

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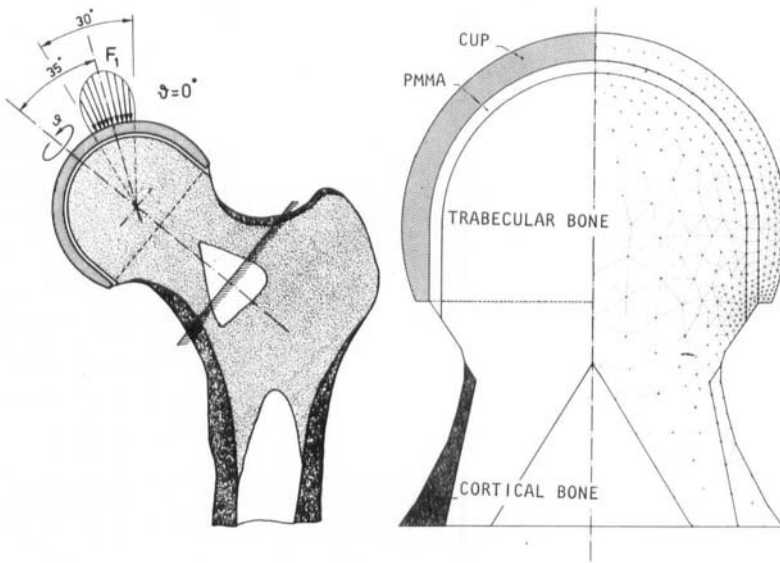


Figure 1. A. Sagittal section of a model for the proximal femur with a surface replacement according to Wagner (1978).

B. Finite element mesh for the model utilizing axisymmetric ring elements.

mensional solution can be obtained as a superposition of two-dimensional analyses, each of which represents a term of a Fourier series describing the complete three-dimensional external loading configuration. This approach saves computer time and space, and hence enables a much finer element mesh to be used, thus enhancing the mesh-accuracy.

We applied 646 6-node axisymmetric ring elements (linear stress distribution within an element cross-section). The three-dimensional loading and stress distribution were described by 11 Fourier terms, hence 11 subsequently superimposed 2-D solutions.

All stress results were normalized to a unit hip-joint force of 1 N. The separate materials were assumed to be linear elastic, homogenous and isotropic, and the material connections rigidly bonded. Young's modulus of cancellous bone was taken as 2,000 MPa, cortical bone as 20,000 MPa, acrylic cement as 2,000 MPa, and metal as 200,000 MPa. All Poisson's ratios were assumed to be 0.3.

Results

The external force F_1 (Figure 1a) applied as an ellipsoidally distributed load over a circular area of the cup resulted in a peak compressive stress of 1.6×10^{-2} MPa/N (0.016 Newton/mm² stress per Newton applied force) on the external cup (Figure 2).

Within the cup, the compressive external load was transformed to bending stresses, in

the same manner as a transverse load on a beam is transformed to bending stresses in the beam. Above the neutral plane with zero stress (n.p., analog to the neutral line in beam-bending) there was increasing compressive (bend-

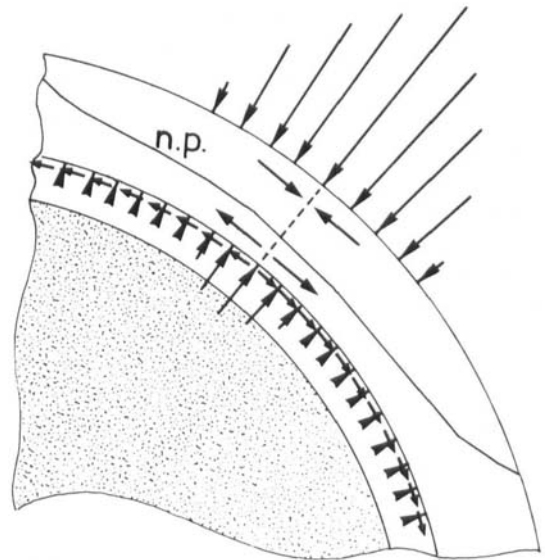


Figure 2. Qualitative illustration of the local load-transfer mechanism. The external compression on the cup is transformed into bending in the cup; the neutral plane (n.p.) with zero stress is indicated. Above this plane the cup is in compression, under this plane in tension. A part of the load is directly transferred to the cement foundation, engendering compressive and shear stresses at the cement-cup interface and at the cement-bone interface.

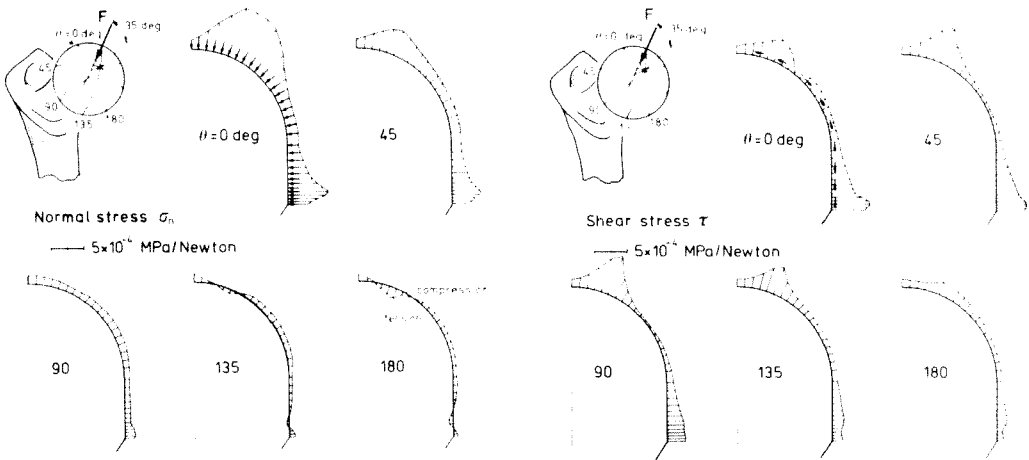


Figure 3. Stress patterns on the bone at the cement-bone interface in subsequent sections through the head, relative to a total hip-joint force of 1 Newton. The stress-orientations are parallel to the interface but otherwise arbitrary in this case; only in the sagittal plane ($\Theta = 0^\circ/180^\circ$) are they directed within the plane itself. Left. Normal, direct stress (compression and tension). Right. Shear stress.

ing) stress up to a peak level of approximately 3.7×10^{-2} MPa/N. Below this plane we found increasing tensile (bending) stress, up to a level of about 2.1×10^{-2} MPa/N.

Although the cup acts as an elastic buffer, part of the external load was transferred directly to the cement foundation (Figure 2), where a peak compressive (interface) stress of about 9.4×10^{-4} MPa/N occurred. A second, even higher compressive peak stress (15.2×10^{-4} MPa/N) on the cement, however, occurred on the lateral side, under the cup rim. The maximal shear stress at the cement-metal interface (12.7×10^{-4} MPa/N) occurred also in this region. The locations of these stress concentrations indicate that at least a part of the external load is transferred in that region, bypassing the central part of the head (stress shielding).

The stress state in an arbitrary point of the cement-bone interface can be represented by a normal, direct stress component (σ_n), which attempts to either compress (σ_n negative) or distract (σ_n positive) the two materials, and a shear stress component (τ) which attempts to shear the materials apart tangentially. The patterns of these stresses over the interface are illustrated in Figure 3. The patterns are shown in 5 sections from $\Theta = 0^\circ$ to $\Theta = 180^\circ$ only, since they are symmetric with respect to

the sagittal plane; hence the plane $\Theta = 90^\circ$ will be a mirror image of $\Theta = 270^\circ$, and so on.

The peak compressive stress (Figure 3) occurred directly under the external load ($\Theta = 0^\circ$), and reached about 7.5×10^{-4} MPa/N. A local peak also occurred in the lateral cup-rim region ($\Theta = 0^\circ$), with a value of about 6.2×10^{-4} MPa/N. Only very small tensile stresses were found.

The peak shear stresses (Figure 3) occurred in the superior cup-rim region, in the sagittal plane ($\Theta = 0^\circ$), and at the apex of the head in the coronal plane ($\Theta = 90^\circ$), amounting to about 5.5×10^{-4} MPa/N and 5.9×10^{-4} MPa/N, respectively.

The patterns of Figure 3 again indicate that a significant part of the external hip-load is transferred to the bone via the superior cup-rim region. Whereas in the normal hip the greatest part of the load is transferred directly through the head to the medial cortex (e.g. Brown et al. 1980), in this case the central part of the head is partly bypassed, and thus "stress-shielded".

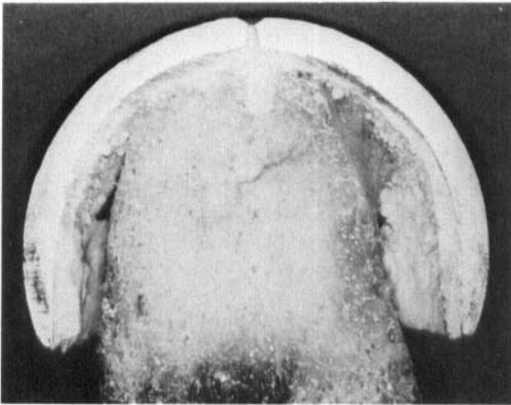


Figure 4. Typical example of a sagittal section through a removed femoral head with a surface replacement.

Discussion

It is important to appreciate that the *concept* of (Wagner) surface replacement was analysed, rather than a particular patient case. The model was an idealized representation of an actual structure, featuring several simplifications and assumptions imperative for the application of the technique. Hence, the stress values should be regarded in a relative, rather than in an absolute sense.

Comparable analyses of other surface replacements were reported by Shybut et al. (1980), Askew et al. (1984), and Schreiber & Jacob (1984). Generally speaking, the trends reported here agree with these studies. However, both the ICLH and the THARIES models exhibited higher stresses in the lateral/superior cup-rim region, higher tensile stresses at the medial/inferior interface, and more stress shielding in the central head region than found in our analysis of the Wagner cup. Although alternative modelling techniques and assumed loading conditions may play a role here too, it is probable that these differences are a result of the higher flexibility of the Wagner cup.

An example of a removed prosthesis is shown in Figure 4, typical for the 28 cases in our own material. Generally speaking, drastic bone resorption has occurred along the peripheral head-neck region, shown here at the medial and lateral sides. The central bone is usually well attached to the cement layer under

the dome of the cup; sometimes a very thin fibrous layer is present at the interface. The remaining bone is usually sclerotic on the medial side, and osteoporotic on the lateral side, suggesting massive remodelling in cancellous bone. Sometimes, but not always, collapse had followed the drastic reduction of the bonded interface area. An interesting detail is that the cement-layer usually remained well fixed to the metal. The common trends found in these cases make this type of arthroplasty ideally suited for analyses of failure mechanisms.

The stress patterns obtained here (Figure 3) were evidently not in agreement with the previously assumed physiologic nature of the stress transfer in joint-replacement structures of this kind (Wagner 1978). The compressive stress peaks at the superior/lateral interface, the tensile stresses at the inferior/medial interface, and the shear stresses over the whole interface are certainly not natural to the bone. In addition, when multiplied by a realistic hip-joint force magnitude, e.g. $3 \times$ body weight, their peak values indicate an uncomfortably low safety factor relative to the interface strength data reported in the literature. Hence, it seems attractive to explain the relatively high clinical failure rates reported in the literature by the occurrence of high, unnatural interface bone stresses in the initial post-operative stage and subsequent mechanical failure.

However, the interface stress magnitudes found in this case are not much higher than those reported in analyses of intramedullary fixed prostheses (e.g. Crowninshield et al. 1980, Huiskes 1980), or in surface-type fixations like the acetabular-cup (e.g. Carter et al. 1983) and the tibial component of the knee (e.g. Bartel et al. 1982, Lewis et al. 1982). Hence, if femoral surface replacements fail relatively early due to mechanical factors, these factors are probably not so much related to the initial, post-operative stress patterns.

It may be hypothesized, based on these findings, that failure initiation due to high initial stresses is not the significant denominator, but rather failure propagation due to a combination of mechanical effects (progressively increasing stresses and micro-motions through local failures) and biological phenomena (gradual interface bone resorption through micro-

motions). According to this hypothesis, the relatively early failure of surface replacements would be caused by high sensitivity to interface loosening or, in other words, by a lesser "secondary" stability when compared to other types of replacements. Prosthetic designs should perhaps be analysed relative to their potential to provoke failure propagation, rather than only initiation.

References

- Askew, M. J., Lewkowicz, S. Z., Lewis, J. L., Hanner, R. S. & Wixson, R. L. (1984) Three-dimensional analysis of the surface replacement hip prosthesis. *Transact. 30th Annual ORS*, Atlanta, GA, February 7-9, p. 249.
- Bartel, D. L., Burstein, A. H., Santavicca, E. A. & Insall, J. N. (1982) Performance of the tibial component in total knee replacement. *J. Bone Joint Surg.* **64-A**, 129-135.
- Brown, T. D., Way, M. E. & Ferguson, A. B. (1980) Stress transmission anomalies in femoral heads altered by aseptic necrosis. *J. Biomech.* **13**, 687-699.
- Carter, D. R., Vasu, R. & Harris, W. H. (1983) Periacetabular stress distribution after joint replacement with subchondral bone retention. *Acta Orthop. Scand.* **54**, 29-35.
- Crowninshield, R. D., Brand, R. A., Johnston, R. C. & Milroy, J. C. (1980) An analysis of femoral component stem design in total hip arthroplasty. *J. Bone Joint Surg.* **62-A**, 68-78.
- Huiskes, R. (1980) Some fundamental aspects of human joint replacement. *Acta Orthop. Scand. Suppl.* **185**, 109-200.
- Huiskes, R. & Chao, E. Y. S. (1983) A survey of finite element analysis in orthopaedic biomechanics. *J. Biomech.* **16**, 385-409.
- Huiskes, R. (1984) Principles and methods in solid biomechanics. In: *Functional behavior of orthopaedic materials*. Vol. I. *Fundamentals* (Eds. Ducheyne, P. & Hastings, G. pp. 51-98, CRC-Press, Boca Raton.
- Lewis, J. L., Askew, M. J. & Jaycox, D. P. (1982) A comparative evaluation of tibial component designs of total knee prostheses. *J. Bone Joint Surg.* **64-A**, 129-135.
- Oh, J., Bourne, R. B. & Harris, W. H. (1979) Strain distribution in the femur after surface replacement of the hip. *Transact. 25th Annual ORS*, San Francisco, CA, February 20-22, p. 35.
- Schreiber, A. & Jacob, H. A. C. (1984) Loosening of the femoral component of ICLH couple cup hip prosthesis. *Acta Orthop. Scand. Suppl.* **207**.
- Shybut, G. T., Askew, M. J., Hori, R. Y. & Stulberg, S. D. (1980) Theoretical and experimental studies of femoral stresses following surface replacement hip arthroplasty. In: *The Hip* (Ed. Riley, L. H.). pp. 192-224. C. V. Mosby, St. Louis.
- Wagner, H. (1978) Surface replacement arthroplasty of the hip. *Clin. Orthop. Rel. Res.* **134**, 102-130.
- Zienkiewicz, O. C. (1977) *The finite element method*, 3rd Ed. McGraw-Hill, London.