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Stress analysis of threaded cups

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Abstract Using finite element analysis we have studied the pelvic bony socket and compared it with radiological imaging using threaded acetabular cups of three different shapes (parabolic, conical, hemispherical). The two-dimensional model depicted a planar section through a left pelvic hemisphere. In all three cups the stress in the bony socket increased from lateral towards medial. Compressive stress was found on the superior and inferior parts of the cup, but mainly on the superior aspect, seen radiologically as new trabecular bone formation. The maximum compressive stresses were seen in the cranial curvature of the conical cup, with less in the parabolic form and least in the hemispheric form. The tensile stress at the bottom of the socket increased from the hemispheric to the conical shape. Radiological rarefaction gave an indication of lower stress. There was lower compressive stress between the teeth of the threads. This FE model uses computer simulation to predict bony changes with different designs of implant. The ability to simulate biological conditions is a valuable addition to the testing of mechanical strength.

Résumé Par éléments finis nous avons étudié le cotyle osseux et l'avons comparé avec les images radiologiques faites avec des cupules vissées de trois formes différentes (parabolique, conique, hémisphérique). Le modèle à

deux dimensions a représenté une section plane à travers une hémisphère pelvienne gauche. Pour les trois cupules la contrainte dans l'os augmente de dehors en dedans réalisant un toit osseux en pince. La contrainte de compression existe sur les parties supérieure et inférieure de la cupule, mais d'une manière prédominante sur le côté supérieur, radiologiquement représentée par une nouvelle formation osseuse trabéculaire. Les contraintes de compression maximales étaient présentes dans la courbure polaire crâniale des cupules coniques, avec une tendance décroissante de la forme parabolique à la forme hémisphérique. Des contraintes en tension ont été démontrées au fond du cotyle osseux, en augmentant de la cupule hémisphérique à la cupule conique. La raréfaction osseuse radiologiquement visible a représenté la contrainte calculée et moindre de contrainte de compression entre les filets. Ce modèle utilise la simulation informatique pour prédire les changements de l'os selon les différents dessins des implants. La possibilité de simuler des conditions biologiques complète utilement les essais de résistance mécanique.

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Introduction

Uncemented hip arthroplasty using threaded cups [3, 4, 8] is a procedure with good medium and long-term results that can be equivalent to those with cemented techniques. Uncemented cup implantation is carried out with a press-fit technique. As soon as the cup is implanted, “pre-stress conditions” are produced [2], and this will occur both when a threaded cup is screwed in and when a slightly oversized unthreaded hemispherical cup is implanted. With both techniques, preparation of the bony socket causes loss of supportive bone, which has to be provided biologically.

The shape and rigidity of the implant causes changes in the stress configurations, which may be converted into hypertrophic or atrophic areas. Radiologically apparent new trabecular bone (Figs. 1, 2) occurs in the zones of implant loading, while atrophy is seen in the low loading

Fig. 1 Directed trabecular bone to the superior pole curvature and to the threads. Bone rarefaction between the threads in the superior part of the parabolic Hofer-Imhof cup, 7 years postoperatively

zones. The regions where the distribution of stress is physiological remain unchanged.

Microradiographs of threaded cups show these phenomena not only on the axis of load, but also in ventral and dorsal regions. After implantation of a threaded cup, newly formed bone is present at the threads and at the load-bearing surfaces (Figs. 3, 4). This is due to differing directions of loading by the patient when undertaking different activities such as walking, bending, climbing stairs and sitting. This induces a spatial distribution of stresses in both implant and surrounding bone.

The finite element (FE) technique is used to elucidate and simulate these biomechanical processes. Former FE models for analysis of the acetabulum represented only a section of it. Huiskes [5], Rappaport et al. [9] and Dalstra and Huiskes [2] have described models. Carter et al. [1] and Vasu et al. [12] expanded the model to include the ramus of the ilium. Levenston et al. [7] also enlarged it medially and cranially into the ilium, and medially and caudally into the superior ramus of the pubis. In 1996 Witzel [13] expanded the Carter-Vasu model [1, 12] to include the pubis and ischium, bringing a three dimensional configuration into a two-dimensional plane.

Fig. 2 Newly formed trabecular bone formation in direction of the load on the bottom of the Zweymüller threaded cup, 6 years post-operatively

In general as the calculated section decreases, the rigidity of the model increases, and the stress pathways to be calculated are reduced. Limited information is obtained when only considering one plane. Vasu et al. [12] and Huiskes [5] distinguished between cortical and cancellous structures.

As the physiological reaction of bone is determined by local mechanical stresses, comprehensive stress analysis of both the implant and the bony socket is necessary when investigating osseointegration. One modern computational approach is the finite element method (FEM), by which both two-dimensional [10, 11] and three-dimensional models can be subdivided into finite elements to which measurements of bony activity at the site of implantation can be given parameters. This study compares the results of FE calculations of three different threaded cup shapes and their bony sockets, with those of the radiological findings.

Materials and methods

A cross-sectional model through the entire left hemipelvis was chosen for this study, and the model contained 1,966 finite elements with 3,846 nodes. Loading through the hip joint corresponded to the standing phase during walking. Compensatory forces passed through the ilium, ischium and lumbosacral spine, and locally at the cranial edge of the acetabulum as a further modification of Vasu's model [11].

Fig. 3 Microradiography of the parabolic cup (Hofer-Imhof) shows trabecular bone in the curvature of pole and on the threads. The cup has a titanium lid on a polyethylene insert

Fig. 4 Microradiography of the conical cup (Zweymüller) shows trabecular bone in the curvature of pole and on the threads. The closed bottom of the cup also enables new bone to form in this region

It was possible to determine displacement at every nodal point and stresses in each finite element using the computation software ANSYS 5.5.3 (ANSYS, Canonsburg, Pa., USA). The calculations were performed using a Power Challenge from Silicon Graphics at the Computer Centre of Ruhr University in Bochum, Germany. The relative movements were visualised by superimposing the contours of the non-deformed structure, or on the basis of gap formation. For greater clarity, the gap width was increased using a scale factor. Simplification of the geometry of the input model and the use of relatively undifferentiated material values had to be accepted. Bone and implant material were assumed to be homogeneous, isotropic and of linear elasticity. For two-dimensional calculations, loading of the model was carried out in such a way that the resulting stresses were approximately $\pm 1 \text{ N/mm}^2$. The presentation of results included comparative stresses, compressive or tensile, or a vector field. In the stress diagrams particular areas were identified and colour-coded for rapid and reliable recognition. Compressive stresses were labelled red when indicating lack of stress, orange and yellow for low, green for moderate, and blue and grey for high stress. The colour coding was reversed when considering tensile stress, except for grey, which again indicated maximum stress. The ranges of measurable stress could be selected and extended if necessary for more accurate analysis. The values of FE calculations were related to radiographs, microradiographs and histological sections. Every threaded cup calculation included direct element coupling of the implant and bony socket to simulate the incorporated status, and also contact elements [9, 14] intended to simulate the non-incorporated, i.e. immediate post-

operative state. The model allowed direct transfer of compressive forces and of thrust forces with selected coefficients of friction at the interfaces, but not the transfer of tensile forces. The femoral head/polyethylene articulation was of a sliding design through contact elements in all calculations. The calculated stresses were compared with each other; there were no absolute values, since this was a two-dimensional model.

The sacroiliac joint was considered a firm bearing area, and the iliac spine a mobile area. Imperfect compressive stress peaks (artefacts) were found in the ischium. These forces were compensated for in vivo by increased bone. The centre of the femoral head shown as a triangle marked the input of distributed load and therefore reduced the deformation that would result from a single loading point.

Three threaded cup models were used for the FE calculation and the analysis of the cups in their bony sockets.

The *parabolic threaded cup* had 12 rows of teeth, four thread turns, a pitch of 5 mm and a thread depth of 2.8 mm. The 28-mm polyethylene insert had a titanium lid to close the shell at the pole opening. Since the lid was not in contact with the shell, it was not included in the calculation, but direct contact between bone and polyethylene was simulated. The peripheral two-thirds of the cup had contact with the cortical bone; the central one-third had contact with cancellous bone. The cup was positioned medially so that the outer diameter was the same as for the spherical cup (Fig. 5).

The *conical threaded cup* had 12 rows of teeth, four thread turns, a pitch of 10 (5) mm and a depth of 3 mm. The cup pole was closed except for an M6 thread aperture. In order to allow

Fig. 5 Finite element analysis simulating the immediate post-operative situation with compressive stresses running from the superior and inferior pole to the sacroiliac joint. Superiorly a stress gradient increased from lateral to medial. Stress-free regions are present at the bottom, opposite to the polar opening and the inferior cancellous bony area

modelling with the same cup size and a cortical bone contact of about 50%, the cup was placed under the bony margin.

The *hemispherical threaded cup* had an outer diameter of 54 mm with nine rows of teeth, five thread turns, a pitch of 4 mm and a thread depth of 2.6 mm. The thread turns were embedded completely within cortical bone.

The hemispherical and parabolic cups were single-threaded, the conical cup double-threaded. The surface of all threaded cups was corundum-blasted with a roughness of 3–5 μm . Pre-stresses in cortical bone which could result in loading of the lateral thread surface were disregarded in the model, since their computational definition is uncertain [2]. The threads were counted starting from the pole.

Results

Parabolic cup

Induction of the hip joint force took place along a pathway from the superior and inferior pole curvatures to the sacroiliac joint, with different accentuation. Further pathways ran cranially and caudally with lower compressive stress (orange area). The polyethylene in the cranial area showed marked compressive stresses (green area). Stress-free regions were seen centrally adjacent to the polar opening, and in the caudal aspect of the polyethylene inlay (red area; Fig. 5). High tensile stress, however, occurred centrally adjacent to the polar opening and in

the central polyethylene. The cranial pressure gradient increased from lateral to medial. There were high compressive stresses in the polyethylene running in a cranial direction. There was a notable peak of pressure cranially between the polar surface and the most medial thread. Further peaks of compressive stress, decreasing laterally, were also seen at the teeth of the second and third thread. Stress-free areas were present in the cortical bone at the lateral thread surfaces of the second and third threads (Fig. 6). All the teeth showed torsion laterally, seen in the tensile stress analysis with high stresses in the medial threads. There was caudal separation of the head due to deformation of the polyethylene. The cup separated more markedly at the bottom than laterally, which is equivalent to tilting it medially (Fig. 7). The bony socket of the thread surfaces on the medial side had high compressive stress, especially at the most medial and lateral thread.

Conical cup

The immediate postoperative situation of the conical cup model was the same as that of the parabolic cup. The principal stress pathway, which extended broadly from the craniolateral margin through the cranial polar curvature to the polar opening, ran to the sacroiliac joint. An additional concentrated area of stress ran from

Fig. 6 Parabolic cup: superiorly the pressure gradient increases from lateral to medial. A pressure peak is located at the curvature of the pole and the first thread. Further stress peaks decrease towards the lateral

Fig. 7 Separation of the parabolic cup more medially than laterally. High compressive stress at the most medial and lateral thread. Separation of the head due to polyethylene deformation (scale factor)

Fig. 8 The bottom of the conical cup is under tensile stress; there is no compressive force

the lower polar curvature to this site. Further low compressive stress pathways ran to the ilium and ischium. The stress distribution within the polyethylene was comparable to that with the parabolic cup. There was no compressive stress in the caudal polar area. The compressive stress in the polyethylene inlay passed from the cranial polar curvature to the bony socket. There was no transmission of forces caudally through the polyethylene, but to a minor degree through the cup close to the inferior polar curvature. There was tensile stress but no compressive force at the bottom of the cup (Fig. 8). Superiorly, high compressive stress values were found in relation to the tips of the threads, and the polar curvature, increasing from lateral to central. The compressive stress extended from the threads to the shell. No stress-free areas were seen in the cranial region. In the caudal position the head of the femur component was detached from the polyethylene insert. The bone between the teeth of the threads was consistently free from compressive stress. There were laterally directed compressive stress-free zones in relation to the teeth of the threads. The thread closest to the central position and the surrounding area was involved in the effect of the caudal polar curvature from which the transmission of compressive stress originated.

Hemispherical cup

The compressive stresses ran essentially from the cranial and caudal polar curvatures to the sacroiliac joint. Further compressive stress pathways of less significance ran cranially and caudally. Stress-free regions were present in the caudal cancellous bone, in the area adjacent to the polar opening and in the caudal part of the polyethylene inlay. In contrast, there were marked compressive stresses within the polyethylene cranially. The medial surfaces of the cortical bone showed high compressive stresses, which decreased laterally. The cortical bone adjacent to the lateral thread showed stress shielding with an area free of compressive forces. Tensile stress analysis revealed lateral torsion for each tooth. In the caudal position there was separation of the head from the polyethylene and loosening of the cup from the cortical socket. Due to this loading the cup may tilt laterally, as seen by the pronounced lateral separation of the cup. Between the teeth of the threads the cortical bone close to the cup showed no compressive stress. On each lateral side of the threads, a press-fit remained, causing a polar bending stress on the thread turns. The cortical bone between the caudal threads was not subject to tensile stresses.

Discussion

Previous FE calculations for threaded cups have yielded theoretical results without clinical correlation. In this study, the results of the FE calculation were compared with the radiological findings. Compared to the hemispherical cup where a compressive stress pathway runs both from the cranial and caudal polar curvatures to the sacral iliac joins, a predominantly caudal path is seen with the parabolic and conical forms.

The attenuated compressive stress profile running to the ilium and ischium was similar for all three cups. The background conditions relating to modelling of the socket and induction of forces was also similar. There is a large concentration of compressive strength at the cranial curvature in the conical cup, less so with the parabolic cup and least with the hemispherical cup. With the conical cup this stress concentration ran cranially as far as the centre of the cup, with the metal floor acting as a force transmission element. In conformity with Wolff's law of transformation [15], the stress concentrations induced new bone formation, as seen radiographically up to the polar opening with the hemispherical and parabolic cups (Figs. 1, 3). With the conical form, which does not have a polar opening, the new bone formation extended closer to the centre of the cup (Figs. 2, 4). The bone stock around the hemispherical and parabolic cups (Fig. 5) was largely free from compressive stress, whereas the bone around the conical cup was only free from compressive stress caudally. All cups were subject to tensile stresses in this region (Fig. 8). These tensile stresses were much less with the conical form and were no longer apparent in the cranial region. The resultant predominant compressive stress induced newly directed trabecular support (Figs. 1, 2, 3, 4). The compressive stresses in relation to the tips of the threads increased from lateral to medial in all three forms; they were lowest with the hemispherical and largest with the conical shapes. Due to the compressive stresses applied at the tips of the threads "forceps-like" bony cover and new trabecular bone formation occurred at the pole curvature (Figs. 3, 4). The compressive stresses were highest in the conical shape, lower in the parabolic form and lowest in the hemispherical form. The directed bone structure was radiographically identifiable (Fig. 2). Radiographic rarefaction was observed as an expression of reduced and absent compressive stress between the teeth of the threads (Figs. 1, 2, 3, 4). The largest compressive stresses were present in the immediate region of the cranial pole curvature of the conical form. The value was decreased with the parabolic cup and smallest with the hemispherical form. The most clearly visible new bone formation was radiographically seen in this region (Figs. 1, 2, 3, 4). The calculations in the caudal region between the femoral head and PE inlay revealed gap formation as the expression of load-induced femoral head displacement. The rigid head assumed an eccentric position in the soft PE socket and this was further exacerbated by wear during use.

The relative movement of the caudal half of the cup in the simulation of immediate postoperative conditions can be used to predict later behaviour. The hemispherical cup separated more markedly at the lateral margin and was in contact in the polar direction. The parabolic form delaminated centrally and was in contact laterally. This could mean that in a left hemipelvis, upon failure of the bony socket or in the absence of contact the hemispherical cup might undergo counter-clockwise tilting, i.e. would become vertically orientated. Conversely, the parabolic form would become reorientated to the horizontal as a result of clockwise rotation. Compared with the hemispherical and parabolic form, the conical form was detached parallel to the bony socket.

Tensile stress is demonstrated on the bony socket at the bottom of the cup, and is largest in the conical shape, and smallest in the hemispherical shape. The polar curvature in the superior part of the cup is the point at which compressive stress changes to tensile stress. Comparison with the radiographs and the FE analysis show the importance of the immediate postoperative stress situation.

This FE model uses computational simulation in order to predict bony changes with different implant designs. As a procedure which simulates biological conditions it is a valuable addition to mechanical strength testing methods. This technique should thus be incorporated in the quality assurance process for implant development.

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