Finite element analysis of acetabular reconstruction : noncemented threaded cups

Acta Orthop. Scand. *58*, 620–625, 1987

Finite element analysis of acetabular reconstruction

Noncemented threaded cups

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The load-transfer mechanism and stress patterns in the acetabulum reconstructed with a threaded cup were investigated relative to the normal hip using the Finite Element Method (FEM). Several models were applied in the analysis to investigate the effects of axisymmetric 3-D versus 2-D approaches, inclusion of the femoral head, and local stress patterns around the threads in three types of implant/bone bonding modes. It was found that the threaded ring behaves as a relatively rigid implant shielding the trabecular bone and enhancing load transfer through the cortical shells. Load transfer from ring to bone is concentrated at the first and last threads where the subchondral bone layer is penetrated. Stress peaks of up to 24 MPa occur in these regions.

A type of cementless acetabular component increasingly applied is the metal threaded cup, often lined with a polyethylene socket (Mittelmeier 1976, Weill 1982, Lord & Bancel 1983). Due to its special features in comparison with conventional, cemented acetabular sockets, the threaded cup displays different mechanical behavior and bone reactions.

I have analyzed the load-transfer mechanism and the stress patterns in the acetabulum, reconstructed with threaded sockets, relative to the normal hip. For this study the finite element method (FEM) was applied (Huiskes & Chao 1983).

Methods

The axisymmetric FEM models applied are shown in Figure 1 as sections in the mid-frontal plane. The acetabular part of the natural hip is identical to the model of Pedersen et al. (1982) in geometric and material properties. However, where these authors applied the hip-joint load as a given constant external parameter, the present model also includes the femoral head, contact-coupled to the acetabulum. The advantage of this approach is that the contact stress distribution is determined as depending on the elastic characteristics of the acetabulum and the femur, rather than a priori assumed. To accommodate this to the numerical characteristics of the axisymmetric model, the resultant force exerted by the femur on the acetabulum must by necessity be prescribed along the axis of symmetry.

The threaded-cup reconstruction model was derived from the natural one using the geometric and material characteristics of the CLW cup (Weill 1982). This model was analyzed for several different boundary conditions at the head/socket connection: 1) a one-legged stance maximal force distributed over the inner socket (Crowninshield et al. 1978), 2) a symmetric load along the femoral neck axis, also distributed over the inner socket, and 3) a symmetric load exerted through the femoral head. In the latter case the femoral head was included in the model, as in the simulation of the natural hip. In the first case, Fourier expansion was applied to describe the 3-D, nonsymmetric load in 10 subsequent periodic terms. The applied load characteristics are in this case identical to those used by Pedersen et al. (1982). In addition, the same geometry was analyzed using

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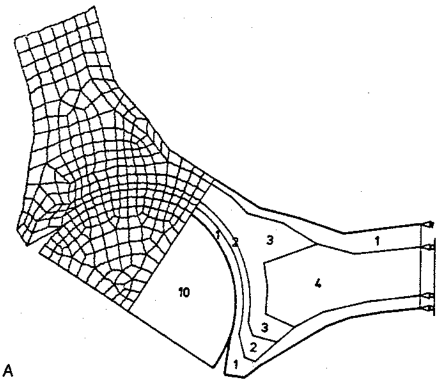

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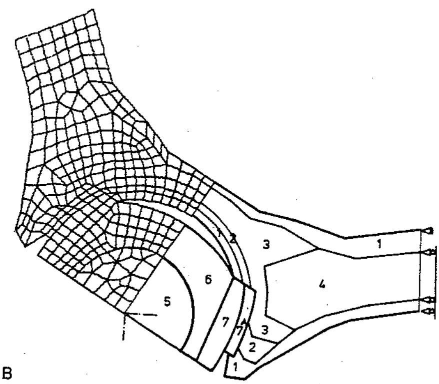

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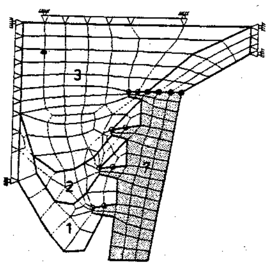

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Figure 1. Sections (mid-frontal plane) of the models used for the FEM analyses:
A. Natural acetabulum
B. CLW reconstruction
C. Local FE model of the screw-thread region
Partially bonded interface: all circled nodes.
Marginally bonded interface: open circled nodes only.
(See Table 1 for the explanation of the numbers.)

a 2-D, plane strain model to evaluate the effects of geometric simplifications.

The threaded-cup reconstruction model uses composite materials theory to describe the thread/bone region as a continuum. In order to determine the local stresses in and around the threads, an additional local 2-D FE model (plain strain) was developed. The local stresses are determined based on prescribed local boundary deformations derived from the global model in Figure 1B. The local analysis was carried out in three steps: a fully bonded implant/bone interface, a partially bonded interface, and a marginally bonded interface.

All the materials were described as linear elastic, homogeneous, and isotropic. Bone elastic constants were chosen identical to Pedersen et al. (1982). The cup liner is made out of ultra high molecular weight polyethylene (UHMWPE), the ring out of c.p. titanium, which, in a separate analysis, was also varied to CoCrMo alloy. Young's moduli values applied are summarized in Table 1.

The elements used were 8-node isoparametric quadrilaterals. The analyses were carried out with the MARC/MENTAT FEM and preprocessing and postprocessing codes (MARC Analysis Corporation, Palo Alto, CA, USA) running on the NAS-9060 main-frame computer of the University of Nijmegen.

Results

Typical stress patterns in the acetabular bone are shown in Figure 2. These mid-frontal sections of the model show equivalent von Mises' stress contours per unit applied force for two loading

Table 1. Materials and elastic moduli used in the FEM models.

No.	Material	E (MPa)	No.	Material	E (MPa)
1.	Cortical bone	1.7×10^4	5.	CoCrMo alloy	20.0×10^4
2.	Cancellous bone	0.3×10^4	6.	Polyethylene	0.07×10^4
3.	Cancellous bone	0.15×10^4	7.	C.p. titanium	11.0×10^4
			7A.	Composite	5.5×10^5
4.	Cancellous bone	0.1×10^4	10.	Cancellous bone	0.5×10^4

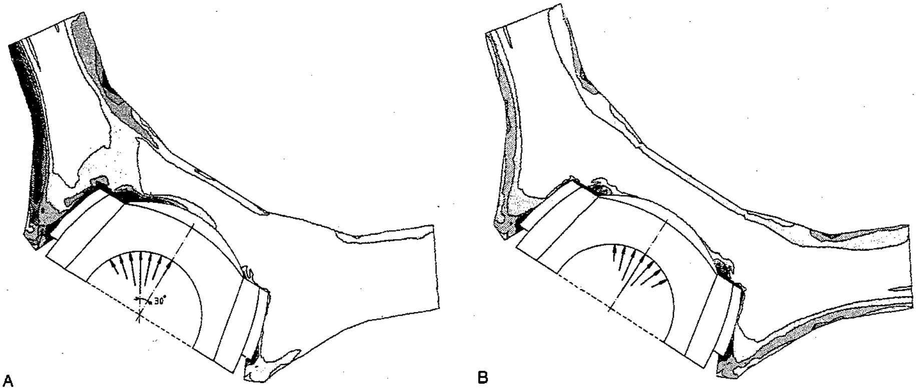

Figure 2. Von Mises' stress patterns per unit applied force in the bone, comparing a hip joint load according to the one-legged stance (a) to a symmetric load (b). ▭ $<2.5\times10^{-4}$MPa/N, ▭ 2.5–5.0×10^{-4}MPa/N, ▩ 5.0–7.5×10^{-4}MPa/N, ▩ 7.5–10×10^{-4}MPa/N, ■ $>10\times10^{-4}$MPa/N.

cases. Two important features of the load-transfer mechanism are visible in these graphs; first, the stress concentrations at the corners of the ring; and secondly, the importance of the cortical shells relative to the inner trabecular bone. Although the stress patterns are different for the two loads, the trends are equal in both cases. Rotation of the applied force from symmetric to 30 degrees lateral to the femoral neck axis results in an almost uniform twofold increase of stress in the lateral, superior part of the bone, and an almost 50 per cent reduction in the medial, inferior part.

The local von Mises' stress patterns in and around the threads of the ring are shown in Figure 3 for the case in which a partly connected implant/bone bond is assumed. The maximal stress peaks occurring at the upper and the lower corners are about 3.2×10^{-3}Mpa/N (symmetric force), which for a 4×B.W. hip-joint force (3,000 N) is 9.6 MPa. The maximal stress in the metal threads is about 16.0×10^{-3} MPa/N, hence 48 MPa for a 4×B.W. force. These values are somewhat lower in the (unrealistic) case that the metal is fully bonded to the bone, and about 25 per cent higher when the threads are only marginally fixed. Assuming a worst-case situation of 4×B.W. one-legged stance force and marginally fixed threads, local stress peaks of 24 MPa would occur locally in the subchondral bone layer near the superior and inferior edges of the ring. It is evident from Figure 3 that most of the load is transferred by the first and the last screw threads; the middle ones even seem to be almost redundant.

The stress patterns in the superior lateral part of the CLW-reconstructed acetabulum are compared with those in the natural bone in Figure 4. In this case the femoral heads were included in the models (symmetric force) because otherwise less realistic results would be obtained in the natural hip. The stresses in the reconstructed model are similar to the ones in Figure 2B. Relative to the natural bone, stress shielding of the inner superior trabecular bone is evident to an extent of about 50 per cent, and more load is transferred directly through the lateral cortical shell. The stresses in the lateral cortical shell are about 25 per cent higher and those in the superior medial shell about 50 per cent lower than in the natural case. In the medial shell over the apex of the acetabular dome, the dominant stress component, direct stress along the length of the shell, changes from tensile to compressive after reconstruction. The central subchondral bone layer, which has a very important mechanical function in the natural hip, is shielded from stress after reconstruction.

Discussion

In the present study axisymmetric FEM models were applied, comparable to the ones used by Pedersen et al. (1982) and Crowninshield et al.

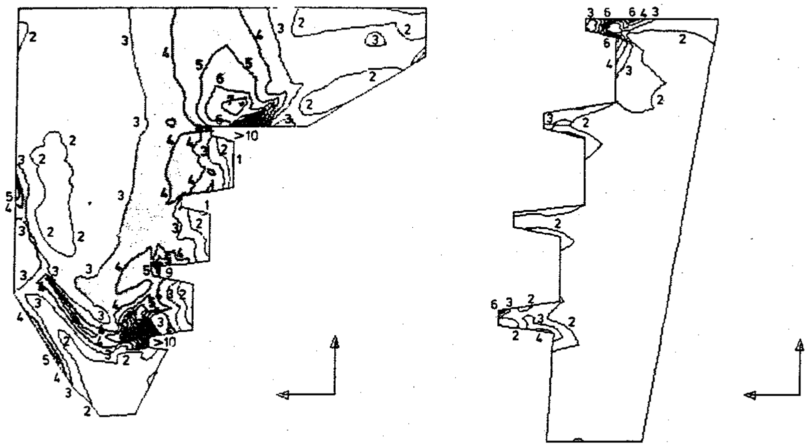

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Figure 3. Von Mises' stress patterns in and around the screw threads assuming a partly connected interface (circled nodes in Fig 1C).

Stress levels (MPa/N):

Bone:	1	0.0	6	17.8×10^{-4}
	2	3.5×10^{-4}	7	21.3×10^{-4}
	3	7.1×10^{-4}	8	25.0×10^{-4}
	4	10.7×10^{-4}	9	28.4×10^{-4}
	5	14.2×10^{-4}	10	32.0×10^{-4}
Metal:	1	0.0	4	12.0×10^{-3}
	2	4.0×10^{-3}	5	16.0×10^{-3}
	3	8.0×10^{-3}	6	20.0×10^{-3}

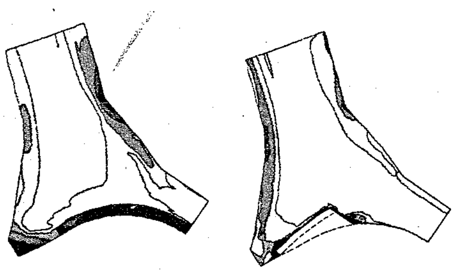

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Figure 4. Comparison of Von Mises' stress patterns in the natural acetabulum (A), and the CLW-reconstructed acetabulum (B). ▭ $<2.5 \times 10^{-4}$MPa/N, ▭ $2.5–5.0 \times 10^{-4}$MPa/N, ▨ $5.0–7.5 \times 10^{-4}$MPa/N, ▩ $7.5–10 \times 10^{-4}$MPa/N, ■ $>10 \times 10^{-4}$MPa/N.

(1983) for their studies of conventional, cemented acetabular reconstructions. However, some essential improvements and additional models were applied in an attempt to obtain reasonable complete and realistic information without having to rely on impractical 3-D FEM models (Huiskes & Chao 1983). One of these improvements was the inclusion of the femoral head in the acetabular model, contact coupled to the polyethylene liner, a method used earlier by Brown & DiGioia (1984) and by Rapperport et al. (1985) to analyze pressure patterns in the normal hip.

It is evident from the stress and deformation patterns obtained that the femoral prosthetic head presents an important deformation restraint to the inner socket surface. It forces the relatively soft polyethylene to conform to the globular shape of the head. As a result, the deformation of the polyethylene is much less than predicted by a model assuming direct load application as an independent variable. Thus, the metal head functions as a "stiffener" for the polyethylene socket. Generally speaking, this stiffening effect would also influence the stress patterns in the bone and at the implant/bone interface (Huiskes & Slooff 1986). However, it turns out that the metal ring is so stiff relative to the bone and the polyethylene that it behaves approximately as a rigid body, thus overshadowing the stiffening effect of the metal head. This effect is due to the structural, rather than the material stiffness of the ring. When the material is changed from c.p. titanium to CoCrMo alloy, which has an elastic modulus almost twice as high, there is virtually no change in the stress patterns in the acetabular bone and in the polyethylene liner.

The metal femoral head restraint to the deformation of the relatively flexible polyethylene cup also implies that the application of a presumed external load at the inner cup boundary in the model, as in previous analyses (Pedersen et al. 1982, Crowninshield et al. 1983, Vasu et al. 1982, Carter et al. 1982, Oonishi et al. 1983) results in considerable overestimations of the cup deformations and the bone stresses. However, in the case that a relatively stiff metal backing or ring is applied, the deviations are limited to the polyethylene liner.

Comparing the essentially three-dimensional, axisymmetric approach used here and in some previous analyses (Pedersen et al. 1982, Crowninshield et al. 1983) to the two-dimensional, plane strain approach as applied by Vasu et al. 1982, Carter et al. (1982) for the threaded cup reconstruction showed a qualitative agreement in stress patterns. However, whereas the stresses in the polyethylene liner are of comparable magnitudes, the bone stresses are overestimated approximately fourfold in the 2-D model. Nevertheless, the agreement found in general trends indicates that 2-D models can be adequate for stress evaluations on a comparative basis. The adequacy, however, is affected by the rigidity of the reconstruction. In analyses of flexible sockets, the 2-D approach is less suitable (Huiskes & Slooff 1986).

The results obtained concerning the load transfer in the natural and reconstructed acetabuli in general confirm earlier findings with respect to the relative importance of the cortical shells and subchondral plate (Jacob et al. 1976, Pedersen et al. 1982, Vasu et al. 1982). The acetabulum behaves as a sandwich construction in which the internal trabecular structure mainly serves to prevent the cortical shells from collapsing; the shells take most of the load. The present results also emphasize the structural significance of the relatively stiff acetabular rim, taking relatively high hoop stresses. Violation of this part in acetabular reconstruction weakens the structure as a whole.

The sandwich feature of the acetabulum is of paramount importance for the design and fixation technique of the socket. In particular when the dense subchondral layer is reamed, the structural rigidity of the acetabulum is considerably reduced. When a conventional, cemented polyethylene socket is then used, stress concentrations occur in the central, superior cement and bone region, which can only be reduced by the application of metal backing (Pedersen et al. 1982, Carter et al. 1982, Huiskes & Slooff 1986). When the subchondral layer is not reamed, cement/bone interlocking is less likely, which reduces the interface strength. The threaded cup concept is a way out of this dilemma.

Load transfer in the threaded-cup reconstructed acetabulum shows some typical general characteristics, dominated by the relative rigidity of the metal ring. Because the polyethylene liner is not backed in the central part, the load passes directly to the ring through the periphery of the liner. This mechanism creates a peripheral contact-stress contour in the joint, which will also affect the friction and wear mechanism. Because the liner tends to be pushed through the central hole in the ring, the long-term integrity of the polyethylene to metal connection is of vital importance. Although the maximal stress predicted in the polyethylene liner (12 MPa) is somewhat higher than is found in conventional cemented cups (Huiskes & Slooff 1986), it is still low in comparison to the yield strength of 20–40 MPa reported for this material (Willert & Semlitsch 1976).

The relative rigidity of the ring and the way in which it is screwed into the dense subchondral bone causes a load-transfer emphasizing the cortical structures, even more so than in the natural case, resulting in some stress protection of the central trabecular bone, understressing the medial cortex and overstressing the lateral cortex. The load transfer from the ring to the bone is concentrated mostly at the lower and upper corners, leading to stress concentrations, possibly up to 24 MPa. Although this is much higher than predicted for conventional cemented cups (Pedersen et al. 1982, Huiskes & Slooff 1986), these values occur only locally in the dense subchondral bone, and are only moderate in comparison with the reported yield strength values of about 80 MPa for this material (Vasu et al. 1982). Nevertheless, these results emphasize the need for a very precise reaming procedure to obtain an optimal intimate contact between ring and bone. In view of the high, local stress concentration, local bone yielding, remodeling, and densification could occur in the immediate environment of the ring.

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Acknowledgement

This project was sponsored in part by Protek, Bern, Switzerland.