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THE APPLICATION OF NUMERICAL SHAPE OPTIMIZATION TO ARTIFICIAL-JOINT DESIGN

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INTRODUCTION

Artificial joints have two distinct functions. Firstly, to provide for an adequate range of motion, and secondly, to tranfer the joint forces to the bone without causing failure. The chances for failure to occur depend on the strength of the implant/bone connection, and on the stresses associated with the load-transfer mechanism. In this paper the feasibilities of minimizing these stresses, thereby reducing the chances for failure, are discussed.

Our attention will focus on the femoral component of Total Hip Arthroplasty (THA), made out of metal, and fixated in the bone with the use of acrylic cement (Fig. 1). The stress patterns in this structure will depend exclusively on four aspects, (i) the magnitudes and orientations of the external loads, (ii)

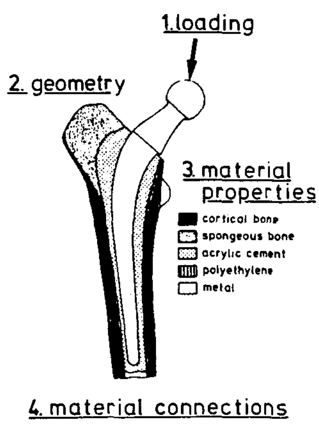

Fig. 1: The stress patterns in the bone/prosthesis structure depend on four aspects, as indicated in the figure.

the geometry, (iii) the properties of the materials, and (iv) the boundary conditions, including those of the connections between the separate substructures (interfaces). This implies that, given a particular loading pattern dependent on patient weight and activity, given the geometry and properties of the bone, and given the fact that acrylic cement (PPMA) is used for fixation, the stress patterns depend exclusively on the material properties and the shape of the prosthetic stem. In order to be able to determine the optimal stem shape, for which minimal interface stresses would be obtained, a computational method called 'numerical shape optimization' (NSO) was developed, to be used in combination with the finite element method (Huiskes and Boeklagen, 1988, in press). The application of this method in a model of femoral THA is discussed in the next pages.

FEM MODEL

The Finite Element Method (FEM) is pre-eminently suited to interrelate the structural parameters identified above (Fig. 1), and to determine stress patterns for particular loads. To develop an adequate model, however, is not an easy task, in view of the complex and often uncertain characteristics of loading, geometry, material properties and interface conditions (Huiskes and Chao, 1983). Rather than investigating these modeling aspect here again, we will postulate the model, and discuss its limitations later.

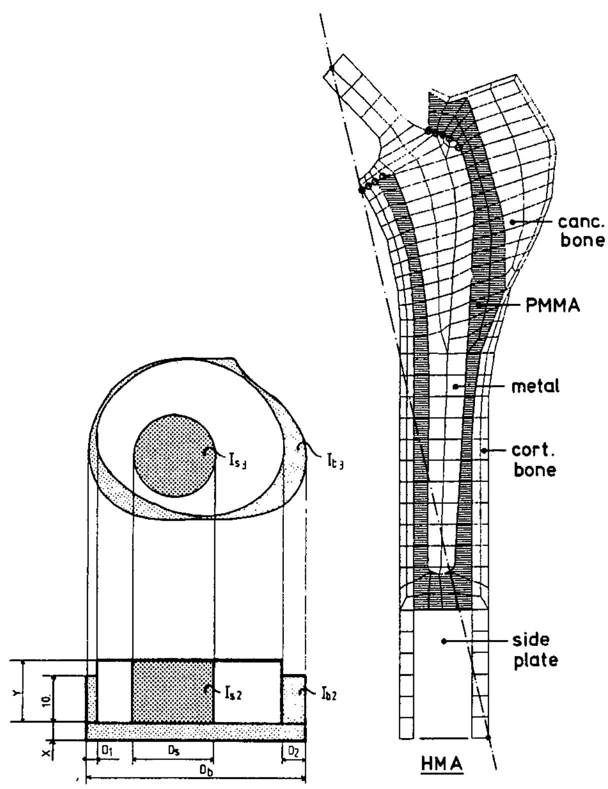

Fig. 2: Example of a 2-D side-plate model of the femoral THA, as used in the FEM analyses. The method of deriving the element thicknesses from a 3-D cross section is illustrated.

The FEM model is 2-D, using a side-plate and nonuniform element thicknesses to take the out-of-plane structural rigidity into account (Fig. 2). Hence, the model consists of two 2-D FE layers superimposed, of which one is the 'side plate', representing the out-of-plane cortex connection, and the other the 'front plate', representing the composite of stem, cement and bone in a mid-frontal section. The thicknesses of the bone elements in both layers are derived from the geometry of a cadaver bone (Huiskes et al., 1981), in such a way that the cortical thicknesses, the cross-sectional areas and the second moments of inertia are reproduced by approximation. A similar derivation is used for the stem elements. The cement elements are equal in thickness to the adjacent stem elements; the cement is not represented in the side plate.

The stem in this particular model of Fig. 2 is made out of CoCrMd and represents the Precision Hip (Howmedica Inc., Rutherford, New Jersey) (Noble et al., 1988); it does have a collar, but this is not modeled as connected to the calcar bone as a rule. Interfaces are asssumed to be rigidly connected, apart from a small region of the stem/cement interface at the proximal/lateral side

(Fig. 2). Materials are assumed to behave linear elastic, homogeneous and isotropic. Elastic moduli are taken as $2.0 * 10^5$ (metal), $2.0 * 10^3$ (acrylic cement), $1.7 * 10^4$ (cortical bone), $0.5 * 10^4$ (methaphyseal cortex) and $1.0 * 10^3$ MPa (cancellous bone). The elements used are plane-stress, 4-node quadrilaterals with a bilinear displacement field. The MARC FEM package (MARC Analysis Corporation, Palo Alto, CA) is used for the calculations.

This model is routinely used to assess the load-transfer mechanism in commercial prosthetic designs on a comparative basis, hence a model for 'numerical bench testing' (Huiskes and Vroemen, 1985; Huiskes, in press). Apart from the Precision Hip, five other prostheses were studied up to this time. Three loading cases on the femoral head are taken into account in these comparative analyses, one (unit) hip-joint force F_1 along the line shown in Fig. 2, a (unit) force F_2 at the same point along a line rotated 10 degrees laterally (clockwise about the point of application), and a (unit) pure bending moment M.

The stresses calculated in the model are represented as the values of the three components (two direct and a shear stress) along six proximal-to-distal lines in the structure: medial outside bone surface, medial cement/bone interface, medial stem/cement interface, and the corresponding ones on the lateral sides. For this purpose the stress tensor is expressed in each nodal point relative to a local basis, which is locally parallel to the line concerned. It must be noted that at the free surfaces no normal stress should occur, and that at the interfaces the normal and shear stresses should be continuous, and the parallel ('bending') stress discontinuous. The actual continuity of normal and shear stresses at the interfaces, however, is seldom fulfilled exactly, due to the approximate character of the FEM, and depending on the mesh density and element charateristics. The better approximation is always delivered by the nodal stress values calculated from elements of the material with the lowest modulus (Angelides et al., 1988).

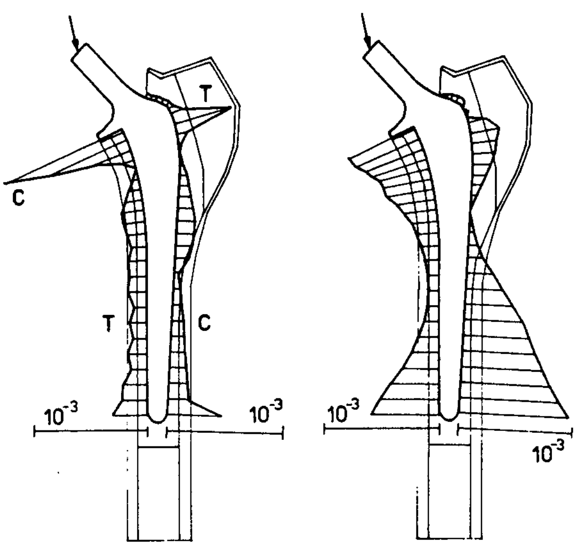

Fig. 3: An example of results for the Precision Hip: normal stresses (left; T=tension, C=compression) and shear stresses (right) at the stem/cement interface. Note the typical stress concentrations at the proximal and distal sides.

Fig. 3 shows an example of results: normal (tension/compression) and shear stresses along the medial and lateral stem/cement interfaces (loading case F_1), as found for the Precision Hip. These interface stresses give a good impression of the load-transfer mechanism, characterized by stress peaks at the proximal and distal sides. Similar patterns are found for the cement/bone interface stresses, and are typical for load-transfer in cemented stems (Huiskes, 1980). But although similar in these cases, the actual stress values depend on the shape of the stem, hence are quite different in a series of prosthetic designs.

Using the above method of 'numerical bench testing', prosthetic designs can be tested and compared relative to their chances to provoke failure. If necessary, parametric analyses can be carried out to investigate the consequences of design adaptations. Any desired material or geometrical parameter, or set of parameters, can be changed easily to investigate their effects on the load-transfer mechanism and the associated stress patterns. Such a process of parametric analysis is quite usefull as a basis for a design process, if only to develop some feeling for the importance of the various structural aspects. However, it is by necessity an arbitrary process of 'one shot at a time', and does not quarantee that the best solution is found.

Going back to the shear-stress patterns of Fig. 3, for example, it is obvious that optimal distributions would be uniform ones, lacking the high peaks on the proximal and distal sides. The question is if, and for which stem shape this could be realized. In order to answer that question, a procedure for Numerical Shape Optimization (NSO) can be applied (Huiskes and Boeklagen, 1988, in press).

NUMERICAL SHAPE OPTIMIZATION

The NSO program package consists of a number of procedures built around the MARC FEM program, illustrated in Fig. 4. Central to the method is firstly the definition of n design variables v_i enabling an adequately variable shape variation of the part to be optimized. These variables are denoted as $\underline{v} = (v_i)^T$, $1 \leq i \leq n$. Secondly, an objective fucntion F is defined, depending on particular stress, strain and/or strain-energy density (SED) values, calculated with the FEM. Because these values depend on the design variables, given specific loads, material properties and nonvariable geometric parameters, the objective function can be written as $F=F(\underline{v})$. Thirdly, m geometric constraints are introduced, expressing, for example, the limitation of the stem thickness to the width of the medullary canal, denoted as $a_j \leq f_j(\underline{v}) \leq b_j$, $1 \leq j \leq m$.

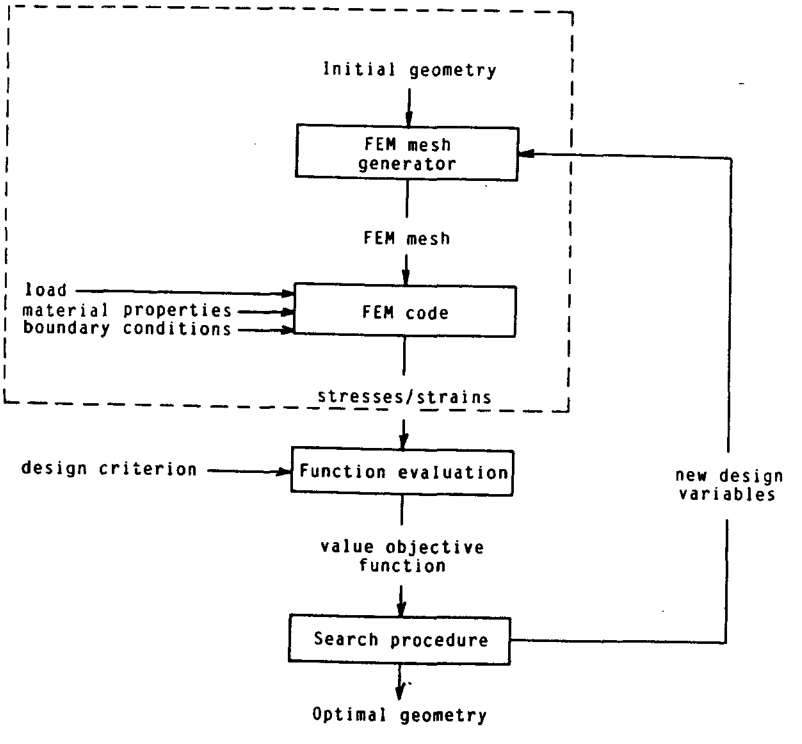

Fig. 4: Scheme of the NSO procedure. The functions within the dotted boundary are provided by the MARC/MENTAT FEM system.

The search procedure (Fig. 4) iteratively minimizes

$$\{ F(\underline{v}) \mid a_j \leq f_j(\underline{v}) \leq b_j\} \quad , \qquad (1)$$

to find the optimal set of design variables, for which the objective function has its minimal value. Several iterative algorithms for the search procedure were tested. A slope-following method called 'least-pth' (Daniels, 1978) was found to work satisfactory. The search direction was not determined analytically (Yang et al., 1984), but by FEM iteration. According to this method, the objective function is expressed in the form

$$F(\underline{v}) = \sum_{k=1}^{\beta} \{w_k e_k(\underline{v})\}^p , \qquad (2)$$

where e_k is a stress, strain, or SED function, w_k a weight factor, and the summation is over β relevant nodal or integration points.

The principles of the procedure are discussed in some more detail by Huiskes and Boeklagen (in press). It was first applied in a simplified, 1-D FEM model of an axissymmetric, straight bone, and later in a more realistic 2-D side-plate FEM model such as discussed earlier. In both cases, the stem was assumed to be symmetric relative to a fixed mid-stem axis, and 9 thickness parameters t_i along the stem were chosen to describe the stem shape. The fixed axis in the 2-D model was, of course, curved. The thicknesses t_i were transformed into arctangent functions and normalized, according to $v_i = \arctan(\pi t_i/2D_i)$, whereby D_i is the medullary width associated with t_i. In this way the geometric constraints were incorporated in the design variables v_i. In the 1-D model, the objective of uniform interface stresses was chosen for the design criterion on which the objective function was based. In the 2-D model, the objective function was based on minimal SED in the cement along the cement/bone interface, using the SED-value $U_k(\underline{v})$ in the appropriate integration points as a substitute for $e_k(\underline{v})$ in equation (2).

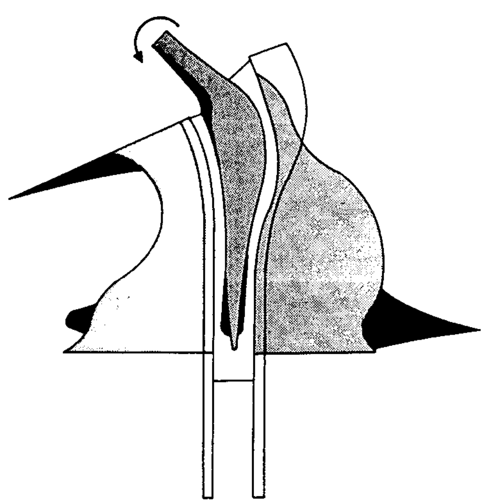

Fig. 5: Schematic representation of an earlier optimization (Huiskes and Boeklagen, in press), carried out for a pure bending load and minimal SED in the cement at the interface as design criterion. Shown are the final shape and its SED distribution (light) in comparison to those of the Precision Hip (dark).

The method showed to work satisfactory, and interface stress reductions of 30-70% could be obtained in both models. Results of the 2-D model are summarized in Fig. 5, showing the final, optimized shape of the stem and its associated SED distribution, in comparison to those of the Precision Hip discussed earlier. Typical features of this optimized shape, also found in the 1-D model, are its proximal taper, its belly-shaped middle region and its sharp distal taper. These features are quite effective in diminishing the stress peaks at the proximal and distal regions, associated with the load-transfer mechanism in conventional stem designs.

In a parametric analysis (1-D model) it was established (Huiskes and Boeklagen, in press) that the actual optimal dimensions of the stem are rather sensitive to the stem length and the bone modulus, and to a lesser extent to the stem and cement moduli. For most variations of these parameters an equally effective optimal shape could be determined. However, a stem length of 120 mm yielded lower stresses than lengths of 80 and 160 mm, and a stem modulus of $1.1 * 10^5$ MPa (titanium) generated up to 90 percent higher interface stresses with its optimized shape as compared to a modulus of $2.0 * 10^5$ MPa (CoCrMd). Hence, the latter material is much better suited to be used in stems designed for reduced interface stress transfer.

In the present analysis, using the 2-D side-plate model discussed earlier, a slightly different course is followed. The stem is assumed to be non-symmetric, and both the stem thickness and its position in the medullary canal are used as design variables. The geometric constraints per cross-sectional level for this case are illustrated in Fig. 6. The FE mesh, similar to the one in Fig. 2, defines 21 cross-sections along element boundaries, each containing nodal points on the medial and lateral interfaces. In each of these sections, the thickness of the medial cement mantle and the thickness of the stem are used as design variables. Hence, 42 design variables are used in total. The admissible space is further reduced by requiring a minimal cement-layer thickness of 1 mm. The arctangent transformation indicated above can not be used, and instead a "Boundary Follow" method is introduced, whereby an inadmissible set of parameters is corrected relative to the boundaries of the admissible space, as illustrated in Fig. 7.

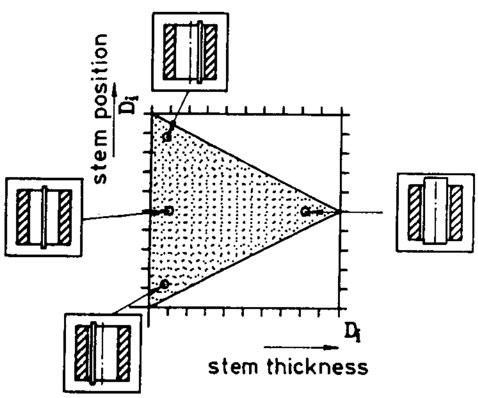

Fig. 6: Schematic representation of the admissible design space in a section as a function of two local design variables: thickness of the medial cement mantle and stem thickness, both bounded by the medullary width D_i.

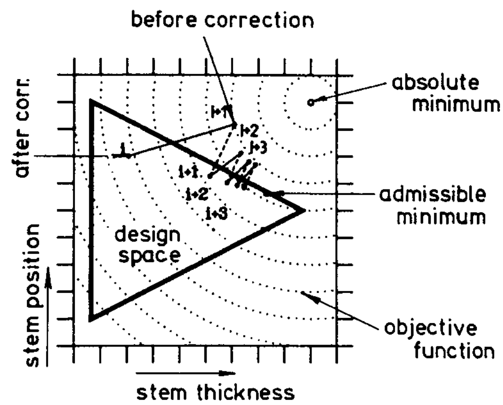

Fig. 7: Illustration of the boundary-follow method used in relation with the search procedure.

The design criterion used this time accounts for individual stress components (normal and shear stresses) in six regions of the cement/bone interface; proximal, middle and distal ones at the medial and lateral sides. The objective function takes the form

$$F = \sum_{i=1}^{\beta} \{ W_{\tau i}(\tau_i - \tau_{io})^P + W_{\sigma i}(\sigma_i - \sigma_{io})^P + W_{di}D_i^P \} , \qquad (3)$$

where β is the number of all cement nodal points at the cement/bone interface, τ_i and σ_i the shear and normal stresses in these points, τ_{io} and σ_{io} associated reference values, D_i geometric parameters to penalize the creation of irregular stem contours, as discussed later, and $W_{\tau i}$, $W_{\sigma i}$ and W_{di} associated weight factors.

The values of $W_{\tau i}$, $W_{\sigma i}$, τ_{io} and σ_{io} were varied per region, and also changed between iterations, using the stress values obtained as guidelines. Hence, the process was conducted this time as an empirical one, with manual interference between iterations, guiding the process towards the shape providing the most desirable, uniform interface stress patterns. In the final iterations, the values shown in Table I were used.

	medial			lateral		
	proximal	middle	distal	proximal	middle	distal
$W_{\tau i}$	1.0	1.0	1.0	0.5	0.8	0.8
τ_{io}	$-4*10^{-4}$	$-4*10^{-4}$	$-4*10^{-4}$	$4*10^{-4}$	$4*10^{-4}$	$4*10^{-4}$
$W_{\sigma i}$	0.8-3.7	1.0	0.4-1.0	1.0	1.0	1.0
σ_{io}	$-4*10^{-4}$	0	0	$-1.8*10^{-4}$	$-1.8*10^{-4}$	$-1.8*10^{-4}$

Table I

This implies that uniform shear stresses were eventually strived for, somewhat emphasizing the medial side in favor to the lateral one. Tensile stresses, predominantly occuring in the medial/distal and lateral/proximal regions were penalized in particular, as were the medial/proximal compressive stress peaks. All these values were estimated empirically in the course of the process. The value p=4 was used predominantly for the exponent. The 'reference' force F_1, discussed earlier, was used for the external load during the entire iterative process.

RESULTS

The resulting shape found with the NSO procedure and its associated interface normal and shear stresses are shown in Fig. 8. Characteristic features are the constriction in the stem contour on the proximal/medial side, causing a wedge-shaped cement mantle, a belly-shaped middle region, and a medialized, strongly tapered distal tip. Although different in details, these general features are very similar to those of the optimized shapes obtained in the earlier analyses with the 1-D and 2-D models (Huiskes and Boeklagen, in press).

There was a strong trend for the stem to form a constriction, or bottle neck, below the belly of the stem, thereby almost cutting it in two parts. This was also observed in earlier analyses, particularly those associated with a transverse force as external load on the femoral head. Some indication of this behavior is still visible in the shape of Fig. 8. It is believed that this mechanism is an effect of the requirement to reduce distal stresses due to the bending moment transferred through the stem. This bending moment is greatly reduced by a stem constriction, which would act somewhat similar to a 'hinge'. This would imply that the optimal stem length in this construction would be much less than presently assumed. Nevertheless, for practical reasons a shorter stem was considered inadmissible. Hence, the constricting mechanism was suppressed by adding the terms D_i to the objective function. These terms describe, in each nodal point i of the stem contour the 3° derivative of a spline function, fitted through a row of four nodal points, thus penalizing irregular shapes.

The interface stress patterns associated with the resulting shape, indicated in Fig. 8, are certainly not as uniform as hoped for. The medial shear stress varies between $-1.8 * 10^{-4}$ and $-8.0 * 10^{-4}$ MPa, relative to a uniform object value of $-4.0 * 10^{-4}$ MPa (Table I). The lateral shear stress varies between $-0.9 * 10^{-4}$ and $5.2 * 10^{-4}$, relative to a uniform object value of $4.0 * 10^{-4}$ MPa (Table I). These shear stress patterns are more uniform as compared to those, for example, associated with the Precision Hip, but the maximal peaks are about equal in value.

Proximal/medial compression is fairly uniform with a peak of $-6.0 * 10^{-4}$ MPa and an average very close to the uniform object value of $-4 * 10^{-4}$ MPa (Table I). The tensile stresses in the medial middle and distal regions are fairly uniform as well, with a highest value of $2.0 * 10^{-4}$ MPa, to be compared to an object value of zero (Table I). Similar results are found for the proximal/lateral region. The middle and distal regions on the lateral side display compression up to $-3.5 * 10^{-4}$ MPa, compared to a projected $-1.8 * 10^{-4}$ MPa (Table I). These compressive and tensile stresses are reduced considerably relative to those in commercial designs. Taking again the Precision Hip for a comparison, since its associated stresses are on the lower side already as compared to other designs, the reductions for the optimized shape are in the order of 50-60%, at least for the loading case F_1 on which the present optimization process was based.

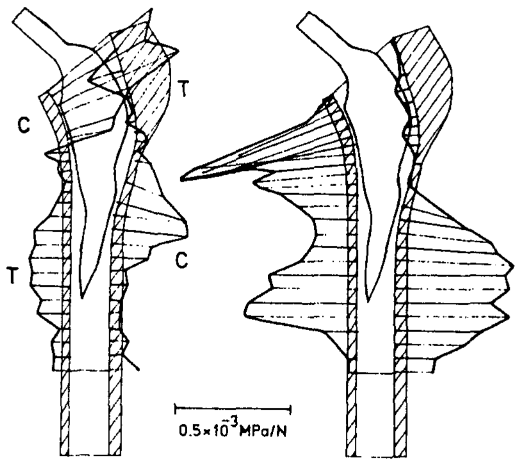

Fig. 8: The final shape, obtained when further iterations did not effect in significant improvements, also showing the associated normal (T=tension, C=compression) and shear stresses at the cement/bone interface. Contrarily to Figs. 3 and 9, tension and compression are represented on opposite sides of the interface lines. The scale factors differ as well.

APPLICATION

Based on the information provided by the outcome of the optimization process, the somewhat awkward shape of the stem was modified to a more practical one, retaining the typical geometric features mentioned above. Evidently, some of the gains in stress reductions could be lost again after this adaptation. Hence, a FEM analysis of this final shape was again carried out, using the standard loading cases and stress evaluation methods discussed earlier in relation with the comparative 'numerical bench tests' of commercial designs.

This modified shape is shown in Fig. 9, together with the associated normal and shear stresses at the lateral and medial cement/bone and stem/cement interfaces (loading case F_1). Evidently (compare Fig. 8), the stress values are not very different from those associated with the original optimized design. When comparing the stress patterns to those around the Precision Hip (Fig. 3), it is found that they have more uniform distributions. The maximal tensile and compressive stresses in particular are reduced. The maximal shear stresses are about equal. When comparing the interface stress values to those found for six popular designs (Huiskes, in press), it is concluded that considerable reductions have been obtained. This is illustrated for normal and shear stresses at the medial cement/bone interface in Fig. 10.

An interesting question is, to which extent the stress values associated with this (modified) optimized stem shape are sensitive to alternative bone properties or to a different placement of the prosthesis relative to the bone. This question was studied in a parametric analysis, using the following variations: (1) a 50% reduced cortical bone modulus, (2) a medialization of the calcar, hence a wider cement wedge at the proximal/medial side, (3) a medialization of the cortex under the calcar, hence an increased cement-mantle thickness medial of the stem belly, (4) a malplacement (lateralization and rotation) of the stem whereby the stem tip is centered in the medullary canal. The effects of these variations on the most relevant cement/bone interface stresses are shown in Fig. 11. Virtually all variations effect in stress increases, but only to a mild extent. Hence, it seems that the stresses are rather unsensitive to small differences in geometrical and material bone properties, and imprecision in the placement of the stem.

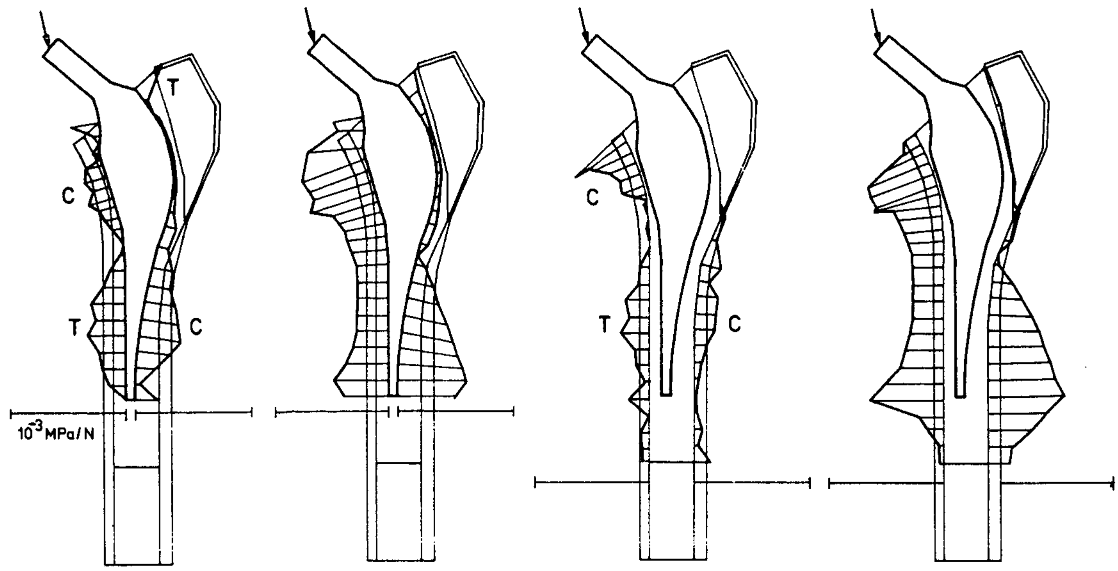

Fig. 9: The modified stem shape and the associated normal and shear stresses at the cement/stem and cement/bone interfaces (T=tension, C=compression).

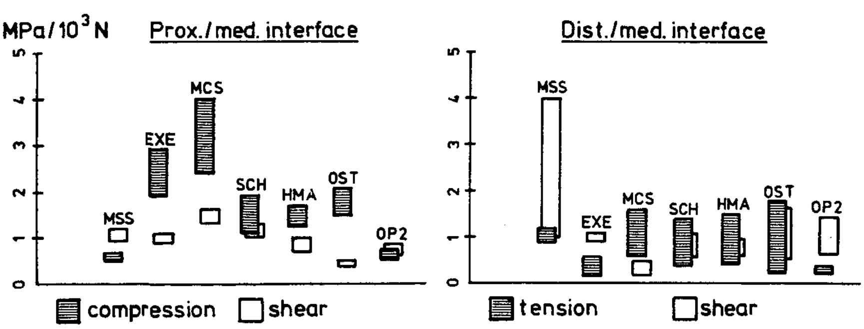

Fig. 10: A comparison of maximal normal and shear stress values at the cement/-bone interface associated with different stem designs, determined in 'numerical bench tests' with the standardized 2-D side plate FEM model, for the loading cases F_1 (lowest values of the bars) and F_2 (highest values). MSS= Mueller Straight Stem; EXE= Exeter; MCS= Mueller Curved Stem; SCH= Scandinavian Hip (experimental design); HMA= Precision Hip; OST= Osteonics; OP2= present (modified) shape.

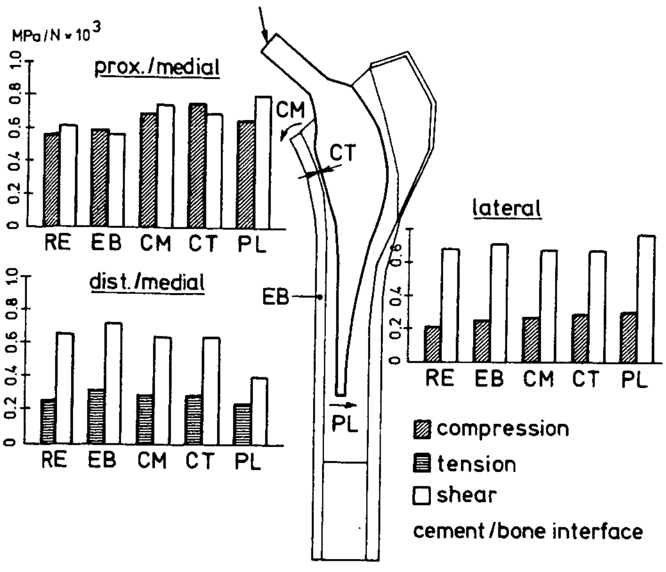

Fig. 11: Maximal stress values (tension, compression, shear) at the medial and lateral cement/bone interfaces associated with the modified stem, as determined for the reference configuration (RE, compare Fig. 10) and four parametric variations; EB: bone modulus reduced 50%, CM: calcar cortex medialized, CT: medial cement layer thickness in the narrowest region increased by medial displacement of the cortex, PL: distal prosthesis lateralized relative to the bone.

DISCUSSION

The NSO procedure showed to be effective in finding a shape for reducing interface stresses. However, the way in which it was applied can be much improved. In the present example, individual stress components in separate regions were considered in the objective function, combined with requirements for a relatively smooth surface contour. The parameters of the objective function were changed manually during the iterative process, using the stress results to send the process in a presumably right direction. Although, evidently, only the end result counts in analyses of this kind, very little basic information can be derived from the process in this way, which is not very satisfying. It is remarkable, that notwithstanding the complex objective function, the effectiveness of the present analysis was not much higher than the earlier one (Huiskes and Boeklagen, in press), in which a single stress parameter (SED) was used in a well-defined objective function, wich was kept intact until the end. Such a process is more comprehensive and gives more information as to the effectiveness of the objective function

Based on these, and earlier analyses we would suggest the following considerations for further use of NSO methods:

- The 'least-pth', in combination with the 'boundary follow' method, works quite well for NSO applications of this kind.
- It is advantageous to use a single stress parameter for the design criteria (e.g. SED in stead of separate stress components) in order to keep the process comprehensive.
- The design criteria should be based on a preset goal for the overall stress or SED patterns, rather than introducing separate regions with associated weight factors. Preferably a theoretical 'optimal' stress or SED distribution should be estimated analytically and used in the objective function (Huiskes and Boeklagen, in press).
- A combination of loading cases in a single stress evaluation should be used, not a separate optimization procedure for different loading cases.

- The shape variations should be restrained, allowing only practical shapes from the beginning, and avoiding the need for manual interference in-between iterations.
- Once the objective function is defined, based on the design criterion, the process should be terminated, without interference, until it converges or must be terminated fur other reasons. Only in this way can any basic information be derived from the procedure, for example relative to the consequences of the objective function for the end results.

With regard to applications of the NSO results to actual stem designs it must be appreciated that some adaptations will usually be necessary to convert a theoretically optimal shape into a practically applicable one, as was also the case in the present example. It must also be recognized, that the results are based on a number of assumptions and simplifications. Firstly, a basic assumption is that the results of THA can be improved if the chances for interface failure are reduced by (further) reduction of interface stresses. Although this seems likely (Huiskes and Nunamaker, 1984; Huiskes, 1986, 1988), we must recognize that other mechanisms, both mechanical and biological, may affect implant loosening as well (Radin et al., 1982; Eftekar et al., 1985; Lewis, 1985). For instance, in the present analyses no consideration is give to the possible effects the stress and strain patterns in bone may have on adaptive bone remodeling and resorption (Huiskes et al., 1987).

Secondly, it must be recognized that the NSO procedure, even if 100% effective, is still executed relative to a particular FEM model which, by necessity, is a rather coarse representation of the complex biological reality (Huiskes and Chao, 1983).

The 2-D side-plate model used here is thought to give a reasonable representation of the predominant characteristics of the load-transfer mechanism in reality, provided that the prosthesis is symmetric relative to the frontal plane, and that the load is predominantly in that plane. This assessment is based on comparisons between 3-D (axisymmetric) models and 2-D side-plate models (Huiskes, 1980), between results of bone surface-strain measurements (Huiskes et al., 1981) and corresponding FEM calculations with the present 2-D model (Huiskes, 1986), both intact and after prosthetic replacement, and on the similarities between stress patterns found in these models and those obtained in (symmetric) 3-D FEM models (Crowninshield et al., 1980). A systematic comparison between 3-D and 2-D side-plate models, limited to an assessment of stem stresses, was performed by Hampton et al. (1980), showing good agreement. In an analysis of a cementless hip prosthesis (Huiskes et al., 1986) we have studied the effects of the side plate, and found it to be an essential feature for realistic stress predictions. In the same project (Huiskes et al., 1986), a (non-symmetric) 3-D model was developed and used to study the stress patterns for a variety of loads. The stress patterns in stem, bone and at interfaces in the frontal section were found to be reasonably similar to those in a 2-D side-plate model, provided that the loads were assumed within this frontal section. Out-of-plane loading tended to affect the stress patterns in and around the distal stem in particular (Huiskes, 1988). Hence, although never systematically documented, the adequacy of the 2-D side-plate model where it concerns the predominant trends of the load-transfer mechanism, has a reasonably strong basis. Nevertheless, the limitation of the model to loads in the frontal plane must be recognized.

The analysis as carried out here also presumes that the interfaces between the different materials are well connected, a description which is most representative for the immediate post-operative situation. When loosening and slip occur, interface stresses can be altered drastically. It was found in earlier, nonlinear FEM analyses that of the four modeling aspects (loads, geometry, material, interface conditions), the interface conditions can be the most influential where it concerns interface stress patterns (Weinans et al., 1987). Hence, when interface loosening has occured, the interface stress patterns of an optimal design may no longer display their optimal features.

The results of an optimization procedure also depend on the external load for which it is carried out. The present analysis was conducted with a (unit) 'physiological' hip-joint force representing the one-legged stance, which is reportedly a 'worst-case' situation (Crowninshield et al, 1978). The earlier

NSO procedure in the 2-D model (Huiskes and Boeklagen, 1988) used a pure bending moment on the femoral head, which is considered as a 'substitute' load for a number of typical loading cases, generating the highest stress peaks in the load-transfer mechanism (Huiskes, 1986). In the earlier 1-D model (Huiskes and Boeklagen, in press), three separate loading cases were used for three individual NSO procedures. Which of these strategies is the most satisfactory in terms of effectiveness and simplicity is not entirely certain. It is remarkable, however, that the most characteristic features of the optimized shapes found were very similar in all these analyses.

In summary, it can be concluded that numerical shape optimization can become an extremely useful tool for artificial-joint design creation and analysis. Obviously, its application is by no means limited to THA stem designs, as in the present example, but equally useful for other kinds of components. It can be equally effective for cementless prostheses, using the appropriate design criteria. However, the application of NSO is still in its infancy. It is still expensive (100-200 FEM runs in the present examples), tedious, and the definition of objective functions, design variables and geometric constraints is not trivial. Much work is still to be done to develop strategies and select mathematical formulations which are both effective and comprehensive. Using a multitude of weighted stress-component values in a number of regions, as in the present example, was found to be less satisfactory in this respect. Attention must also be given to the external loading system for which the NSO procedure is carried out. In the hip joint the loads are variable, and it is not entirely clear which loads create the most damaging stresses in which region. Also in this respects a balance between effectiveness and simplicity should be found in the strategy. Preferably, an optimization should integrate a series of representative loading cases out of a daily cycle.

Notwithstanding these complications and uncertainties in the execution of NSO, and given the relative simplicity of the FEM model used, a stem shape with superior load-transfer characteristics could be derived from the present analysis. This shape has some typical novel features, which make it distinctly different from present prosthetic designs, and provokes cement and interface stresses which are considerably reduced relative to at least the six commercial designs previously tested. How this shape could be realized in a practical design, and whether such a design would also be 'optimal' in a clinical sense, remains to be seen.

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