

# Adaptive bone remodeling and biomechanical design considerations : for noncemented total hip arthroplasty

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# Adaptive Bone Remodeling and Biomechanical Design Considerations

## for Noncemented Total Hip Arthroplasty

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**ABSTRACT:** Clinical problems with noncemented total hip arthroplasty (THA) stems, directly or indirectly related to load transfer, include mid-thigh pain due to relative (micro) motions or excessive endosteal interface stresses, subsidence and loosening due to inadequate primary stability and fit, and proximal femoral bone atrophy due to stress shielding. In this article, the load-transfer mechanisms associated with noncemented THA stems and their resulting stress patterns are discussed in relation to design features, bonding characteristics, and materials choice. Nonlinear finite-element models and computer simulation programs for strain-adaptive bone remodeling have been used for this study. Canal-filling, fully bonded metal stems have been found likely to cause proximal bone atrophy, possibly leading to long-term failure of the implant/bone composite. The use of flexible (isoelastic) materials and/or press-fit fixation reduces stress shielding, but also reduces the potential for interface stability. The stem material, the stem shape, and the coating geometry interact in relation to the load-transfer mechanism, and it is suggested that optimal combinations of these characteristics can be determined through the computer simulation methods presented.

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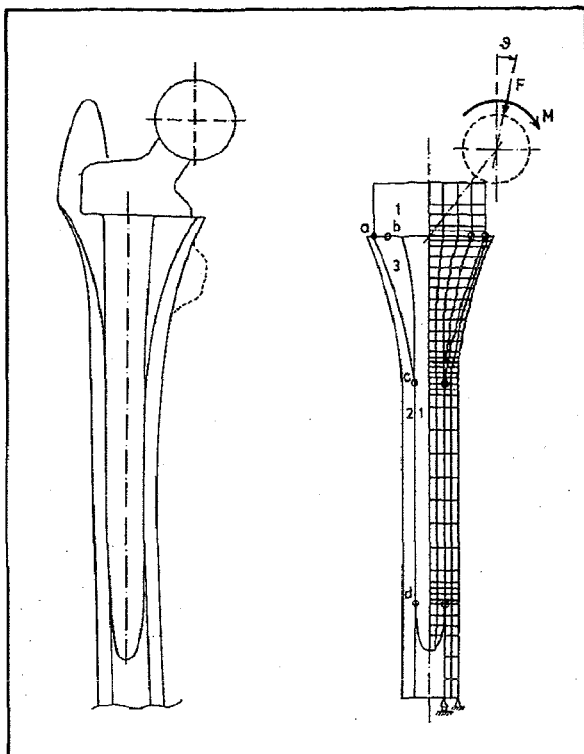
### Introduction

The transfer of load from prosthesis to bone without causing mechanical failure or pain is one of the most important functions of a total hip arthroplasty (THA). The load-transfer mechanism generates stresses in prosthetic materials, in bone, and at interfaces. Excessive stresses may cause fatigue failure of components if they occur within the implant, or interface disruption if they occur at the implant/bone fixation. Interface disruption causes relative (micro) motions, with pain and interface bone resorption as a possible long-term effect. Unnatural stresses in bone may cause adaptive remodeling and bone atrophy, possibly weakening the bone bed to the point of failure and creating an unfavorable basis for a revision arthroplasty. Hence, it is important to study the stress patterns generated by the load-transfer mechanism and to understand their relationships with hip-joint loading characteristics, prosthetic design, materials, and the characteristics of the fixation.

Owing to an extensive clinical experience and a large number of laboratory experiments and computer analyses, the effects of load transfer and stresses associated with cemented prostheses are reasonably well documented and understood. Correlations between clinical results and finite element analyses have shown that the cement/bone interface is the weakest link in the system, and that any measure taken to increase strength or reduce stress peaks also reduces the incidence of loosening.<sup>1</sup> Whereas interface strength is predominantly a matter of surgical and cementing techniques, interface stress is mostly a matter of implant design for given patient weight and activity level.<sup>2</sup> On the basis of clinical and experimental data gradually obtained, improved prosthetic designs and fixation techniques have been developed.

Where noncemented THA is concerned, however, less information is available. The clinical experience is relatively short and diverse, involving a multitude of fixation

**Fig 1: Axial symmetric FEM model of the Lord (PUP) stem, based on a geometry study.<sup>9</sup>**



principles and designs. Reported clinical problems directly or indirectly related to biomechanical phenomena include postoperative mid-thigh pain due to relative (micro) motion or excessive endosteal interface stresses,<sup>3</sup> initial postoperative relative motion, subsidence and loosening due to inadequate primary stability and fit,<sup>4</sup> and proximal femoral bone atrophy due to stress shielding.<sup>4,5</sup>

This article documents the principles of the load-transfer mechanism in noncemented femoral components of THA and investigates the relationship between stress patterns and stem designs in terms of the clinical problems mentioned earlier. Particular emphasis is given to the different effects of press-fit and bonded prostheses (hydroxyapatite or porous-ingrowth coated) and the effects of materials choice. Means of investigation are the finite element method (FEM) and computer simulation models for strain-adaptive bone remodeling.

## Methods

### Finite Element Analysis

The stress patterns in a THA structure are complex in the sense that the stress values vary among and within the

different materials for each single loading situation. Several stress components are at work at any given point.<sup>6</sup> In the cortex of a femur, for instance, we find longitudinal (or bending) stress, hoop stress, and radial (or transverse) stress, all with a specific distribution within the material. In addition, three mutually perpendicular shear-stress components can be identified; hence, we have six independent components in every point of the structure. Fortunately, not all of these components are equally important. The predominant stresses in a femoral THA structure are the longitudinal (bending) stresses and the hoop stresses in the cortex, the longitudinal (bending) stresses in the stem, and the shear stresses and normal direct stresses (tension or compression) working at the implant/bone interface.<sup>7</sup>

The patterns of these stresses, their values, and their distribution within the THA structure depend on: the magnitudes and directions of the external loads, the elastic moduli of the materials, the shape of the individual structures (implant, bone), and their fixation characteristics (bonded or loose). If these four characteristics can be described quantitatively, then the stress patterns can be determined with the finite element method.<sup>6,8</sup> The FEM is a computer simulation method combining location-dependent material properties, geometry and interface properties, and external loads as the input data, and generating location-dependent stress values. Conceptually, an FEM analysis can also be considered as a numerical experiment. As in all computer simulations and experiments, the information provided can be correct only insofar as the basic input data adequately describes reality. In this respect, an FEM model is always a generalized, schematic representation of reality whereby geometry and properties are described by some degree of approximation. Hence, the resulting stress predictions are also schematic, generalized, and approximate relative to reality. The extent to which an FEM model is schematized and simplified depends on the purpose of the study, and judging whether the stress patterns are realistic and if so, in what detail, is mostly a matter of experience.<sup>8</sup>

Figure 1 shows an FEM model of the femoral Lord (PUP) THA. The PUP stem has a circular cross section with fluted, polarized corrugations on the surface to provide resistance against torsion.<sup>9</sup> Its geometry is derived from the earlier Lord madreporic stem.<sup>3</sup> The model is schematic in the sense that its geometry is assumed to be axially symmetrical, whereby the circumferential stress variations are described by Fourier series.<sup>6</sup> Materials are described as linear elastic (elastic modulus for cortical bone:  $E = 1.7 \times 10^4$  MPa; cancellous bone:  $E = 1.0 \times 10^4$  MPa; CoCrMn alloyed stem:  $E = 20.0 \times 10^4$  MPa). A hip joint force  $F$  or a bending moment  $M$  are assumed as external hip-joint loads. The prosthetic collar is assumed not to be in contact with bone, in accordance with experimental information.<sup>9</sup> The implant/bone interface is either

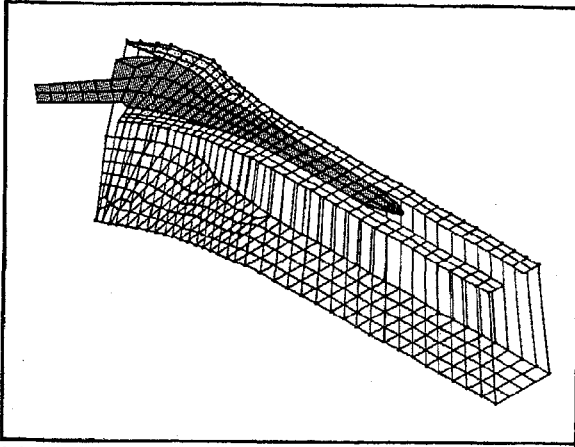


Fig 2A.

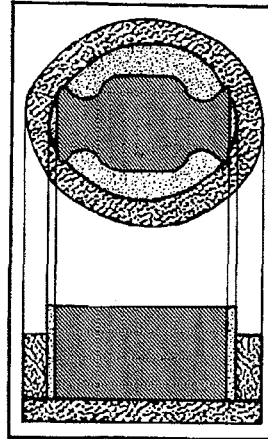


Fig 2B.

Fig 2: Two-dimensional sideplate FEM model of the Zweymüller stem, in which two element layers are superimposed, one representing the midfrontal section through the bone-prosthesis structure (front plate), and one representing the three-dimensional out-of-plane cortical integrity (side-plate) (A). Inset shows how the cross-sectional properties are derived (B).

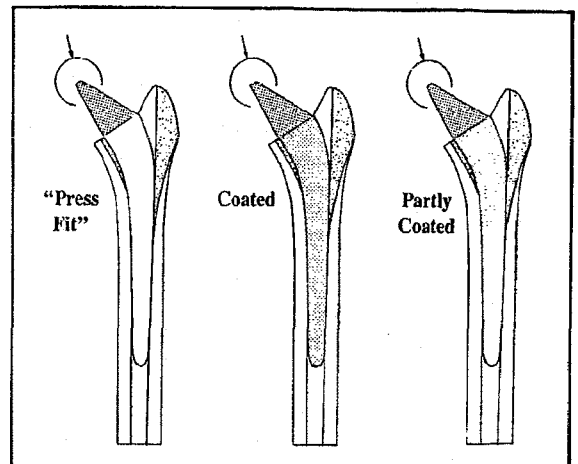
assumed as bonded or as sliding with no friction. In a parametric analysis, the prosthetic material was varied to titanium ( $E = 11.0 \times 10^4$  MPa) and to a hypothetical isoelastic material with the same elastic modulus as cortical bone ( $E = 1.7 \times 10^4$  MPa).

Another kind of schematized FEM model is shown in Figure 2. This is a two-dimensional sideplate model, whereby the three-dimensional structural integrity of the bone is accounted for by a separate two-dimensional mesh, a sideplate. This FEM plate is connected to the front plate, which represents the midfrontal section of the structure (Fig 2). Such a two-dimensional sideplate model gives a reasonable approximation of the stress patterns if the loads work predominantly in the frontal plane.<sup>7</sup> Of course, the effects of torsional loading on the prosthesis cannot be studied in such a model, and the hoop stresses in the cortex cannot be determined.

This two-dimensional sideplate model was used to analyze load transfer associated with two types of Zweymüller stems, the traditional model<sup>10</sup> and the newly designed SL stem, and with the Osteonics (Allendale, NJ) stem.<sup>11</sup> In the latter case, the effects of an uncoated (press-fit) stem, a fully coated, and a partly hydroxyapatite-coated stem<sup>12</sup> were investigated (Fig 3). The external load in these models is a hip-joint force directed according to the one-legged stance in gait.<sup>13</sup> But variations in load directions are studied as well. The elastic moduli are the same as those given for the Lord-stem model. The bone geometry and its relation with the external force are always equal for each prosthesis or variation studied.

The interface connections in the two-dimensional sideplate models are assumed to be either fully bonded or loose. In the latter case, nonlinear-gap elements are used,<sup>14</sup> allowing local slip and separation between implant and bone to occur. Friction is not accounted for.

Fig 3: Hydroxyapatite coating geometry studied in the Osteonics model.

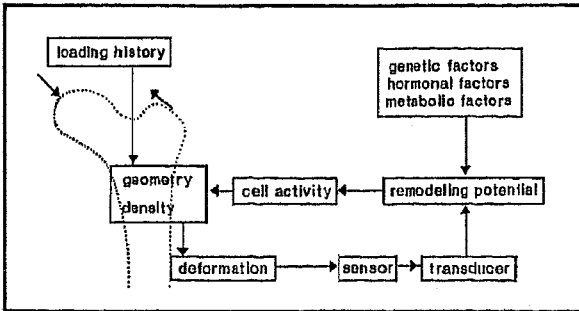


### Strain-Adaptive Bone-Remodeling Analysis

The ability of bone to form an optimal structure for load resistance and to adapt its structure to alternative loads is qualitatively described in Wolff's law.<sup>15</sup> This ability to adapt implies that bone must have internal strain sensors and transducers that enable it to react to an actual state of deformation and to translate mechanical signals to biochemical ones. Since it is unlikely that bone would react immediately to every single change in load, it is probable that bone also has a means of integrating the loading history over a certain period of time and of storing this information in order to react gradually to a definite change in the general loading environment.<sup>16</sup>

After THA, the hip-joint and muscle forces may or may not be similar, but the bone around the prosthesis will

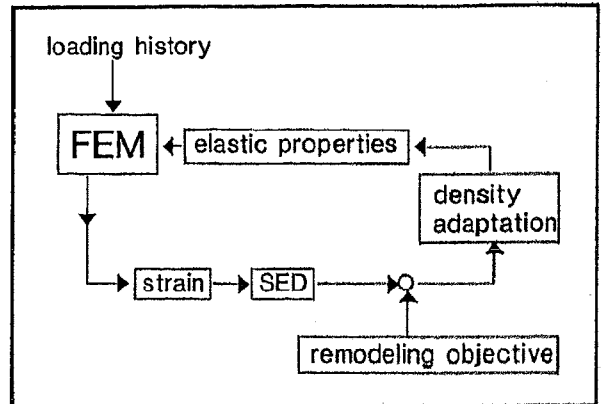
Fig 4: Schematic, hypothetical chain of events governing strain-adaptive bone remodeling according to Wolff's law.



certainly experience different stresses and strains related to the presence of the prosthesis in the bone. Increases or reductions in bone mass, depending on the new state of deformation, may then occur. Obviously, normal bone remodeling associated with osteoclast and osteoblast activity is a continuous process. Hence, net increases or reductions of bone mass occurring in adaptive remodeling are associated with a disturbance of the remodeling balance. Figure 4 shows schematically the chain of events that is assumed to govern Wolff's law.<sup>17</sup> A local change in strain is sensed by the bone and translated by an, as of yet, unknown transducer to a chemical strain-remodeling potential which, integrated with genetic, hormonal, and metabolic factors, causes a remodeling signal that activates osteoblasts or osteoclasts. The actions will affect either the porosity (or density) of the bone, and thereby its elastic modulus (internal remodeling) or its geometry (external or surface remodeling). In both cases, the deformations in the bone change again, owing to the geometric or material adaptations, thereby also changing the strain-remodeling potential that signals cell activities. The process thus continues, or so it is assumed, until the state of deformation is normalized to the physiologic strain values (homeostasis) whereby bone density and shape are again optimally adapted to the loads.

In recent years, quantitative strain-adaptive remodeling theories have been formulated.<sup>16,18,19</sup> These are essentially mathematical descriptions of the scheme mentioned earlier and quantitative formulations of Wolff's law. These theories are characterized by a remodeling signal, a remodeling objective, and a remodeling rule. The remodeling signal is a mechanical variable representing the actual state of deformation that is assumed to be sensed by the bone cells. The remodeling objective is a mathematical criterion representing the assumed equilibrium relationship between the remodeling signal and bone mass. The remodeling rule is a mathematical formula incorporating the remodeling objective and represents the way in which bone is assumed to reach this equilibrium objective.

Fig 5: Schematic characterization of the iterative FEM-integrated computer simulation model of strain-adaptive bone remodeling.



The present analysis is confined to internal remodeling, whereby bone mass is measured by the local apparent density. For a number of reasons, the apparent strain energy density (SED)  $U$  is chosen as the local remodeling signal.<sup>16,19</sup> This variable can be directly calculated from the stresses and the strains in each point of the bone. The remodeling objective used was first proposed by Carter et al<sup>16</sup> and assumes that bone is a self-optimizing material, adapting its local apparent density  $\rho$  to the local SED  $U$  until the ratio  $U/\rho$  reaches a given uniform value. Using the relationship<sup>20</sup> between elastic modulus and apparent density  $E = C\rho^3$ , this remodeling objective can be incorporated in a remodeling rule for the rate of change of the elastic modulus

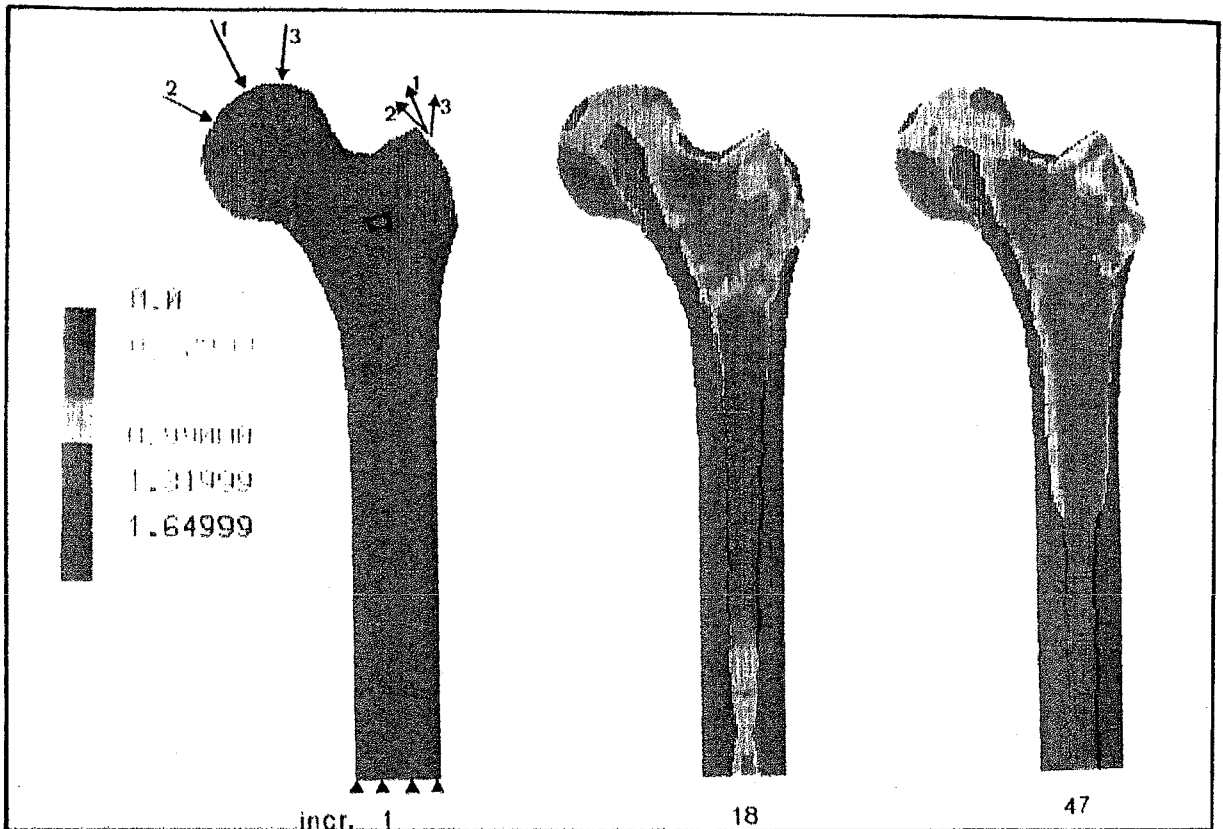
$$dE/dt = A (E - CU^3)$$

whereby  $A$  and  $C$  are constants.<sup>21</sup> This remodeling rule is used in a computer simulation program, incorporating an FEM model, as illustrated schematically in Figure 5. In the iterative program, in which each iteration represents a time step, the elastic modulus in each element of the FEM model is adapted stepwise until it satisfies the above relation with the SED  $U$  determined by the FEM for each element. The possible variations of the elastic modulus are bounded between 0.01 MPa and 20,000 MPa ( $\rho$  between 0.014 and 1.74 g/cm<sup>3</sup>), representing complete resorption or cortical bone in an element, respectively.

This scheme was applied in a two-dimensional FEM sideplate model of a natural proximal femur, using an integrated loading series (loading cases 1, 2, and 3) representing daily activity.<sup>16</sup> The process was started with a uniform apparent density throughout the whole bone (including the intramedullary cavity) of 1.28 g/cm<sup>3</sup>. The density distributions predicted by the computer simulation program after the first, 18th, and final 47th iterations are shown in Figure 6. The density distribution gradually adapts to a final configuration that very closely mimics the

Fig 6: Starting from a bone with uniform apparent density distribution, the optimal density distribution is predicted by the computer simulation model. A series of loading cases (1, 2, and 3) representing daily activity is assumed. Shown are

the density distributions after the 1st, 18th, and final (47th) time steps. Colors represent apparent-density values, shown in  $\text{g}/\text{cm}^3$ .



real femur, including a medullar cavity; Ward's triangle; and characteristic structures in the femoral head and greater trochanter. This result lends confidence to the simulation method in general and to the remodeling objective chosen in particular. Recently, however, in comparing different kinds of remodeling objectives, it was found that the final density distribution predicted is not very amenable to precise mathematical form.<sup>21</sup>

This same scheme was applied to investigate the effects of stress shielding on strain-adaptive bone remodeling around the fully bonded Osteonics prosthesis. For this purpose, the natural density distribution of Figure 6 was used as the initial configuration.

## Results

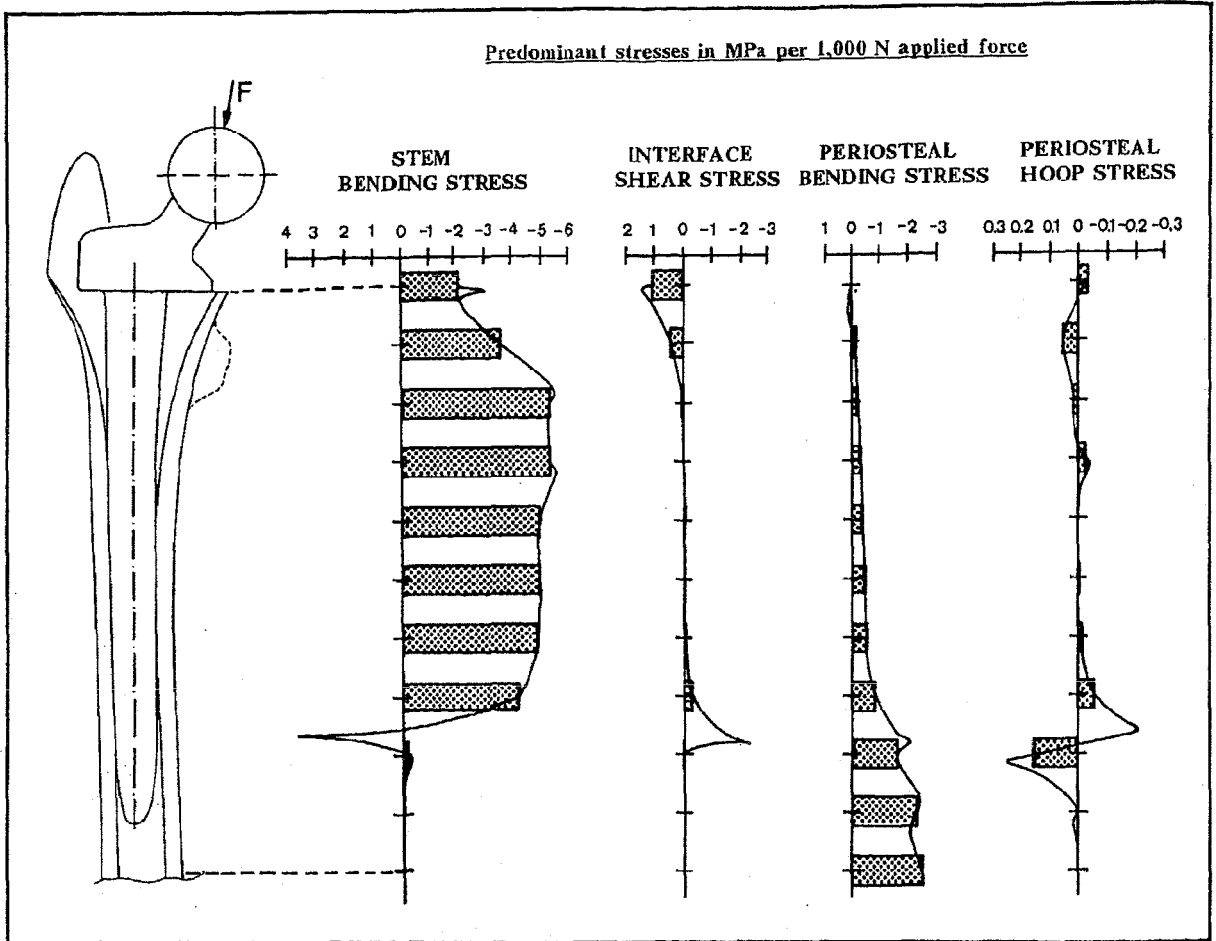
### Stress Shielding and Prosthetic Material

The predominant stress patterns resulting from the (bending) load-transfer mechanism in the Lord THA sys-

tem are illustrated in Figure 7. The stem (bending) stresses display their maximal values in the middle region of the stem but are low relative to the (fatigue) strength of the CoCrMd material. Of more interest are the interface stresses, with their typical peak values near the distal and the proximal sides of the stem. Since interface shear stress represents the potential for interface slip, it is evident that slippage would be initiated predominantly at these locations. Because the implant/bone interface of the Lord (PUP) prosthesis is rather straight, the interface shear-stress values are relatively high, owing to the axial component of the hip-joint force particularly. If axial interface slip can be resisted by interface friction only, it is not unlikely that it will indeed occur. It is also evident from the interface shear-stress patterns in Figure 7 that the extreme length of the stem serves no purpose at all because load is transferred predominantly at the proximal and distal ends and not in the midregion. A similar conclusion can be drawn from the interface compressive and tensile stresses, which are not shown in Figure 7.

Fig 7: Stresses in stem, bone, and at the implant/bone interface as determined in the axial symmetric Lord model.

Bonded interfaces; CoCrMd alloyed stem; hip-joint force loading.



The periosteal bending stresses (Fig 7) are reduced relative to the natural distribution in the proximal region of the bone owing to the presence of the stem. This reduction represents the stress-shielding effect of the stem. The bone stresses in general are somewhat augmented by the presence of elevated hoop stresses, but their peak values are still an order of magnitude lower than the bending stresses (Fig 7).

The extent of the stress-shielding effect is shown again in Figure 8A, giving periosteal bending-stress values relative to the natural ones, comparing CoCrMd, titanium, and the hypothetical isoelastic material. The extent of stress shielding is directly related to the stiffness of the stem, which depends on its thickness and on the elastic modulus of its material. The effects of stem thickness on stress shielding and cortical resorption were studied earlier.<sup>19</sup> In the case of noncemented stems, which are usually canal filling, stem thickness is a given quantity

depending on intramedullary width, and stem stiffness can be varied only by materials choice. As is evident from Figure 8A, the lower-modulus materials such as titanium, or in particular a hypothetical isoelastic one, may reduce the stress-shielding effects considerably, but only below the lower proximal region. Near the upper proximal side, stress shielding remains extensive, whatever the value of the stem's elastic modulus. The reduction of stress shielding by using low-modulus materials has a price also, as indicated in Figure 8B. For the low modulus stems, as compared with the higher modulus ones, load transfer shifts to the proximal side at the expense of the distal side. This implies that the proximal interface stresses increase and the distal ones decrease. As shown in Figure 8B, this effect would be particularly dramatic for the hypothetical isoelastic material, in which proximal shear stresses would increase by a factor of four relative to a CoCrMd stem.

Fig 8: Relative stress values found in the Lord model for a bending-moment applied load and bonded interfaces. Periosteal, medial bone bending stress for CoCrMd, titanium, and isoelastic stems, relative to the natural values (A). Interface shear stresses generated by titanium and isoelastic materials, relative to those found for a CoCrMd alloyed stem (B).

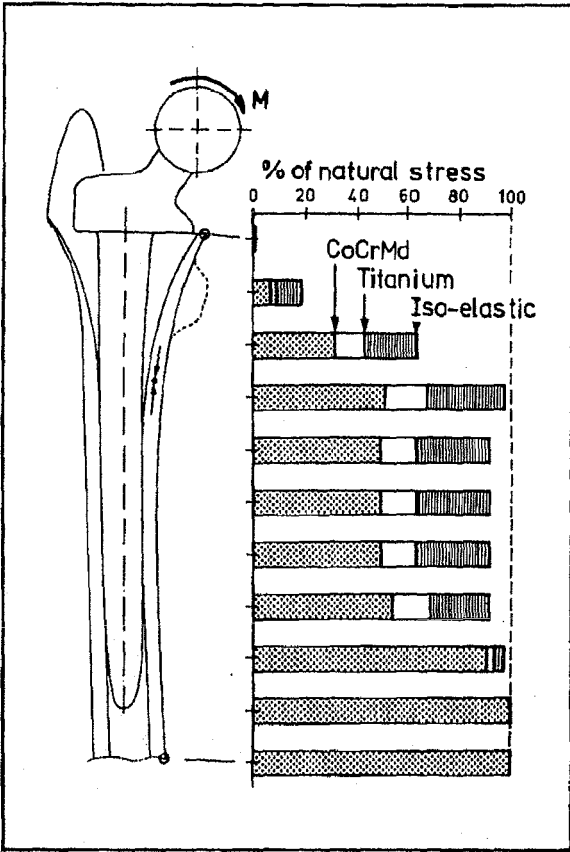


Fig 8A.

Similar effects to those found in the analysis of the Lord THA are evident in the results of the Zweymüller and the Osteonics models, when comparing the effects of materials choice on stress shielding. Figure 9 shows a typical example from the Zweymüller SL model, comparing medial endosteal bending stresses and implant/bone interface shear stresses for a titanium stem (which is the actual material of the prosthesis) and a CoCrMd stem. A change to CoCrMd, relative to titanium, would result in a reduction of about 32% in bone stresses 10 mm below the calcar (hence, more stress shielding), about 14% higher stem stresses (hence, higher chances for stem failure, although this would still be very unlikely), and about 27% higher interface shear stresses at the distal side (hence, greater probability of distal slip). Considering these effects, titanium would obviously be the material of choice. On the other side, however, we would obtain 20% to 35% reductions in proximal/medial interface shear stress and

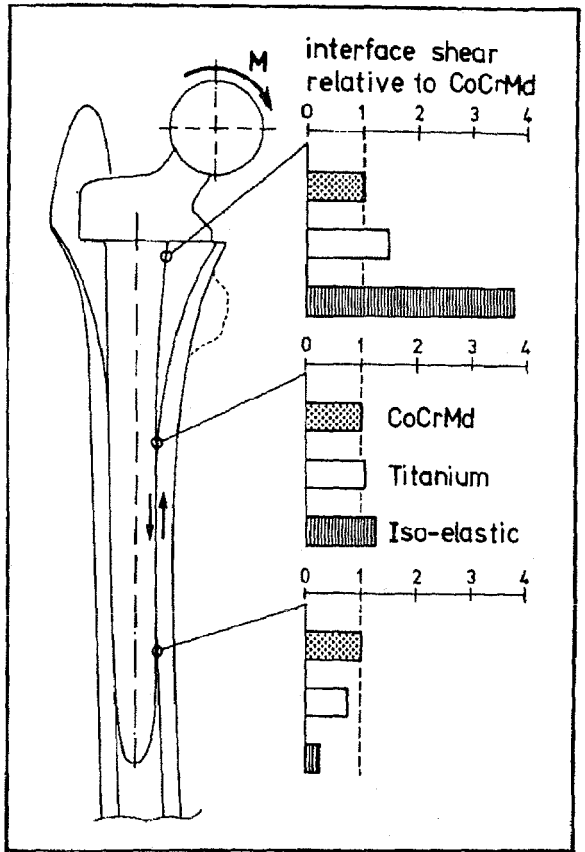


Fig 8B.

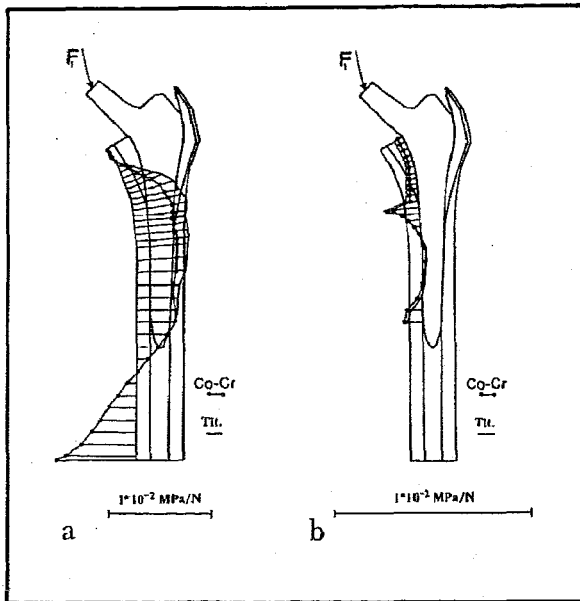
compression when choosing CoCrMd, giving less probability of proximal slip and interface failure.

### Stress Shielding and Bone Resorption

According to Wolff's law, the stress-shielding effects discussed in the previous section would cause net bone resorption until the stresses and strains were again normalized. This is indeed reported frequently, based on radiographic patient evaluation and animal experimental studies associated with stiff, canal-filling noncemented prostheses.<sup>4,5,22,23</sup> For the purpose of designing implants and choosing materials, it would be very helpful if the extent of bone resorption associated with a particular prosthetic design could be predicted, if only by rough approximation. The adaptive-remodeling computer simulation models discussed earlier were actually developed for this purpose, and the results of such an analysis are presented in Figure 10.



Fig 9: Periosteal, medial bending stress patterns (A) and medial interface shear stresses (B) found in the Zweymüller SL model, comparing a titanium stem (drawn curve) to a CoCrMn alloyed one (dotted curve). Interfaces bonded; values shown in MPa per  $N$  applied force.



A two-dimensional sideplate FEM model is again used, whereby the natural apparent-density distribution determined by the earlier analysis (Fig 6) is taken as the initial configuration. An Osteonics stem, fully coated with hydroxyapatite and assumed fully osseointegrated (bonded interface), is introduced into the model. Using the same three loading situations representing a typical daily loading cycle (Fig 10), the iterative remodeling simulation is restarted and continued until homeostasis is obtained in each element; in other words, until the objective  $E = CU^3$  (equivalent to  $U/\rho = \text{constant}$ ) is reached in the elements, until the bone has resorbed (minimal value of apparent density) or until it has turned into cortical bone (maximal value of apparent density). This stage is reached after 18 iterations. The new equilibrated apparent-density distribution, shown in Figure 10, suggests severe cortical resorption in the upper region of the bone.

The question arises as to what extent this prediction is realistic and, if so, whether it would become a clinical problem. These questions are discussed later. From a biomechanical point of view, the proximal resorption has the effect of increasing distal load transfer at the expense of the proximal side. This is illustrated in Figure 11, in which the colors do not represent apparent density as in Figure 10, but represent stress intensity (von Mises stress). Evidently, the proximal bone is almost fully bypassed in the load-transfer mechanism after proximal bone resorp-

tion has occurred. This mechanism illustrates the self-perpetuating effects of the proximal bone resorption process, whereby bone resorption increases stress shielding, which then again stimulates bone resorption, and so on. As a result, very high interface stresses occur near the distal stem tip in the homeostatic configuration, possibly causing eventual mechanical failure.

### Bonded Versus Press-Fit Stems

As seen earlier, the stress-shielding effects of the canal-filling noncemented prostheses, and hence the extent of bone atrophy, can be reduced somewhat by using more flexible materials but with the adverse effect of creating increased stress at the proximal interface. Another way of reducing stress shielding is by using alternative fixation methods. In all three model studies reported here (Lord, Zweymüller, and Osteonics), it was found that the assumption of a loose (press-fit) interface produced higher proximal cortical bone stresses as compared with bonded interfaces. These effects are evaluated in Figure 12 for the Zweymüller and the Zweymüller SL models. Particularly for the latter model, the effect is quite extensive.

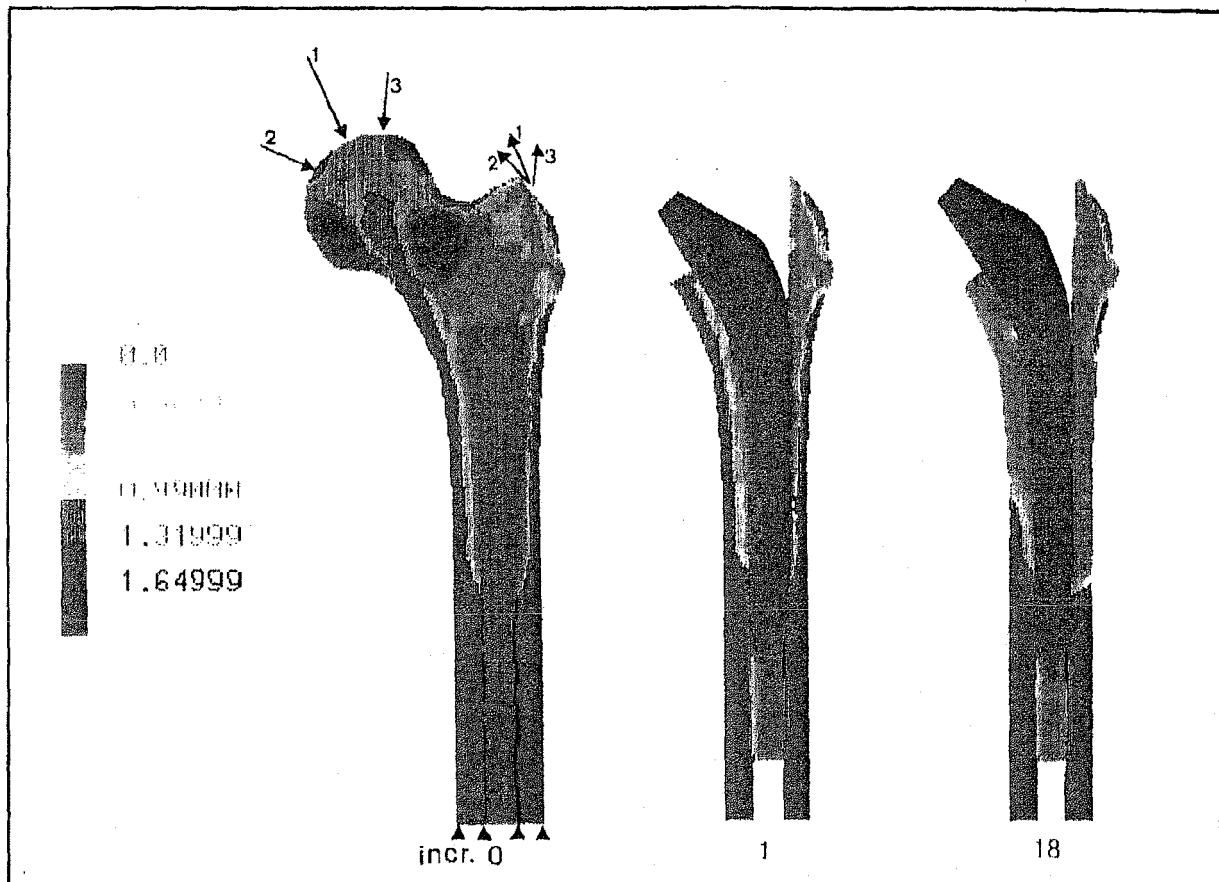
These elevated cortical stresses are mainly caused by increased deformation of the bone in bending, due to a reduced stiffness of the whole structure when the implant/bone interface is no longer bonded. The corresponding effects in the Osteonics model are illustrated in Figure 13, comparing medial cortical bending stresses for the fully hydroxyapatite coated, partly coated, and press-fit configurations (Fig 3) to those in the natural bone. The press-fit model creates stresses in excess of the natural ones, but only below the lower proximal region. Near the calcar, extensive stress shielding is still apparent. The stress-shielding effects of the fully coated (and osseointegrated) stem are the same as discussed earlier. The partly coated stem generates moderately elevated cortical stresses, and hence reduced stress shielding around the noncoated part of the stem. However, near the proximal side, the stress-shielding effects are as severe as for the fully coated stem.

An interesting phenomenon is seen with regard to the endosteal bone stresses near the lower edge of the coating in the partly coated configuration (Fig 14). Particularly in this area, stress concentrations are caused by the abrupt transition from a bonded to a press-fit interface. Geesink<sup>12</sup> reported radiographic denseness of bone in his patient series with the partly coated Osteonics stem precisely at this location.

According to the press-fit concept, prosthetic fixation is predominantly maintained by the generation of interface compression. It is intuitively obvious that the shape of the interface contour of a stem design, in relation to the site-dependent elastic characteristics of the bone by which the stem is contained, has an overwhelming influence on the compressive stress patterns generated at the interfaces—much more so than in the case of a bonded interface. This

Fig 10: Strain-adaptive remodeling prediction of bone around the fully bonded Osteonics prosthesis ( $E = 20.0 \cdot 10^4$  MPa). Colors represent the apparent density distribution with values in  $\text{g}/\text{cm}^3$ . The initial configuration of the natural

bone (see Fig 6) is shown with the three loading cases out of the daily loading cycle. After the final, 18th time step, when homeostasis is obtained, considerable proximal bone atrophy is predicted.



is illustrated in Figure 15, comparing average compressive interface stresses in two medial and two lateral regions generated by the Osteonics, Zweymüller, and Zweymüller SL designs in the bonded and in the press-fit cases. A few features can be noted from these figures:

1. The stress values in the press-fit cases are much higher than those in the bonded cases, and interface tension occurs in the latter case only, both of which are rather obvious phenomena in view of the interface characteristics modeled in both cases.
2. Despite the equal FEM model characteristics and external loads, and the relative similarity of the stem shapes, the stress values for the three prosthetic types differ considerably in the press-fit cases, much more so than in the bonded configurations. This is due precisely to the point made above relative to the importance of the interface contour.

The Osteonics stem, compared with the Zweymüller models, lacks the lateral/proximal wing for additional lateral support. As a result, the Osteonics stem is pressed toward a valgus position, thereby developing elevated compression at the lateral side near the isthmus region (Fig 15A), an effect similar to pushing a curved rod into a straight hole. The two Zweymüller designs, although very similar, also display quite different compressive stresses when comparing the four interface regions identified (Fig 15B and C). The differences are almost exclusively due to a more lateralized position of the proximal medial stem contour in the SL design (Fig 15C). This creates a thicker cancellous-bone buffer in the proximal medial area, giving a more flexible bone constraint to the press-fit stem. Consequently, distal stress transfer increases and proximal stress transfer decreases compared with the traditional Zweymüller design (Fig 15B).

Fig 11: Stress transfer in the resorbed prosthesis/bone configuration of Figure 10. The colors represent (von Mises) stress intensity, high for red, low for dark green.

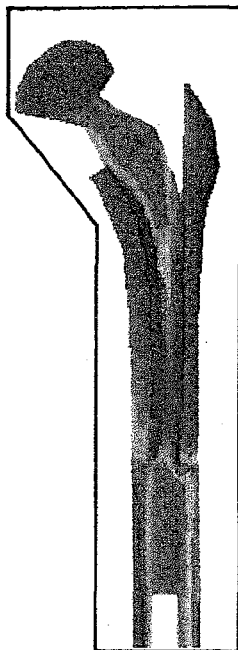


Fig 12: Medial periosteal bending stresses, 10 mm below the calcar, as determined in the Zweymüller models, relative to the natural value (ZF: Zweymüller, bonded; ZSLF: Zweymüller SL, bonded; ZP: Zweymüller, press-fit; ZSLP: Zweymüller SL, press-fit).

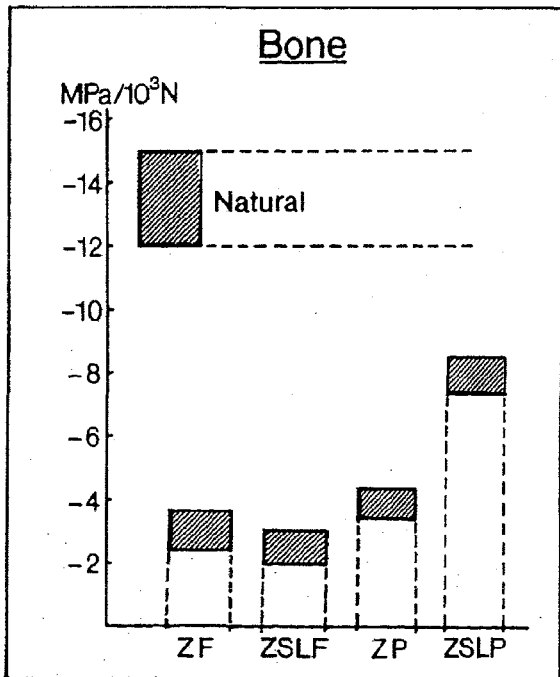
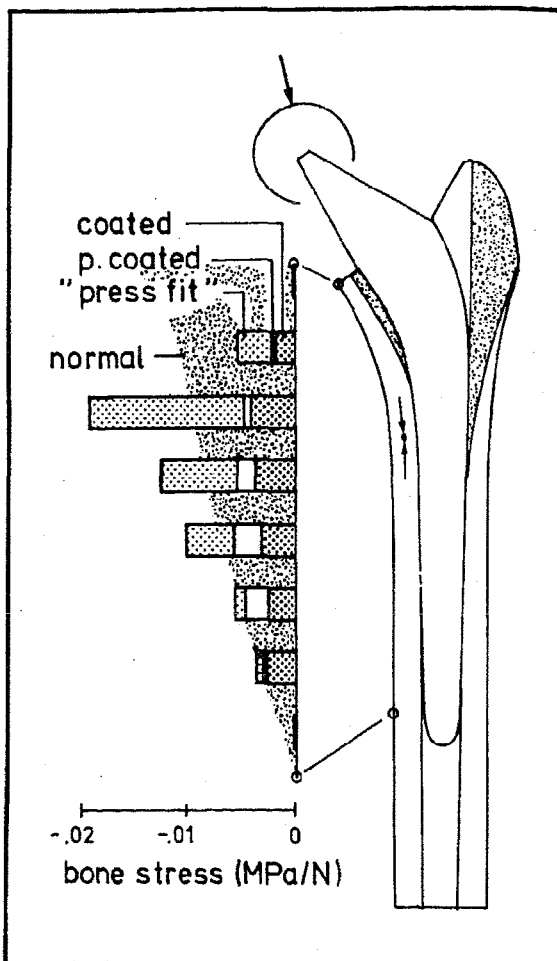


Fig 13: Medial, periosteal bone stress distributions as determined in the Osteonics model (CoCrMd material), comparing fully coated, partly coated, and press-fit stems to the natural stress values (shaded area).



**Discussion**

It must be appreciated first of all that the results presented here are not meant to compare the relative merits of the commercial THA designs analyzed. These designs were based on different philosophies involving more considerations than just the load-transfer mechanism. To establish whether they are effective requires long-term follow-up studies, for which the information presented here may nevertheless be helpful. The purpose of this article is to investigate the relationships between prosthetic design and load transfer, whereby the actual designs analyzed are used as examples.

Second, it must again be noted that the FEM models are schematic by nature, and suitable for investigating the

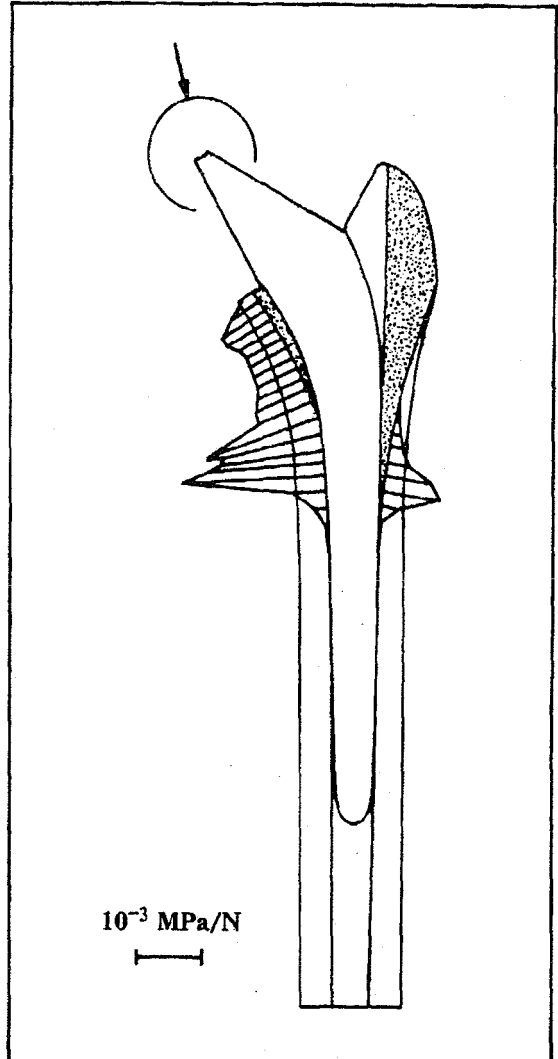
load-transfer mechanism and its relationship with prosthetic design aspects in general, but not realistically in every detail. Noncemented fixation of hip stems is not a readily reproducible process—even less so than the cemented prostheses. This is predominantly a matter of fit. Although both press-fit and coated prostheses are meant to obtain a precise fit with bone over the whole interface, this is in fact hardly ever realized in the fixation procedure for any design.<sup>9,24</sup> Contact is usually obtained in a few individual regions, typically amounting to a total area of some 20% of the whole interface.<sup>9,24</sup> As a result, the stress patterns in bone tend to vary considerably from case to case.<sup>25</sup> It is obvious, therefore, that the press-fit models in particular are idealized, being based on an assumption that a more or less uniform medial and lateral fit has been realized, or at least that the individual contact points are more or less uniformly distributed. Hence, the models represent design concepts, rather than particular patient cases.

The strain-adaptive remodeling theories used are, of course, hypothetical by nature, since the bone's true remodeling mechanisms, its actual remodeling signal, and its remodeling objective are as yet unknown. Nevertheless, predictions of proximal/femoral density distributions obtained with these theories are quite realistic,<sup>16,19,24</sup> as is also found in the present analysis. In addition, it appears that results of animal experiments relative to bone resorption around noncemented stems can be predicted reasonably well with the computer-simulation procedure presented. Examples are the experiments of Miller and Kelebay,<sup>22</sup> and those of Turner et al.,<sup>23</sup> both simulated with strain-adaptive remodeling theories.<sup>19,26</sup>

The FEM analyses indicate that the degree of proximal cortical stress shielding around canal-filling, bonded metal stems is quite extensive. As suggested by the remodeling analysis, this may cause extensive proximal bone atrophy. These phenomena have also been reported to occur in clinical patient series,<sup>5,4,12</sup> although to a much less dramatic extent than predicted here or found in animal experiments. In a way, this may be deceptive, because net bone remodeling cannot be measured very accurately from the patient radiographs usually used in follow-up studies. As reported by Draenert,<sup>27</sup> based on morphologic studies of post-mortem material, bone-atrophic phenomena similar to the predictions reported here are indeed found around noncemented hips. However, the remodeling predictions are somewhat deceptive, too, because they are based on the assumption that patient activity levels, and hence the hip joint and muscle force characteristics before and after the operation, are similar. For the most part, it seems reasonable to assume that the trends of the remodeling phenomena predicted are realistic in a general sense, but the extent is somewhat exaggerated.

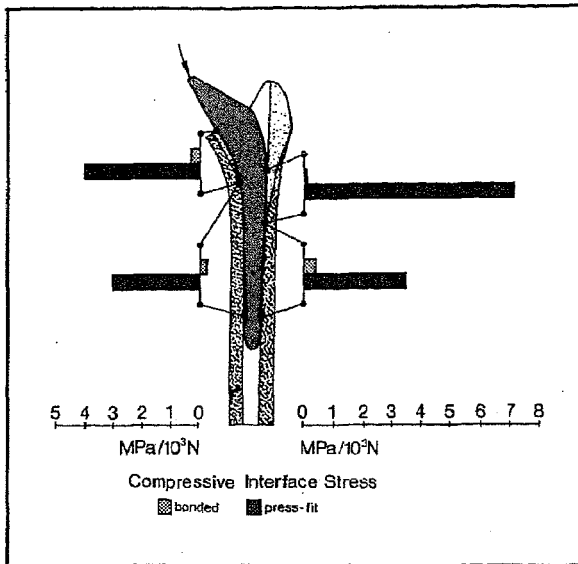
An important question is whether and to what extent proximal bone atrophy will become a clinical problem. As

Fig 14: Medial and lateral shear-stress distributions at the endosteal cortex associated with the partly coated Osteonics prosthesis. Stress concentrations near the lower coating edge correlate with radiographically determined bone densifications in patient series.<sup>12</sup>

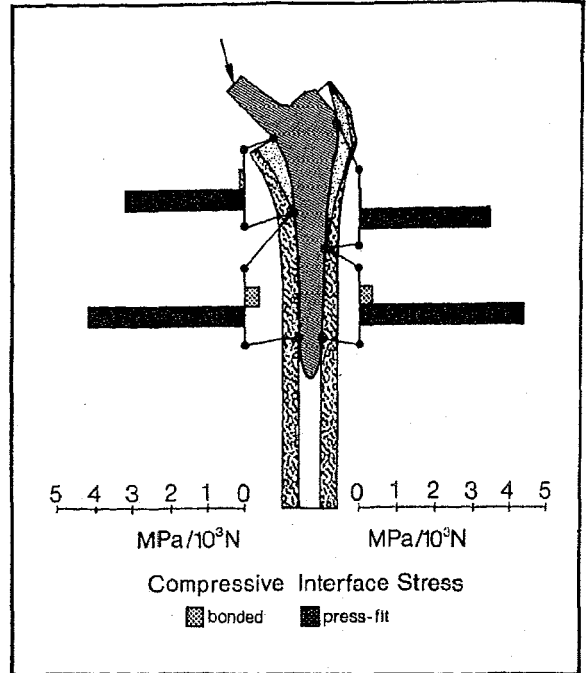


shown here, it causes a shift in load transfer from proximal to distal so that the stem stresses and the distal bone and implant/bone interface stresses increase. This is similar to what occurs with cemented stems after proximal loosening. Such a mechanism may eventually cause mechanical failures, especially in active patients. Another important consideration is that proximal bone resorption is a negative factor for the revisability of the arthroplasty. Hence, designing noncemented prostheses that reduce the chances for bone resorption appears to be worthwhile. In

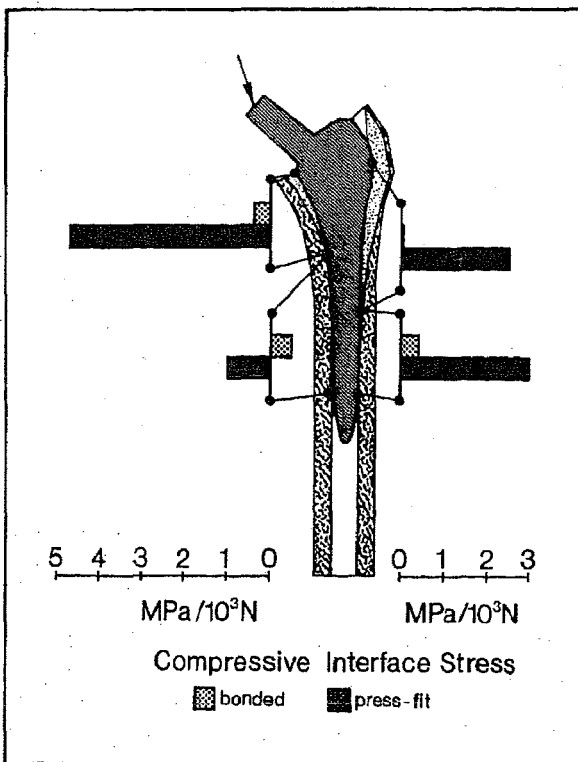
**Fig 15:** Average compressive interface stresses in four regions (medial/proximal, medial/distal, lateral/proximal, lateral/distal), comparing bonded and press-fit stems of the Osteonics (A), Zweymüller (B) and Zweymüller SL (C) stems.



**Fig 15A.**



**Fig 15C.**



**Fig 15B.**

such a process, strain-adaptive remodeling simulation programs can be versatile tools.

In principle, stress-shielding effects can be reduced by reducing stem stiffness or by changing the implant bonding characteristics from osseo-integrated to press-fit. Both measures have adverse mechanical side effects, of which the extent depends on implant design. Low modulus stems reduce stress-shielding effects but increase proximal interface stresses, hence stronger implant/bone bonds would be needed to prevent problems. It is probable that the clinical problems which have been reported in relation with the Isoelastic (RM) prosthesis<sup>28</sup> are due to elevated proximal interface stresses.<sup>7</sup> It is not unrealistic to assume that truly isoelastic materials will become successful in total hip replacement only if interface bonds can be made as strong as bone itself.

The use of titanium instead of CoCrMn stems does not dramatically increase proximal interface stress, but the stress-shielding effects of titanium stems are only slightly less. From a mechanical point of view, however, titanium appears to be a reasonable material for noncemented hip stems, as opposed to cemented ones.<sup>2</sup>

The press-fit stem does reduce stress-shielding effects considerably compared with the bonded type. However, it is not without potential problems that may be worse than those it is meant to solve. First of all, the press-fit concept has a built-in potential for interface micro motions, which will be effectuated when interface shear stresses exceed

friction locally. As shown earlier, this will predominantly occur at the proximal and distal sides of the stem. Distal motions causing midthigh pain are probable, especially when the distal stem is stiff, as in the PUP prosthesis. In addition, press-fit prostheses may generate elevated interface compressive stresses, possibly causing interface necrosis, resorption, and subsidence. If that occurs, a situation of a proximal stress bypass may be created, causing bone to resorb proximally in a manner similar to that associated with a bonded stem.<sup>26</sup> As found in the present analysis, the actual compressive stress values depend greatly on the actual shape of the prosthesis, whereby small details may have notable effects. The precise design of press-fit stems is, therefore, rather critical. Since prostheses coated for bony ingrowth or osseointegration usually begin as press-fit devices in the immediate postoperative phase, their design is critical as well. Obviously, the interface compressive stresses will also depend on friction and on the actual interface contact configuration.<sup>9,24</sup>

In summary, design and materials choice for noncemented THA stems is subject to conflicting design criteria relative to chances for proximal bone atrophy and interface stability. It appears that a noncemented stem should be neither very stiff nor very flexible; it should be coated, but not over its full surface; and it should be tapered instead of straight, but in a very subtle way. Obviously, these three most important aspects of stem design (material stiffness, coating geometry, and stem contour) act in concert relative to the load-transfer mechanism. Hence, their characteristics should be in tune.

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