Biomechanical aspects of hydroxylapatite coatings on femoral hip prostheses

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Biomechanical Aspects of Hydroxylapatite Coatings on Femoral Hip Prostheses

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The presumed advantage of a hydroxylapatite (HA) coating on noncemented prostheses is its fast postoperative osseous integration. This produces strong and lasting bonds between implants and bone, suitable for transferring hip joint loads to the bone without interface failure or relative motions. The biomechanical questions related to this behavior, and its presumed advantages, are whether the early integration process is reproducible, whether relative interface motions due to early prosthetic loading may prevent it, and whether the eventual bonds are indeed strong enough to prevent long-term interface failure. Although the osseo-inductive capacities of HA coatings have been documented in animal experimental and human retrieval studies (1,2), it has also been shown convincingly that integration will not occur when relative motions are beyond certain threshold levels (3). The extent of these early motions depends on surgical, patient and prosthetic parameters, such as implant fit, hip joint loads, and mechanical implant characteristics, i.e., stiffness and interface friction properties. It is important to quantify the mechanical interrelationships and effects of these parameters on the osseous integration.

The same can be said about the probability of late mechanical interface loosening. The likelihood of interface failures depends on the balance between interface stress and interface strength. Whereas interface strength is a function of the biologic bonding characteristics, the HA resorption characteristics and the bonding strength between coating and implant, interface stress depends on surgical, patient, and prosthetic parameters, similar to those that determine the early relative interface motions mentioned above. These effects, too, require quantification. Although it is now often suggested that HA coatings resorb over time, whether this will cause late loosening depends equally on surgical and design factors, because it is not only the interface strength but also the stress that matters.

In addition to interface biomechanics, the long-term behavior of the reconstruction is governed by adaptive bone remodeling. The stresses and strains that occur within a bone depend on its external loads, its shape, and its internal structural organization. This implies that when a part of the bone is replaced by an implant of different mechanical properties, the stresses and strains within the remaining bone change, even if the external loads remain the same. In accordance with Wolff's law, a process of strain-adaptive bone remodeling then takes place, changing the shape and internal structural organization of the bone to adapt to the new mechanical requirements. Although this adaptive remodeling is obviously an important biologic asset, it does not necessarily have positive effects when implants are involved. The reason for this apparent contradiction is simply that the implant does not adapt with its host bone.

A notorious adverse effect of adaptive bone remodeling is resorption around femoral hip stems. After a stem is placed in the intramedullary canal of a femur, two important changes occur in the load-transfer mechanism (4). First of all the hip joint load is no longer transferred downwards through the metaphyseal trabecular structures and the cortex but now involves the implant–bone interface. Second, the load that was earlier carried by the bone alone is now shared with the stem. This phenomenon, called "load sharing," causes "stress shielding"of the bone; i.e., the bone is shielded by the stem from the stress to which it is normally subjected. As a result, the bone stresses become subnormal and the bone resorbs to adapt to this new situation. Hence, contrary to common usage, "stress shielding" is not synonymous with bone resorption but rather is its cause.

An example of the stress-shielding mechanism, as it can be determined in finite-element models, is shown in Fig. 1. The bone stresses are shown here as they occur in reconstructions with a noncemented femoral stem, relative to what they would have been preoperatively case for the same external loads. The stress shielding is evident and clearly reduces from proximal to distal. Below the tip of the stem the stresses are again normal. Stress shielding is more severe for noncemented stems than compared for cemented ones (5). This is mainly due to the difference in flexibility of the two reconstructions. Because noncemented stems are bulkier, and therefore stiffer than cemented ones, they remove a larger share of load from the bone and thus create more stress shielding. It is to be expected that, as a consequence, more adaptive bone resorption will also be seen around noncemented stems, and this is indeed

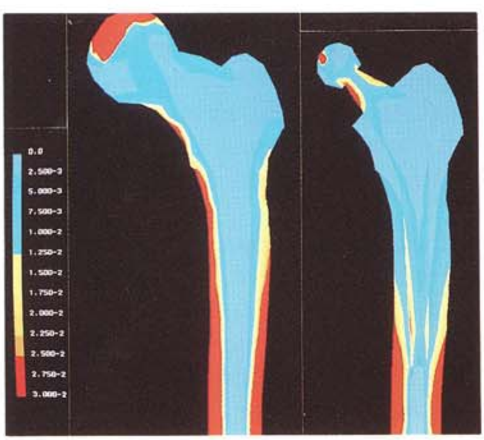

FIG. 1. Stress shielding around a noncemented stem; shown are strain-energy-density patterns (related to stress patterns) in the bone before and after the replacement, for the same loading case.

generally observed in the clinical setting.

Stress shielding and bone resorption around noncemented stems have worried clinicians for a long time, particularly because it is not known where and when this slow process ends. Its adverse effects may be that the prosthesis breaks out of the bone when its holding power is reduced below a certain minimum and the patient makes an unfortunate step, or that not enough bone stock is available in case a revision is needed. However, very few problems of this kind have been reported in the literature.

The degree of stress shielding depends not only on stem flexibility but also on other mechanical factors. As a result, the amount of bone resorption is also multifactorial. Table 1 gives an overview of prosthetic design, surgical and patient parameters that can potentially affect the extent of long-term bone resorption around noncemented stems. The purpose of this chapter is to investigate these relationships, using strain-adaptive bone-remodeling analysis. Because of their important implications for osseous integration and risk of interface failure, the effects of these parameters on interface mechanics are also discussed.

TABLE 1. *A few selected prosthetic, surgical, and patient factors that influence adaptive bone resorption and interface mechanics around femoral stems*

Design Factors	Surgical Factors	Patient Factors
Stem shape	Interface fit	Bone quality
Stem stiffness		Bone reactivity
Coating placement		Weight
Features		Activity

WOLFF'S LAW

The concept of strain (or stress) adaptive bone remodeling was first emphasized in the literature of the last century, a development that culminated in what we now know as "Wolff's law." Although the implications of this "law" are basic assets of orthopaedic surgery and most orthopaedic surgeons have acquired an intuitive appreciation of its significance, the law itself and its scientific background are hardly known at all.

Wolff's law (6) is not a law in the sense of a quantitative, falsifiable statement in line with the tradition of the physical sciences, but rather consists of a series of observations. The most important of these are the "trajectorial hypothesis" and the concept of adaptive remodeling (or "Transformation" in the original German). The first one was based in particular on the work of the anatomist Meyer and the engineer Culmann, who discovered a remarkable similarity between the trabecular structure of the proximal femur and the patterns of stress trajectories calculated in a mathematical model of this structure, using the new theory of graphic statics developed by Culmann. As argued by Roesler (7,8), it is evident from his text that Wolff did not really understand the mechanical implications of that similarity, and hence did not interpret it correctly. The second observation, adaptive remodeling, had been discussed extensively in the earlier writings of Roux (9). Dibbits (10) convincingly argued, based on an extensive review of the literature from those days, that Wolff had never accepted Roux's adaptive remodeling theories earlier and maintained that bone is a static entity once it is formed by interstitial growth. He refused to acknowledge or appreciate advances in histology and biology, such as indications that bone is subject to continuous resorption and formation (10). Nevertheless, he adopted Roux's ideas in his 1892 book and became known later as the conceiver of the law. This "law"– Wolff called it a mathematical law, which it definitely is not–he summarized in five parts. The first two concern the trajectorial theory (incorrectly interpreted, as Roesler wrote). The next three state roughly that the internal architecture of a bone remodels after pathological alterations of its external shape have occurred (he shows a number of examples from his surgical practice and postmortem anatomic studies in his book), that bones have "functional shapes" in both normal and pathologic conditions, and that the "remodeling force" can be used for therapeutic ends. It may have been due to the fuzziness around its emergence or the unclarity of its form that it has taken such a long time before the "law" and the challenge contained in its last statement were addressed scientifically. In 1881, Roux suggested that the adaptive remodeling process was governed by a "quantitative self-regulating mechanism"(9), nothing else, Roesler wrote, "but what nowadays would be described as a biological control process" (8). Although the existence of such a process had never been denied, the presumption of a load, stress or strain mediator for such a process was generally accepted, and the mathematical tools to analyze such a process were available, it was "...not until the late seventies of this century that Cowin and co-workers (11, 12) proposed a first quantitative form of 'Wolff's Law'..." (8). Much progress has been made since then, particularly owing to the combination of finite element analysis with mathematical remodeling rules (13-15).

Strain-Adaptive Bone-Remodeling Analysis

Strain-adaptive bone-remodeling analysis takes Roux's concept of a quantitative self-regulating mechanism or biologic control process, as a basis, according to which bone cells locally appraise loads and mediate bone formation and resorption. A schematic representation of this model is shown in Fig. 2A. The sensor cells, we assume, are the osteocytes (16), and the actors the osteoclasts and osteoblasts, although these assumptions are not critical for the remodeling theory. The sensors measure a strain-related mechanical signal and compare that with a normal reference value. If the signal is too high, the sensor mediates the actors to form bone; when it is too low, bone is resorbed. This process continues until the mechanical signal is again normalized. The values of the signal in each location of the bone depend on the external loads and the mechanical properties of the bone, i.e., on its shape or geometry and on its internal structural organization or architecture. While the remodeling process is enacted, and shape and architecture are changing, the signal values change as well, providing the feedback control loop for the sensors which govern the process.

The computer model used to simulate this process is illustrated schematically in Fig. 2B. Stresses and strains are determined in a finite element model of the bone, representing its shape and architecture (i.e., density patterns) and simulating its external loads. The model is then used iteratively whereby, during each iteration, the mechanical signals (S) are determined from the stresses, strains, volumes, and masses in each element and compared to reference values (S_{ref}). Using a mathematical remodeling rule, the local amounts of bone mass per element to be formed or removed are calculated and adjusted in the finite element model by changing the element volumes or densities.

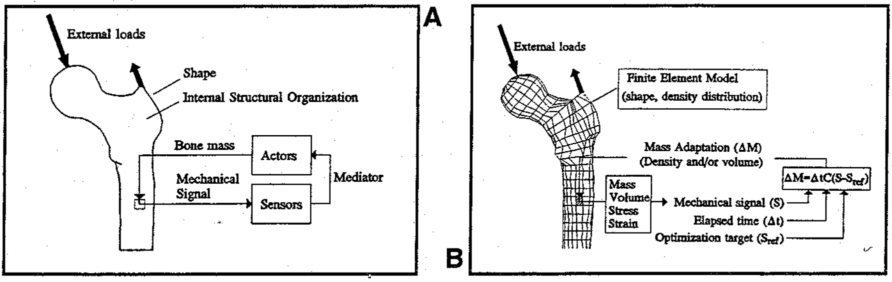

FIG. 2. Adaptive bone remodeling can be considered as a local biologic control process, governed by a mechanical signal, and appraised by sensors (osteocytes), that mediate actors (osteoclasts and osteoblasts) to regulate bone mass (**A**). Such a process can be described by a computer-simulation model in which a mathematical remodeling rule is coupled to a finite-element model (**B**).

Of course, the values of many of the parameters needed in such a computer-simulation model are unknown or uncertain. Of some quantities and relationships, e.g., the remodeling signal and the mathematical remodeling rule, we do not even know the character. This problem is approached in a way typical for modeling in physical sciences, by trial and error. First, sensible assumptions are made for quantities,values and relationships. These are then tried in the computer-simulation model relative to remodeling configurations of which the solution is known. The theoretical and real solutions are compared, and if they do not match the model parameters are adjusted accordingly until we are satisfied that its predictions are valid. In this way we have triggered and verified our model and its parameters relative to the density distribution of the normal femur (17,18), as illustrated in Fig. 3, and to three series of canine experiments with different types of hip prostheses (19-21), illustrated in Fig. 4. We are now confident that the model and its predictions make sense.

We use the elastic energy stored per unit of mass in the bone by the external loads as the remodeling signal, calculated as the product of the stress and strain tensors determined in the finite-element procedure, divided by the actual density. As illustrated in Fig. 3, this quantity gives an excellent representation of local bone loading, to the extent that normal bone density patterns can be predicted in bone growth and maintenance analyses (17, 18). The mathematical remodeling rule, in which the signal values are compared to their normal references, is a nonlinear one, in the sense that a threshold level, or dead zone, for bone reactions to abnormal loads is adopted (15). This implies that, locally,

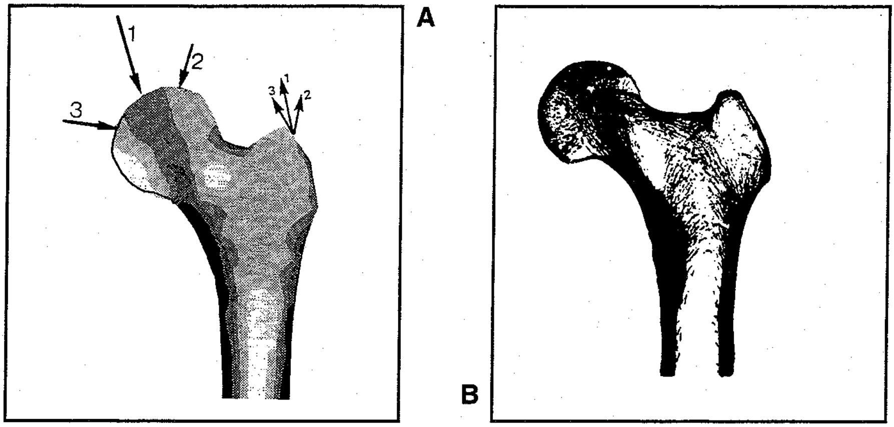

FIG. 3. Using the computer-simulation model illustrated in Fig. 2A, the normal density distribution of the femur, according to Wolff's law, can be predicted (18). Density distribution of a computer simulation (**A**) compared with the density distribution in a midfrontal section (**B**).

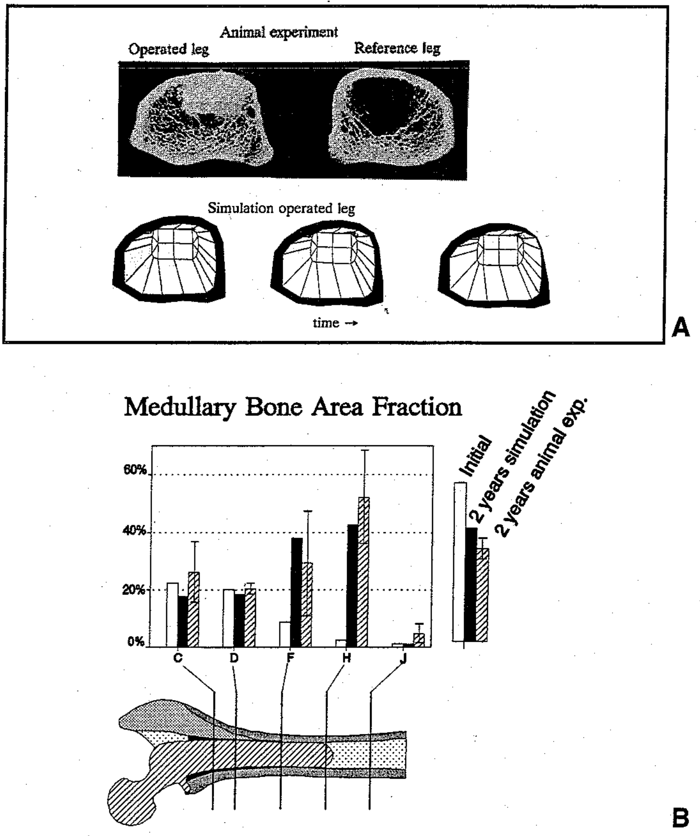

FIG. 4. By simulating animal experiments with canine hip replacement, the postoperative morphologic changes in cortical area and intramedullary density due to adaptive remodeling could be predicted, both in qualitative detail (**A**), and as quantitatively relative to averages and variations in an animal series (**B**) (19,20).

bone must be under or overloaded by at least a certain percentage before it reacts. This percentage depends on bone reactivity, and was established at average values of 35 for the dogs (19) and 75 for humans (22). This dead zone represents the "mean effective strain" (MES) concept of Frost (23).

For the reference signal values we use the distribution of elastic energy per unit of mass as it occurs in a normal bone subjected to typical loads. The procedure for simulating bone remodeling around hip stems is then illustrated in Fig. 5: finite-element models are made of the intact femur and the same femur with a prosthesis, which are subjected to the same external – hip and muscle – loading cycles. The model with the implant is subjected to the remodeling simulation procedure, whereby the element signal values after each time step are compared to those in the intact model, and the element-density values are adjusted accordingly for the next time step. This process continues until the signal values in the replacement model are again equal to those in the intact one, minus the threshold level. Some elements will not reach that stage because they have either resorbed completely in the process or reached the maximal density value of cortical bone. An example of such an end-stage density configuration is shown in Fig. 6 (24). The resorption patterns, particularly at the proximal side around this fully bonded titanium prosthesis, are evident, and are very similar to what can be seen on long-term postoperative radiographs.

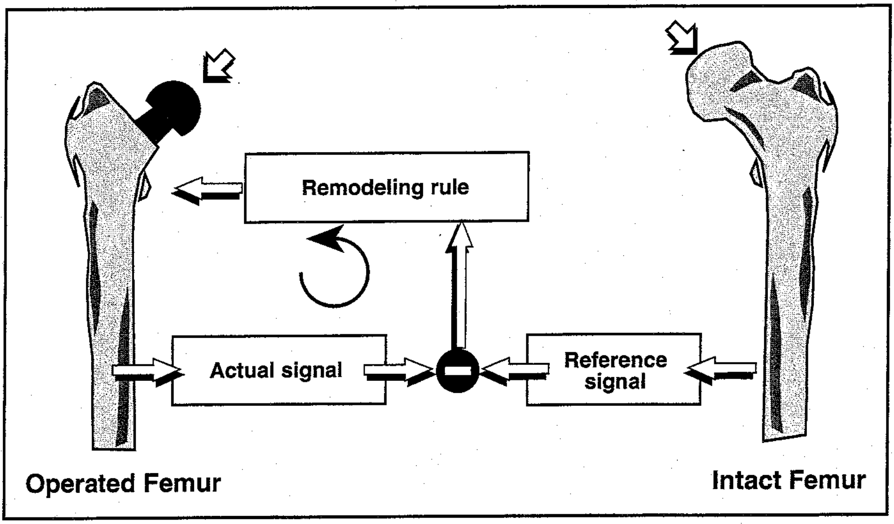

FIG. 5. In the analyses presented here, two finite-element models are applied. One, of the intact femur, provides for the reference values of the mechanical signal. In the other, with the prosthesis, bone density is gradually adapted to equalize the actual signal values to the reference ones. When this is accomplished, a new equilibrium has been established.

The Effects of Prosthetic, Surgical, and Patient Factors

The bone-remodeling computer-simulation model is a versatile tool, particularly to investigate how remodeling and resorption patterns relate to all kinds of parameters, such as those listed in Table 1. The model is then used in a relative fashion, comparing one parameter value to another while other parameters remain unchanged. In this way the pure effects of a single factor can be evaluated, and the imprecisions inherent to models become of lesser importance. Some examples are shown in the remainder of this chapter, whereby effects of parameters on bone resorption and on interface mechanics are discussed. Bone resorption patterns are presented in the Gruen zones 1 through 7 (25), illustrated in Fig 7.

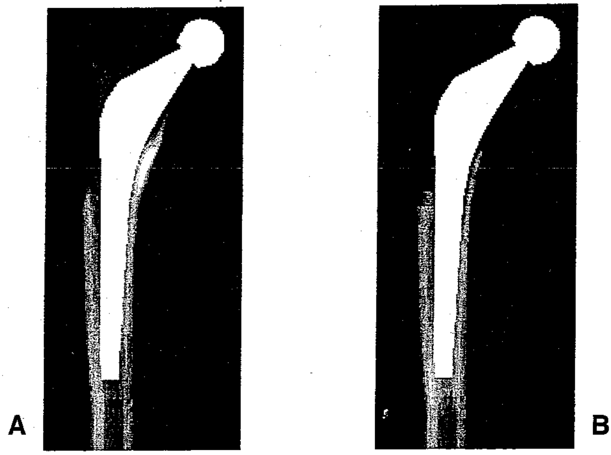

FIG. 6. Result of a remodeling analysis as in Fig 5. Both pictures show density distributions in a three-dimensional finite-element model, projected in the frontal plane, simulating a radiograph; (**A**)the situation before remodeling has started (immediately postoperative), (**B**) after a new equilibrium has been established. Bone resorption, predominantly proximally, is evident.

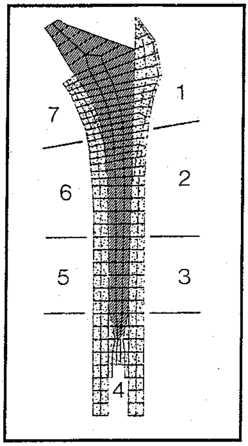

FIG 7. The Gruen zones (25) in which the bone-remodeling patterns are quantified in the parametric analysis (Tables 2-4).

Stem Stiffness

The stiffness,or flexibility, of a stem depends on its shape and dimensions, and on the elastic modulus of its material. Bobyn and co-workers (26) performed experiments with dogs, implanting two kinds of hip stems with different stiffnesses, one made out of solid titanium and the other hollow. After two years the amount of bone resorption was evaluated postmortem and was shown to be much less around the flexible stem, thus confirming the assumed relationship between stiffness, stress shielding, and bone resorption.

In our computer-simulation analysis we look at the effects of material stiffness only; therefore stem shape and dimension remain constant. In each case the same external loading cycle is assumed and the stem is taken as fully bonded (osseointegrated) to the bone along its entire length. The only difference is the material, which is either a cobalt–chromium alloy (elastic modulus 210 GPa), titanium (elastic modulus 110 GPa), or a hypothetical "isoelastic" material with a similar elastic modulus as cortical bone (20 GPa). The results of the simulation analyses for these three cases are presented in Table 2 (27). Evidently, the effects are quite extensive. The drastic reduction in resorption from cobalt–chromium to titanium material (a reduction in stiffness by a factor of about two) from 76% to 54% resorption at the proximal, medial side, illustrates that titanium is a much better material for the bulky, noncemented prosthetic stems. The resorption reduces almost to nil in the case of an isoelastic material.

The results listed in Table 2 relate to analyses with two-dimensional FE models (27), but have also been confirmed in three-dimensional ones (22). In that case, for instance, it was found that the resorption reduces from about 67% to about 34% in the bone around the proximal third of the stem, when a change is made from titanium to an isoelastic material. Although 2-D and 3-D results are not well comparable in an absolute sense, relatively speaking the same trends were found.

TABLE 2. *Bone loss due to adaptive remodeling for three different stem materials (27)*

Stem material (Elastic Modulus, GPa)		Proximal		Middle	
		Medial	Lateral	Medial	Lateral
CoCrMo	(210)	-76%	-45%	-29%	-32%
Titanium	(110)	-54%	-38%	- 4%	-14%
"IsoElastic"	(20)	-7%	-1%	0%	+2%

Hence, it is evident that a flexible stem is advantageous for reduction of bone resorption. However, in what way does it affect interface mechanics, the process of osseous integration, and the probability for late interface failure and loosening? Because so little is known quantitatively about the integration and resorption processes of the HA coatings and their relationships with interface strength, these questions are not easily answered at present. However, it is obvious that the beneficial effects of flexible stems on proximal bone resorption are the result of higher load transfer at the proximal side. This implies higher interface stresses proximally, and more chances for initial interface motions. Figure 8 shows a relative comparison between the amounts of stress shielding and maximal proximal interface stresses for stem moduli between 20 and 110 GPa (22). When stiffness reduces, stress shielding reduces as well, but interface stresses soar. Conversely, the 110 GPa (titanium) stem produces more stress shielding but only half the interface stresses of the 20 GPa (isoelastic) stem. Although this has not been documented here, it is fairly obvious that the effects on relative interface motions are very similar, in the sense that they are higher for the flexible stems. Therefore, although flexible stems have definite advantages for bone resorption, without additional means of initial stability and enhancement of interface strength it is questionable whether they would really be clinically successful.

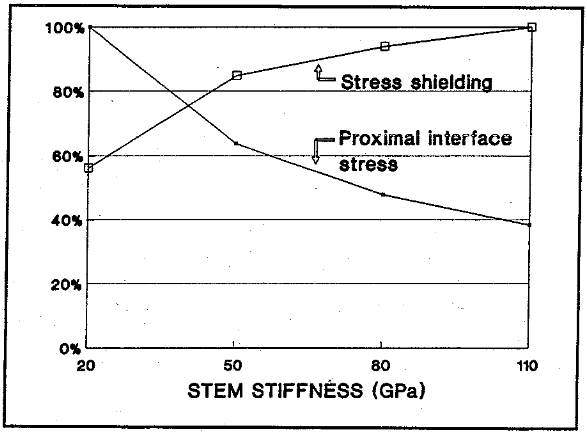

FIG. 8. When stem stiffness decreases, stress shielding also decreases, but proximal interface stresses increase; hence, although flexible stems produce less resorption, they increase the likelihood of interface failure and motions (22).

Coating Placement and Integration

There are different philosophies about the placement of coatings. Coating the surface over the full length of the stem, enhancing the probability of integration, is one; limiting the coating to the proximal part only, enhancing proximal load transfer and reducing stress shielding, is another. There are two questions to be answered before a sensible choice between these alternatives can be made. The first concerns the relationship between coating location and the reproducibility of integration, of which, obviously, very little is known as yet. The second question, addressed here, concerns the relationship between ingrowth location, stress-shielding and interface mechanics. Evidently, in reality, the bonding patterns over the stem surface can be quite diverse, ranging from true biochemical integration to essentially unconnected.

In the analyses we disregard the subtleties and assume either a solid bond or a contact without friction between bone and implant. The first condition occurs, in the model, at the locations where the surface is coated; and the second prevails where the surface is not coated. The configurations thus analyzed and compared are full stem-length coating, one-third proximal coating, small-band proximal coating, and uncoated (press-fitted), as illustrated in Fig. 9. Again, all other parameters remain the same, i.e., identical loading cycles and a titanium stem. The results of the simulations are shown in Table 3 (24). Relative to the fully bonded stem (which is the same case as the titanium stem in Table2), one-third proximal coating reduces the amount of bone resorption at the proximal sides from 54% to 50% and from 38% to 22%. This is significant, but not so dramatic as sometimes hoped for. We see more reduction of bone loss around the midregion of the stem, which is caused by stress-transfer concentrations occurring at the edge of the coated region, as shown in Fig. 10. If coating is assumed only as a small band under the resection plane, we see a dramatic reduction of bone loss, e.g., to 18% and 13% at the proximal sides.

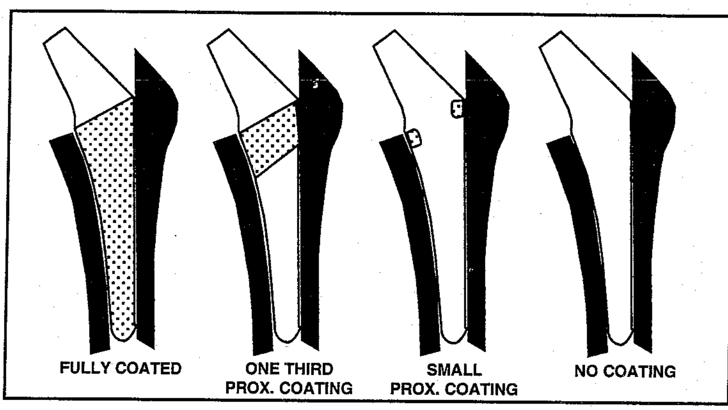

FIG. 9. Four different coating configurations, from left to right: a fully coated stem, a one-third proximally coated stem, a small proximal coating, and no coating at all (24).

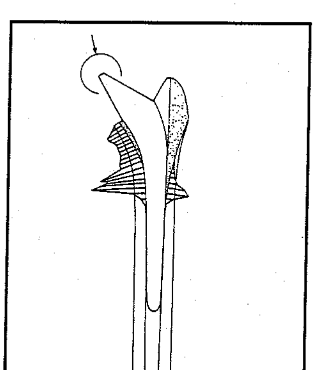

FIG. 10. Shear stresses at the stem–bone interface for a proximally coated stem; stress concentrations occur precisely at the edge of the coating, enhancing bone formation (Reproduced with permission from Huiskes R, Weinans H, Dalstra M. Adaptive bone remodeling and biomechanical design considerations for noncemented total hip arthroplasty. Orthopedics 1989;12:1255-67).

When we assume no bonding at all (Table 3), we do not obtain further reductions relative to the third case of the small proximal coating. When we compare the unbonded case with the one of the proximal one-third coating, we see that bone loss reduces from 50% to 35% in the proximal–medial region, and from a 5% loss to a 3% gain in the midlateral region, but no reductions are seen in the other regions. This confirms that the load-transfer mechanism of unbonded, press-fitted stems is very different from that of bonded ones, even if the bonding occurs in a small area only (5). What makes them so different is the fact that the unbonded stem subsides elastically when loaded, thereby stressing the interface, while the bonded stem is held in its position.

TABLE 3. *Bone loss due to adaptive remodeling for four different coating configurations, illustrated in Fig. 9 (24)*

Bonding characteristics	Proximal		Middle	
	Medial	Lateral	Medial	Lateral
Fully coated	-54%	-38%	-4%	-14%
One-third prox. coated	-50%	-22%	0%	-5%
Small-band prox. coated	-18%	-13%	-5%	-4%
Uncoated	-35%	-26%	-13%	+ 3%

In summary, coating placement and bonding conditions have distinct effects on stress-shielding and bone resorption. Relative to a fully bonded configuration, bone resorption can be limited either by concentrating the coating on the proximal side or by having an uncoated smooth surface instead. In the first case, load transfer is concentrated at the proximal side and, as in the case of the flexible prosthesis discussed earlier, this implies not only less stress-shielding but also higher proximal interface stress concentrations, hence a higher probability of interface failure. It is obvious that in the case of the small proximal coating band (see Table 3), these stresses may become excessive (24). Again, a compromise must be found between acceptable resorption and acceptable loosening risks, by applying a coating that is located proximally but is still extensive enough to carry the interface loads.

The second way of limiting resorption relative to the case of full bonding, by press-fitting an unbonded prosthesis, also has its vagaries for interface mechanics. Because of its lack of bonding, the implant moves (subsides) relative to the bone each time the hip joint is loaded (5). Although this relative displacement stresses the surrounding bone, thereby reducing the amount of stress-shielding, it also produces continuous interface motions which may eventually cause interface–bone resorption and implant loosening.

Implant Fit

Implant fit is considered a surgical parameter, although of course its precision and reproducibility can also be influenced by the design of stem and instrumentation. To evaluate the effect of this factor we compare three different configurations (Fig. 11): one with a precise (line-to-line) interference fit of the stem over its full length, one in which there is a 1 mm gap at the distal interface–hence, the case in which the distal bone has been overreamed by 2 mm, or the distal stem has been undersized by 2 mm – and one in which there is a proximal gap of 1 mm, hence proximal overreaming or undersizing by 2 mm. Again, all other parameters remain the same, i.e., the same loading cycle, titanium stem material, and no interface bonding. Thus, the first case is the same as the last one in the previous analysis.

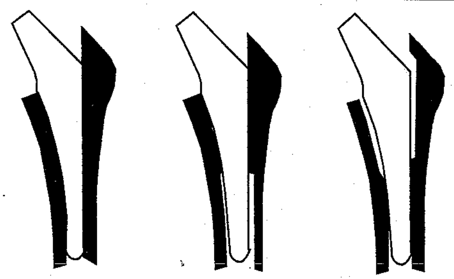

FIG. 11. Three different configurations of fit, from left to right: line-to-line interference fit, distal gap (overreamed or undersized), and proximal gap (24).

The results of the remodeling analysis are shown in Table 4 (24). Relative to the precise interference fit, the distal gap causes load to be transferred more proximally, reducing bone loss to about the same amounts as the small-band proximal coating case in the previous analysis (seeTable 3). Conversely, proximal overreaming or undersizing produces dramatic amounts of bone loss, up to 91% in the proximal–medial region. This is the worst case of resorption in our entire series, and is a nice illustration of the importance of surgical technique relative to implant design.

TABLE 4. *Bone loss due to adaptive remodeling for three different cases of fit, illustrated in Fig. 11 (24).*

Fit characteristics	Proximal		Middle	
	Medial	Lateral	Medial	Lateral
Line-to-line fit	-35%	-26%	-13%	+3%
Distal overreamed	-18%	-13%	+1%	+6%
Proximal overreamed	-91%	-48%	-34%	-7%

Patient Factors

Many patient factors can interfere with the bone-remodeling process, systemic and local, behavioral, physiological, or pathologic. We have summarized all factors in relation to their effects on three parameters (Table 1), hip joint loading, bone physical quality, i.e., density and stiffness, and bone reactivity, by which is meant the threshold strain deviation, or mean effective strain deviation, to which the bone reacts, represented in our models by the width of the dead zone.

Of these three parameters, we have not analyzed the effects of hip joint loading, probably because it seems so trivial. Obviously, increased hip joint loads increase the bone stresses proportionally. Because the bone-remodeling process is nonlinear–or so it is assumed in our models–the reduction of bone resorption is not necessarily nonlinearly proportional with the increase of load. In any case, there will be a reduction. However, there are also increases in relative interface motions immediately postoperatively, and perhaps a less effective integration process, with higher stresses and chances for interface failure, late postoperatively.

The effects of the other two patient parameters were evaluated relative to a three-dimensional finite-element model, based on a bone of which the normal density distribution was determined from CT scans (22). A fully-bonded titanium stem was inserted in the model, and the long-term remodeling patterns were determined with the simulation analysis. The results for this (reference) case are illustrated in Fig. 12, in which the resorption patterns are evident. To simulate a denser, hence also stiffer, bone, the normal preoperative density values measured were multiplied by a factor of two, albeit keeping the maximal density at the initial value of 1.73 g/cm^2, representing cortical bone. The result is a stiffer bone, for which the simulation analysis was repeated, keeping all other parameters the same. In the third analysis the original bone-density values were again used, but this time the threshold level for bone reaction was reduced from 75% to 35% of the normal signal values, thus simulating a bone with about twice the degree of reactivity.

The results are shown in Table 5, this time for four regions in the bone around the prosthesis, at the level of the Gruen zones 1 plus 7, 2 plus 6, 3 plus 5, and 4, respectively. Notable is the drastic reduction in resorption patterns for the stiffer bone, e.g., from 67% to 11% in the most proximal region, and the drastic increase for the bone with the higher reactivity, e.g., from 67% to 82% in this same proximal region.

Because these parametric variations are both rather hypothetical in extent we do not really know the real variability in bone stiffness or in bone reactivity, therefore it is not really possible to interpret them in terms of actual patient conditions. However, these examples nicely illustrate three important general points. The first is that in clinical series the extent of bone resorption will vary, even if prosthetic and surgical factors are equal in all cases, because of differences in patient factors. This may sound trivial, but it does suggest that to optimize results of joint replacements these patient factors should, ideally, be somehow evaluated preoperatively. The second point is that the younger patients, with presumably higher bone reactivities, could experience more postoperative bone loss, even if they are more active than older ones. The third point is probably the most important. The reduction in amounts of bone resorbed in the case of the stiffer bone indicates that bone stiffness is at least as influential a parameter as stem stiffness. This makes sense, of course, because

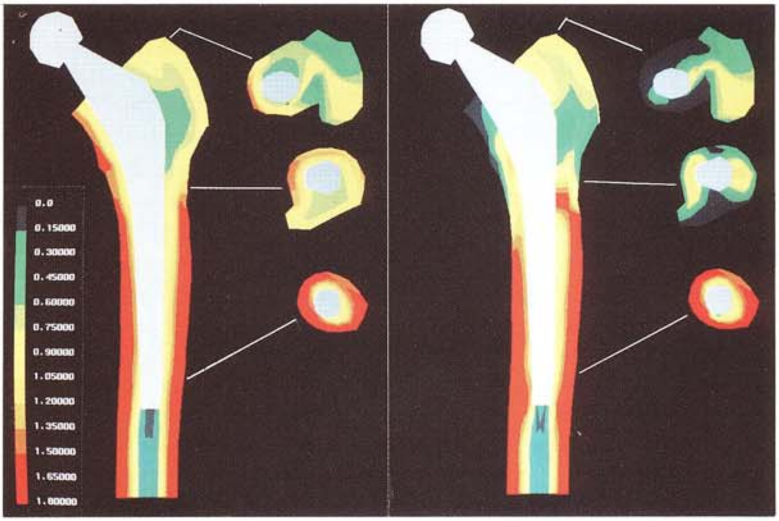

FIG. 12. Immediate postoperative density distribution, as based on CT scan, on the left, and density distribution after long-term remodeling simulation on the right, corresponding with the "normal" bone in Table 5 (22).

TABLE 5. *Bone loss due to adaptive remodeling in three different kinds of bones, a "normal" one (see Figure 12), one with a higher bone density and one with a higher reactivity (22)*

Bone characteristics	Proximal	Upper middle	Lower middle	Distal
"Normal"	-67%	-35%	-4%	+5%
Dense (stiff) bone	-11%	-3%	-1%	+1%
High reactivity	-82%	-64%	-27%	0%

the load-sharing mechanism of stem and bone is governed by the ratio of stem and bone stiffnesses rather than by each separately (28). Recently, Engh and co-workers (29) published a report of bone-density measurements around hip stems in retrieved postmortem specimens, using dual-energy x-ray absorptiometry analysis, which produces accurate estimates, contrary to conventional radiographic measurement techniques. They compared results to those obtained from the contralateral femur, which was assumed to represent a density distribution similar to that of the treated femur preoperatively. They found 7% to 52% bone loss around the stem relative to the contralateral bone, with the largest part–a range of 30% to 80% and a mean of 45%–in the proximal region. They also found a good inverse correlation between the amount of bone loss in the treated bone and the density of the contralateral one (Fig. 13), thus confirming the relationship found in the model. Since it is possible in principle to estimate the density of bones preoperatively, this presents a means of customizing implant stiffness to the patient.

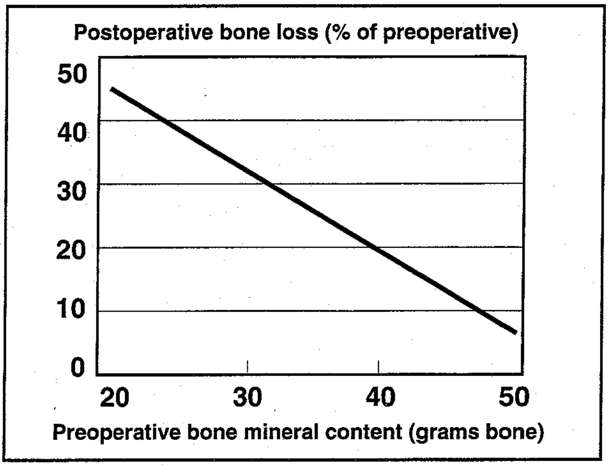

FIG. 13. Illustration of results obtained by Engh and co-workers (29) from precise radiodensity measurements in five postmortem specimens. An inverse correlation was found between postoperative bone loss around noncemented femoral stems and preoperative bone density, estimated by measurements of the contralateral bone. These findings confirm the predictions of the computer-simulation model (Table 5) in a qualitative sense.

Discussion

The simulation model we developed is nothing more than a simple mathematical description of the concept of bone as a "quantitative self-regulating mechanism," suggested by Roux in 1881 (9), combined with finite-element analysis to make it applicable to the morphologic complexities of bone structures. Although the finite-element models are still crude relative to these structures and the remodeling model features a number of assumptions, the results it produces are surprisingly realistic. As said, the morphologic adaptations around canine prostheses in a number of experimental series could be predicted in detail (19-21), but predictions for human cases also are confirmed in the clinic. Earlier, these predictions had been called unrealistic, based on information from postoperative radiographs. However, these conventional radiographs are very imprecise. Recent radiographic studies, using precise objective measurement methods, have shown that the extent of bone resorption around hip stems predicted, on the order of 30% to 80%, is also found in reality (29-31). Effects of implant stiffness on bone resorption, predicted by the model, are similar to those found in animal experiments (26, 32).

The simulation model is particularly useful to analyze and explain the quantitative relationships between the morphologic changes and the various prosthetic, surgical and patient factors that affect them. Generally speaking, the results of the analyses performed for that purpose and discussed in this chapter are not surprising. Anything feasible that increases the external loads on the joint or lets them be transferred to the bone as far proximal as possible, such as flexible stems, small proximal coatings, and distal interface gaps, increases proximal bone stresses, reduces stress shielding, and diminishes bone resorption, but at a price. All these measures also tend to increase early interface motions, hence less effective osseus integration, and interface stresses (hence a higher probability of late interface failure and loosening). It seems that we are caught between a rock and a hard place, and compromises are required. Since stem loosening is a real problem and bone resorption only a potential one (remember that clinical problems as a result of it have not been reported as yet), it would probably be wise to stay on the conservative end where it concerns interface stress enhancing techniques or designs. Of course, conflicting design requirements are great candidates for computer optimization, and these methods will certainly be used in the near future, in combination with the simulation models discussed here.

Not all parameters were considered here, an important one of which is the shape of the stem, which was more or less similar in all analyses. Neither were design features, such as a prosthetic collar, analyzed. These factors can certainly have extensive effects on both bone resorption and interface mechanics.

Effects that were surprising, at least to us, are the dramatic influences of proximal fit and bone initial density (or stiffness) on the amount of bone resorption. It seems that these are parameters that can successfully, and safely, be considered to minimize bone loss. The degree to which uncoated, press-fitted stems limit resorption relative to proximally-coated ones is much less than expected and, considering the interface motions they inherently provoke, one wonders if they should be applied at all. Somewhat disappointing was the moderate effect of proximal one-third coating relative to full coating on bone loss at the proximal side. The gain is predominantly in the midregion of the stem, because a relatively large share of the load is transferred near the distal edge of the coating. Although this is a gain nevertheless, a more subtle coating configuration might be worth considering. Finally, the effects of flexible, so-called isoelastic materials were, to us, all but surprising. However, the risks of excessive interface motions and stresses that these materials present can not be emphasized too often, particularly since a number of companies consider them experimental at present.

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