Failed innovation in total hip replacement

Diagnosis and proposals for a cure

Rik Huiskes

Many new hip prostheses and fixation techniques have been introduced in orthopedics in recent years. Yet, none have provided superior total hip replacements (THR) in comparison to the traditional cemented Charnley concept, which nevertheless has limited long-term endurance. This review article investigates why the THR innovation process has failed.

The predominant causes for long-term failure of THR are discussed. A framework of generic failure scenarios is proposed to provide guidelines for a scientifically-oriented approach to THR design, testing and clinical evaluation. It is shown that THR components are subject to incompatible design goals as regards prevention of the different failure scenarios. Neglect of those has been one important factor in the present innovation impasse. A second factor is the trial-and-error culture in orthopedic surgery, in which new devices run through the innovation cycle without proper testing or rigorous postoperative analysis. The third factor is ineffective regulation of marketing approval with respect to orthopedic implants.

Scientific research in orthopedics and related sciences has produced new methods for systematic design evaluation, pre-clinical testing and clinical trials. These can provide a basis for self-regulation and self-control by the orthopedic community, in all stages of the innovation process.

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After hundreds of thousands of operations during a quarter of a century the cemented total hip replacement (THR), based on Charnley's Low Friction Arthroplasty, has proven outstandingly successful. Although initial problems occurred due to stem fracture, inadequate cementing or malplacement, the introduction of high-grade metal alloys and improved cementing techniques, the emergence of a better understanding of failure mechanisms, and adequate instruction and training, have provided a dependable treatment for the seriously disabled. Its success can be measured by the results of multi-center patient-outcome studies (Herberts et al. 1989, Malchau et al. 1993), showing that all the orthopedic surgeons in a nation together, including the less proficient, could perform Charnley reconstructions in patients of whom, 10 years later, 93 percent were still essentially pain-free and functioning reasonably well.

It is, however, also true that "... in a sense, THR is now threatened by its own success ..." (Poss 1992), due to reductions in age indications. Although the 10-year results of THR are good, it has become clear that many will not endure another 10, that younger patients are subject to higher loosening rates and that revision procedures do not compare with the success of primary ones. As a reaction to these problems, enormous efforts have been devoted to the development of new materials, designs or techniques during the last 10–15 years, in the hope of improving the endurance of the Charnley concept. One may question, however, if these efforts have borne fruits. Perhaps the fruits are still pending, but not even a blossom is apparent as yet. Numerous new THR designs were marketed, none of which has a better clinical record than the Charnley. New fixation methods were introduced, many of which again silently disappeared, leaving disappointed surgeons and patients in their wake. The number of alternative designs, materials, fixation methods and surgical instruments for THR in use, many of which are of uncertain benefit, is staggering. The question whether these new devices are marketed for the sake of the patients or primarily to keep orthopedic companies afloat is occasionally posed (Sarmiento 1991).

When weighing the energy put into marketing efforts against the lack of improvement, is it surprising that surgeons lose faith in the innovation process itself? The issue whether it is at all possible to create a reconstruction that will last for a lifetime, also in the relatively young, or if wear and interface processes simply set natural time-limits to endurance, has

become a very valid one indeed. If there is such a natural endurance limit, should THR not be considered as a multi-stage treatment in principle? Should not postoperative patient-monitoring methods and better revision procedures be emphasized, rather than experiments with devices which may or may not extend the average endurance limit, but marginally? In view of the many adverse innovations marketed in the past, is it at all fair that patients participate in the experimental developments and contribute to what is so elegantly called a learning curve? Should they rather be protected against inventive surgeons? Should surgeons at large be protected against the folly of the day promoted by inventive surgeons, scientists and companies? Suppose that a new invention would significantly extend the endurance limit of THR, how and when will that be established? Herberts et al. (1989) estimated on statistical grounds that a series of almost 3,000 patients with a new device would have to be followed for 5 years, before it can be proven significantly better—that is, enduring longer—than the Charnley prosthesis. Which patient, surgeon or company wants to participate in such an endeavor? And who is still interested by the time the results are published? Or are there dependable methods for pre-clinical or clinical testing to substitute for long-term trials?

These are all very timely questions indeed. The field of joint replacement seems to have reached a turning point recently, an impasse (Bauer 1992). In knee replacement many recent design improvements have created more problems than they have solved, treatments worse than the disease they were supposed to cure. It is even suggested that the last 10 years of knee-prosthetic development have essentially been lost (Bauer 1992, Goodfellow 1992). In hip replacement a similar effect is apparent; many surgeons have returned to the traditional cemented concepts, after disappointing experiments with noncemented designs or other innovations.

I shall attempt to analyze why THR constructions fail. I propose a framework of failure scenarios, paradigms of failure mechanisms and guidelines for a more scientifically oriented approach to THR design, testing and clinical evaluation. Next, I propose an explanation for the impasse, hoping to convince the reader that the empirical trial-and-error culture in orthopedics at large, the disregard of incompatible prosthetic design goals and the lack of effective marketing-approval regulations have contributed substantially to problems in the innovation process. Finally, the scientific accomplishments in recent years and their potential applications in methods of pre-clinical testing and clinical trials are discussed.

Failure scenarios

There has been much discussion about when a THR can be said to have failed. From a clinical point of view, excessive pain and impaired function are indications for a revision, which by itself then defines a failure. However, it is also possible that radiological signs of loosening or migration already define failures, even if the patient is still happy. At the root of these differences of interpretation lies the fact that failure is virtually always a process, hardly ever an event. Such a process develops as a mechanism and produces radiological and clinical signs along its way. Understanding the failure mechanisms, preventing their occurrence and correctly reading their signs are priorities for progress. For that purpose I find it useful to define failure mechanisms in terms of scenarios, particular courses of events that may or may not occur, but are always latent.

The most common failures in cemented and noncemented reconstructions are of different origins. The experience with noncemented THR revealed clinical problems of a nature not earlier seen in cemented ones. When we discuss these problems in terms of failure scenarios, however, a separation between cemented and noncemented prostheses is less obvious.

Accumulated damage

The accumulated-damage scenario is based on gradual accumulation of mechanical damage in materials and interfaces due to repetitive dynamic loading. The damaging process proliferates eventually to disruption of the implant from the bone, interface micromotion, bone resorption and fibrous interposition, and finally gross loosening. This scenario was emphasized as a mechanism of loosening of cemented THR by Stauffer (1982) and played a dominant role in the concepts of THR endurance for many years (Figure 1). Recent investigations of retrieved specimens by Jasty et al. (1991) have actually shown cement damage. In noncemented THR, a scenario of this kind could jeopardize the long-term integrity of the implant-bone interface. Interface stresses may locally exceed the strength of the implant-bone bond and initiate disruption. Cracks may then grow under the influence of dynamic loading, producing interface motions, bone resorption and, eventually, gross loosening (Bragdon et al. 1991).

The likelihood of failures due to accumulated damage may be different for cemented and noncemented THR, but the underlying principles are the same. Mechanical failure is a matter of stress versus strength (Figure 2). Even a highly stressed material or bond does not fail if adequately strong, and even weak materials or bonds do not fail, if not stressed beyond

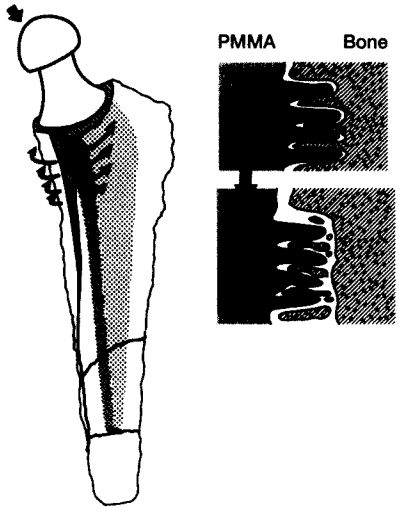


Figure 1. Stauffer (1982) emphasized mechanical cement and interface failure as the cause of long-term THR loosening.

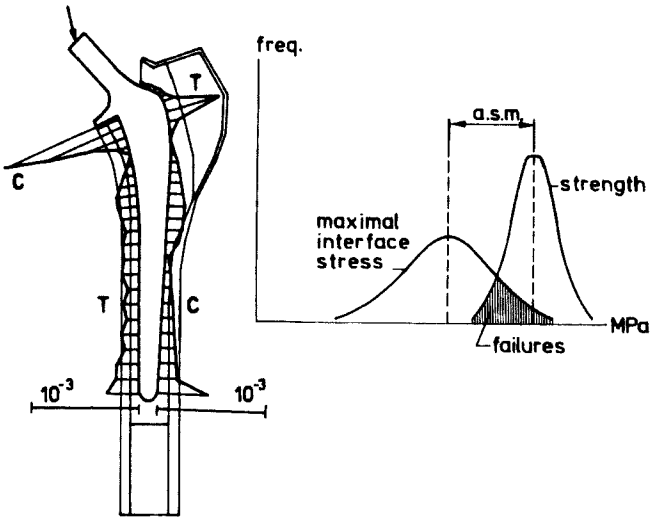


Figure 2. Left: tensile (T) and compressive (C) stresses along the stem-cement interface; peak values occur proximally and distally.
Right: mechanical failure is a matter of stress versus strength, which are both distributed stochastically in a patient population. The average safety margin (a.s.m.) determines the number of failures.

their limits. The stress patterns associated with load transfer in hip reconstructions depend on pure mechanical factors, such as component shape and material properties, interface bonding characteristics, bone mechanical properties and hip-joint loading (Huiskes 1991). Most of the load transferred from a well-bonded femoral stem to the bone is concentrated at the proximal and the distal sides, with relative magnitudes depending on the stiffness of the stem, i.e., stem thickness and elastic modulus (Figure 2). When a flexible stem is used, the highest interface stress peaks occur at the proximal side, whereas a relatively stiff stem produces the highest stresses at the distal side. This generic behavior is reflected in the different stress-transfer patterns of cemented and noncemented stems, since the latter tend to be much stiffer than the former. The interface stresses transferred from the cup to the acetabulum also follow a generic rule, in that a cup which is relatively flexible emphasizes stresses behind its central dome in particular, and a relatively stiff cup, such as a metal-backed one, does so at its rim (Huiskes 1993).

However, whether or not failure is initiated depends not only on the stresses, but also on the strengths of the materials and the bonds. In this respect in particular, cemented reconstructions are at odds in comparison with noncemented ones. Cement is a weak material relative to the stresses it is subjected to in the hip, and the same is true for the cement-bone and implant-cement bonds (Huiskes 1993). Conversely, the noncemented prostheses lack the weak cement link and the implant-bone interfaces can be relatively strong (Pilliar et al. 1986, Geesink et al. 1988, Sumner and Galante 1992). Hence, it is much more likely that the accumulated-damage scenario is initiated in cemented THR than it is in noncemented THR. It is not unthinkable, however, that this failure mode may become more marked in noncemented THR in the future, since interface bonding strengths may gradually weaken, due to resorption of coatings.

Particulate reaction

The particulate-reaction scenario debonds the implant-bone interface for reasons other than load transfer. The most likely course of events is that wear particles—polyethylene from the acetabular bearing surface, acrylic cement abraded from the cement-bone interface or even metal debris from elsewhere—collect in the joint and migrate into the cement-bone interface from the periphery (Figure 3). These small particles activate macrophages at the interface into inflammatory responses of local bone resorption—or lysis, thereby gradually debonding cement and bone (Amstutz et al. 1992, DiCarlo and Bullough 1992, Schmalzried et al. 1992). Eventually this process produces relative interface motions and proliferates to gross loosening, in much the same way as the final stage of the accumulated-damage scenario described above.

For many years, particulate-reactions have been referred to as cement disease (Jones and Hungerford 1987). Since wear particles are generated in all hip

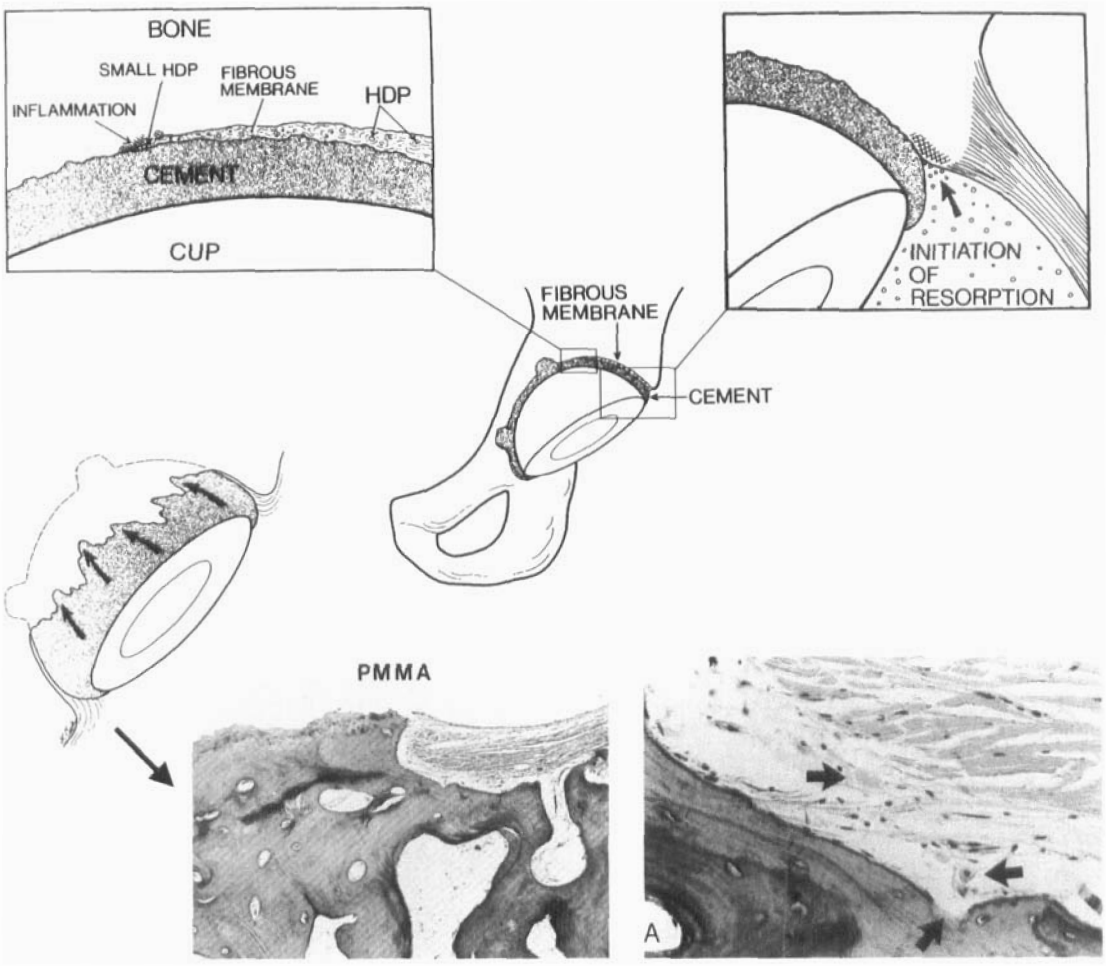


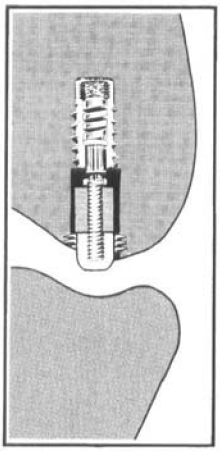
Figure 3. Schmalzried et al. (1992) found cement-bone interface loosening in acetabular cups due to particulate reactions in retrieved material. Mechanical damage was absent.

prostheses, and macrophage cells occur in interfaces of any kind, there is no reason to distinguish between the cemented and noncemented scenarios. That more failures have been reported for cemented THR is probably due to its longer history; failures of this kind are now also reported for noncemented reconstructions (Sumner and Galante 1992). Although noncemented THR does not suffer from abraded cement particles, polyethylene wear particles from the articular surfaces are as common as in cemented THR, while polyethylene or metal particles with other origins might even be more numerous, due to the common use of modular parts and wear-prone connections.

Failed bonding

Typical of noncemented THR is the failed-bonding scenario, which implies that ingrowth—or osseous integration—does not occur, due to gaps and relative motions at the implant-bone interface (Sandborn et al. 1988, Engh and Massin 1989, Søballe et al. 1992). The biological bonding or ingrowth processes require a certain degree of quiescence at the interface to succeed (Figure 4). It has been estimated that repetitive local displacements of more than 150 μm will prevent it (Pilliar et al. 1986, Sumner and Galante 1992). If bonding does not occur, a fibrous membrane is formed. Although this interface may be stable for a while, factors such as relative motions or wear particles are likely to provoke inflammatory reactions, as a result of which interface bone is resorbed (Spector et al. 1990).

A sub-scenario of this is characteristic of uncoated, press-fitted components, around which a fibrous membrane is usually formed (Brunski 1988, Sumner and Galante 1992). Hence, these prostheses are likely to follow this scenario while omitting its first act.



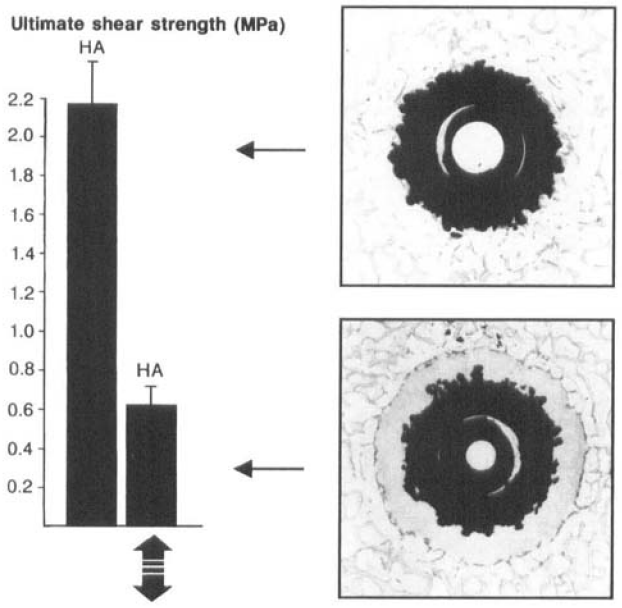


Figure 4. In an animal implant model, Søballe et al. (1992) found a fibrous interface around HA-coated plugs which moved in the bone, and low bonding strength. Stable plugs produced osseous integration and high bonding strength.

Stress shielding

The stress-shielding scenario particularly involves the cortical bone around the femoral stem. In any intramedullary reconstruction, the stem and the bone engage in a process of load-sharing; that is to say, the joint load, which was formerly carried by the bone alone, is now shared between the stem and the bone (Huiskes 1991). Basically, the share depends on their relative rigidities. If the stem is stiff relative to the bone, it takes a higher share of the total load; if it is flexible relative to the bone, it takes a lower share. Because the stem takes a part of the load away from the cortical bone, this is said to be stress-shielded (Huiskes 1991). As a result, the stresses in the cortex around the stem are subnormal (Figure 5). In accordance with Wolff's law, the bone then reacts with resorption in the form of osteopenia and reduction of cortical thickness (Huiskes et al. 1992, Sumner and Galante 1992). Hence, this scenario involves a natural process of strain-adaptive bone remodeling, as a reaction to an abnormal situation. The question whether this natural process must be seen as a failure is debated because, apart from alarming resorptive patterns on radiograms, clinical failures were hardly ever reported. Note, however, that the same was said many years ago about radiolucent lines at cemented interfaces (Harris 1992). Bone resorption reduces the support of the stem and may cause stem fracture (Engh et al. 1990) or breakage of the stem out of the bone, either by fatigue failure of the implant-bone bond or an impact force caused by a fall. Most alarming is the effect it may have on a revision procedure.

One of the problems for the clinical appreciation of this phenomenon is the imprecision of traditional radiograms when used to evaluate bone resorption (West et al. 1987); bone-mass reductions up to 30 percent may occur before they become visible. Recent applications of dual-energy X-ray absorptiometry have shown that the amounts of bone resorbed around nonce-

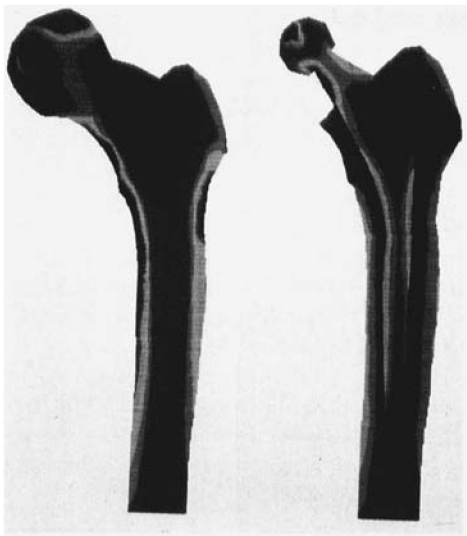


Figure 5. In the intact femur (left), most of the load is transferred through the cortices. After a femoral THR component is placed (right), a part of the load is transferred through the stem. The bone is thus stress-shielded by the stem (frontal mid-section through a 3-D FE model).

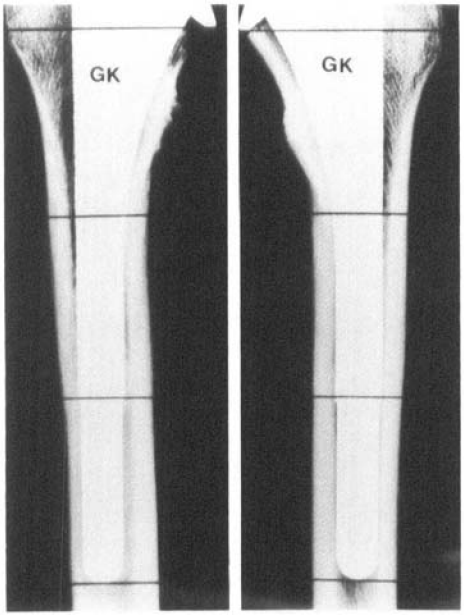


Figure 6. In a retrieval study of porous-coated femoral THR, in which the immediate postoperative (early) situation was reconstructed with the contralateral bone, Engh et al. (1992) found bone losses of up to 50 percent after 5 to 7 years postoperatively. Left retrieval, right reconstructed contralateral.

mented stems may be quite substantial, and that the resorptive process continues even several years postoperatively (Engh et al. 1992, Kiratli et al. 1992; Figure 6).

The degree of stress shielding in the bone around a femoral stem depends on mechanical factors only, such as stem stiffness, i.e., shape and elastic modulus, pre-operative bone stiffness, interface bonding characteristics, and hip-joint loading (Huiskes 1991). The only difference between cemented and noncemented reconstructions in this respect is that the latter tend to be much stiffer, hence produce more stress shielding. Using strain-adaptive bone-remodeling analysis, it can be shown that when comparing typical cemented stems with noncemented ones, of similar shapes, identical material, and both fully bonded and subject to the same loading history, the former would generate less than half the amount of bone loss than the latter (Weinans et al. 1992). Although the effects are much less alarming than for noncemented THR (Harris 1992), the differences are of degree rather than of kind, and the stress-shielding scenario constitutes a potential failure mechanism for both.

Stress bypass

Somewhat similar is the stress-bypass scenario. It develops in the bone around the femoral stem, when proximal load transfer is bypassed in favor of distal load transfer. As a result, the proximal femur is again understressed and becomes subject to strain-adaptive resorption, as in the previous scenario, but now from a different cause. This scenario is typical of uncoated, press-fitted stems in particular, and can develop due to inadequate proximal fit, either as a result of overreaming of the bone or undersizing of the stem, or due to gradual subsidence of the stem, secondary to proximal interface resorption. As a consequence, the stem jams in the distal canal (Draenert 1988, van Rietbergen et al. 1993).

Essential to this scenario is the fact that no bonding between stem and bone develops. Hence, it can be seen as an offspring of the failed-bonding scenario. Where for stress shielding the relative stem stiffness is the important design parameter, the emergence of a stress bypass is influenced predominantly by the surface-contour shape of the stem, for instance, the degree of tapering (Huiskes 1990).

Destructive wear

According to the destructive-wear scenario the bearing surfaces of the components gradually wear out, to the extent that mechanical integrity can no longer be maintained (Saikko et al. 1993).

Normally, the annual wear rate of articular surfaces in the hip is not seen as an important problem, because it is relatively minute (Amstutz et al. 1992)—of course the particles may cause problems, as discussed above. However, when patients tend to be operated on at a younger age, and fixation endurance improves, it is not unlikely that total hip joints simply wear out. In addition, wear is not limited to the articulating surfaces alone. As the modularity of prostheses is increased, more connections between separate parts are required, such as the conical fit of femoral heads or the connection between acetabular metal shell and polyethylene liner; not to speak of stems in two parts, spacers or other additions presently used to improve the fit. The connections are all subject to relative motions, however small, which create wear and abrasion, and corrode. It is not certain that they will be able to withstand, for instance, 20 years of heavy loading at one million cycles per year (Saikko 1993).

Destructive wear may occur in cemented and noncemented components alike. Problems emerging from the connection between liner and metal backing of cups, for example, have also been reported in cemented cases (Morscher 1992), and the cemented Christiansen prosthesis failed repeatedly due to this scenario (Herberts et al. 1989). Thus there is no principle differentiation in this scenario, although it will prevail in noncemented THR, simply because these tend to have coatings and more modular parts.

Table 1. A subjective attempt to score the prevalence of failure scenarios for different components and fixation modes

	Cemented	Noncemented	
Failure scenario	stem/cup	coated stem/cup	uncoated stem
Accumulated damage	+++ / ++	+ / +	–
Particulate reaction	++ / +++	++ / ++	+++
Failed ingrowth	– / –	++ / ++	+++
Stress shielding	+ / –	+++ / –	–
Stress bypass	– / –	++ / –	+++
Destructive wear	+ / ++	++ / +++	++

– not relevant; + possibly relevant; ++ probable failure mechanism; +++ predominant failure mechanism.

Biomechanics of THR failure

In summary, most of the failure scenarios are generic ones, potential failure mechanisms for cemented and noncemented reconstructions alike. Some prevail in cemented THR, some in noncemented THR; some are seen more often in the femur and some in the acetabulum (Table 1). Their prevalence is very much related to our contemporary experience, whereas the scenarios as such can also serve as references—or constraint conditions—for future developments. They can be seen as causative models, which are by nature conceptual and schematic. The real courses of events are more complex, and have more variability and interdependence than the scenarios suggest. The accumulated-damage and particulate-reaction scenarios, for example, are very difficult to separate in analyses of post-operative radiograms or retrieved tissues. There is no doubt that polyethylene wear particles are virtually always present in the fibrous interface tissues of a loose prosthesis. It is also well established that these particles can activate macrophage cells to resorb bone, and it is likely that such a process is enhanced by interface motions, which generate dynamic deformations and irritations of the tissues. However, it is also certain that relative interface motions create these resorptive reactions when the wear particles are not present (Brunski 1988, Spector et al. 1990), as would occur in the final stages of the accumulated-damage and the failed-ingrowth scenarios. Hence, it is virtually impossible to separate their final stages, and in reality there will be a strong interaction between both (Figure 7). Harris (1992) suggested that acetabular cups loosen mainly because of particulate reactions, in contrast to stems. This may well be the case, but as potential failure mechanisms, both scenarios are worth preventing in both components, even if their prevalence differs.

Damage accumulation is often called a biomechanical failure mechanism, and particulate reaction a biological one (Harris 1992). This has an unfortunate connotation—it sometimes seems to take the form of a contest—and it confuses the issues, because mechanics and biology play crucial roles in all scenarios. For example, stresses are purely mechanical variables, but the strengths of interfaces depend mostly on biological factors; stress shielding is a purely mechanical phenomenon, but its relationship with bone remodeling is a biological one; inflammatory reactions to particles are biological processes, but they are influenced by relative motions and wear processes, both mechanical phenomena. In fact, there is such an intimate relationship in THR failure mechanisms between mechanics and biology, that we might as well call their study biomechanics.

Incompatible design goals

It is likely that every THR reconstruction eventually fails, given enough post-operative years. The continuous dynamic hip- joint loads, up to one million cycles per year, are bound to accumulate fatigue damage in cement and interface bonds of any kind, are bound to wear out connections between modular parts, and are bound to produce a continuing supply of wear and abrasive particles. The question is not so much how to prevent the failure scenarios from being enacted as it is how to postpone them to maximize endurance. Obviously the answer is to be found in the optimization of prosthetic design, surgical factors and patient characteristics. The problem, however, is that we are

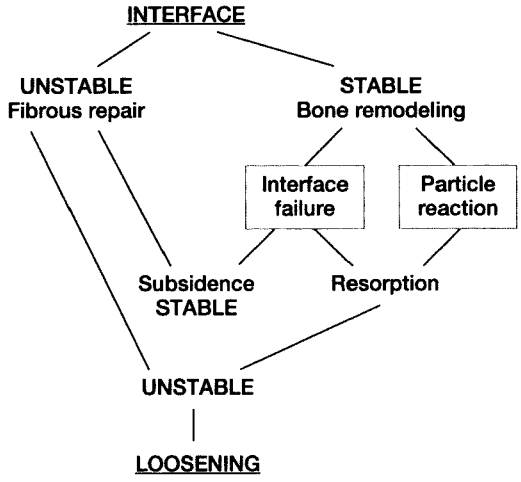


Figure 7. Factors which contribute to bone resorption and interface instability. Mechanical failure and inflammatory reactions act in concert to provoke instability (adapted from Spector et al. 1990).

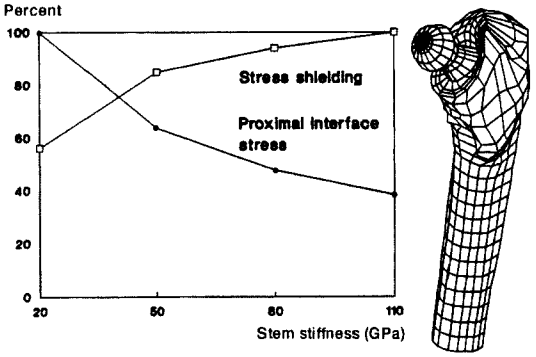


Figure 8. Conflicting design requirements for stem stiffness. A high stem modulus produces excessive stress-shielding, but moderate interface stresses, implying much bone loss, but a reduced likelihood of interface failure; these odds are reversed for the flexible material. A modulus of 110 GPa represents titanium; 20 GPa is close to the modulus of cortical bone (result of a 3-D FE analysis; from Huiskes et al. 1992).

often faced with incompatible design requirements. What may be good for postponing one failure scenario, may enhance another. An obvious example of that is modularity in noncemented component designs. In order to reduce initial relative motions—hence increase initial stability—the fit must be maximized. Because of the variability in anatomy and the imprecision of the reaming procedures, it is advantageous in this respect to use modular designs which allow for peroperative assembly. However, the connections between the modular parts are subject to wear and corrosion. Hence, the more connections, the greater the likelihood of destruction and the more particles will be produced. Thus one problem is addressed, but another is introduced. There are many examples of this in contemporary THR design, some quite obvious, some more subtle. I will identify a few on the following pages.

Stem stiffness versus flexibility

If a stem is stiff, it transfers more load distally and less proximally, and it takes a larger share of load away from the proximal femur. If it is flexible, we see the opposite. As a result, flexible stems provoke less stress shielding in the bone, but more interface stresses—cement stresses, if cement is present—on the proximal side. Thus excessive bone resorption is more likely to occur around stiff stems, but proximal interface (and cement) failure is more likely to occur around flexible stems (Figure 8). An additional problem of flexible stems is the enhancement of relative interface motions when stems are unbonded (Jakim et al. 1988, Nistor et al. 1991). In short, flexible stems reduce the probability of stress shielding, but they promote accumulated damage and failed ingrowth; a real design conflict, because a stem cannot be flexible and stiff at the same time.

Although adaptive bone resorption is presently not seen as an important problem for cemented stems, a number of flexible cemented prostheses have been marketed in the past, featuring reduced thickness and made out of titanium rather than cobalt-chromium alloy, the former material having about half the elastic modulus of the latter. And, indeed, one problem was addressed but another emerged: these flexible cemented designs have generally shown higher loosening rates than the conventional, stiffer ones, most probably due to accumulated damage (Robinson et al. 1989, Ebramzadeh et al. 1992). Presently, a number of companies are experimenting with flexible, composite noncemented stems, hoping to minimize stress shielding (Sumner and Galante 1992). Certainly, stress shielding will be reduced, but there is no indication yet that the coating strengths of these stems will be higher and the initial stability better, to compensate for the higher proximal interface stresses and motions which are bound to occur.

In fact, the wrong questions are posed, because the incompatibility of design goals is neglected. The problem is not just how to minimize stress shielding, but rather how to minimize stress shielding while keeping interface stresses and motions at an acceptable level; or how to minimize interface stresses and motions while keeping bone resorption at an acceptable level.

Bonded versus unbonded stems

The strength of the stem-cement bond in cemented THR is relatively low. Hence, the failure of this connection in the course of time is not unlikely. Once the stem-cement interface is debonded, the stresses in the cement can increase dramatically, enhancing the probability of damage accumulation, cement failure (Figure 9), cement-bone interface loosening and, eventually, gross loosening (Jasty et al. 1991, 1992, Huiskes 1993). This damage accumulation can be prevented by two measures: increasing the strength of this bond and reducing the stress to which it is subjected. The first measure requires a rough grid-blasted surface-finish or PMMA pre-coating (Harris 1992). The second measure requires a profiled stem shape which minimizes cement stress (Huiskes 1991). Thus, stem design can be optimized to reduce damage-accumulation.

However, what if these two measures are not sufficient to prevent stem-cement debonding? In that case, it must be accepted that this connection is loose as a rule. The metal will rub against the cement mantle, producing metal and cement particles and enhancing the probability of failure due to particulate reactions

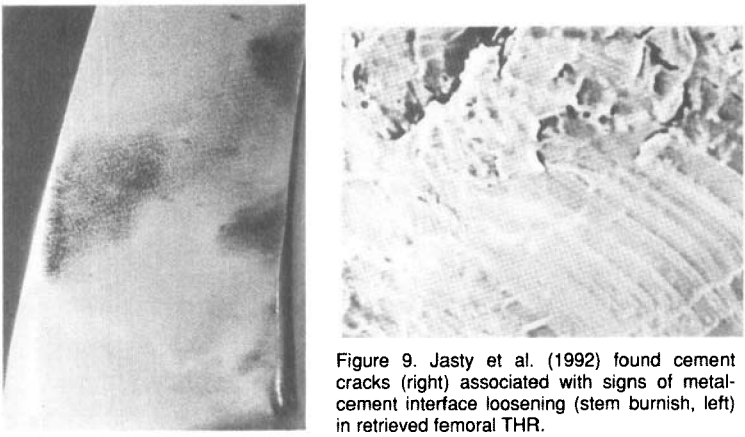


Figure 9. Jasty et al. (1992) found cement cracks (right) associated with signs of metal-cement interface loosening (stem burnish, left) in retrieved femoral THR.

(Ling 1992). That scenario can be countered best by using a smooth stem—which reduces metal abrasion—of a straight-tapered shape, giving the stem the opportunity to sink and find new stability. But clearly a stem cannot, at the same time, be smooth and straight-tapered to prevent particulate reactions, and rough and profiled to prevent accumulated damage. So here, too, we see a design conflict. Note that the heart of this matter is the question whether it is at all possible to keep the stem-cement bond from loosening, a question without a clear answer.

Where noncemented stems are concerned, another conflict occurs. An unbonded, press-fitted stem compresses the endosteal bone envelope due to its elastic subsidence, necessary to balance the external forces (Huiskes 1990). As a result, the stresses in the cortical bone are increased and the effects of stress shielding are diminished. And indeed, smooth, tapered press-fitted stems were marketed, promoting this feature. On the other hand, however, due to the difficulties of obtaining a reproducible fit, many of these stems are likely to subside and jam distally, promoting stress bypass. Lack of bonding also exposes the reconstruction to a failure of ingrowth. So this is another example of a solution with adverse side-effects; a design objective for one failure scenario, which neglects the constraints presented by the others.

Fully versus partly coated stems

If a stem is coated to allow for bone ingrowth, should the coating be placed over its entire surface in order to maximize the probability of bonding? Or should it be limited to a proximal part only, to promote proximal load transfer and reduce bone resorption? Again we have a conflict between design requirements for different objectives. Full coating provides better resistance against failed ingrowth, but promotes stress shielding. The important question is not whether the stem should be fully or partly coated, but how much should the coated area be reduced in order to provide adequate resistance against failed ingrowth, while substantially reducing stress shielding. This key question has not been addressed explicitly by prosthetic companies, and is more complicated than first meets the eye.

A small proximal bonding area is subject to high interface (shear) stresses and provokes bonding failure. Secondly, the uncoated part of the stem does not bond and moves against the endosteal surface, producing abrasive particles, interface-motion-related bone resorption and thigh pain. Sumner and Galante (1992), experimenting with various canine hip replacements, have found consistently that a fibrous interface develops where a stem is uncoated. This has two potential effects. First, the fibrous membrane may, in time, become unstable under the influence of relative motions or accumulated wear particles (Spector et al. 1990), enhancing progressive endosteal resorption. Secondly, the layer around the distal stem plays a smaller role in the load-transfer mechanism, thus becoming superfluous. Hence, although the distal stem may be required initially to enhance stability, once the proximal coating has grown in, it develops into an undesirable element. Thus the design conflict involves not just the coating area, but also the length—or even the presence —of the distal stem. A stem cannot be short, long, fully and partly coated, all at the same time. The search should be for the optimal compromise between incompatible design goals, not for lopsided inventions.

The innovation process

Many innovative devices and techniques introduced in recent years have failed, and a substantial number of patients might have been better off with a conventional

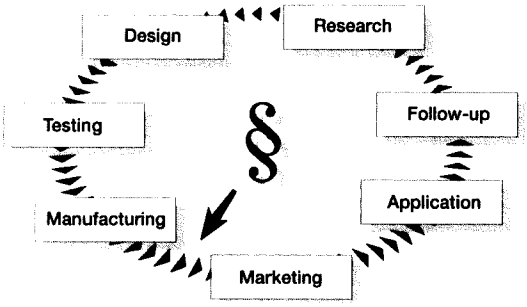


Figure 10. The prosthetic (innovation) life cycle; the paragraph sign represents market introduction (from Faro and Huiskes 1992).

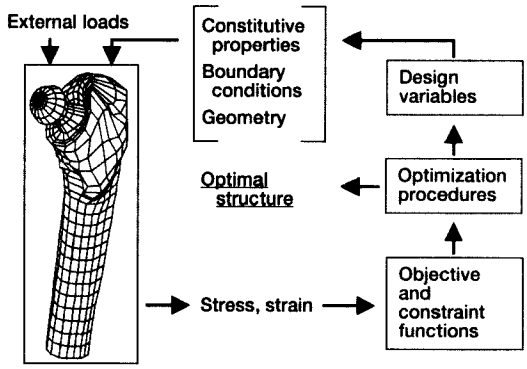


Figure 11. The method of structural optimization with FE models can be useful to establish guidelines for compromises between conflicting design goals, taking mechanical objectives and constraints into account (courtesy of J H Kniper, Univ. of Nijmegen, Netherlands).

Charnley—or similar—procedure. In that sense, the innovation process has "failed to deliver". The question is then, is the traditional method simply the best one conceivable, or is something amiss with the innovation process itself? The latter, I believe, is the case. Figure 10 depicts the life cycle of THR (Faro and Huiskes 1992), which represents the innovation process as well. Ideally, research—clinical and experimental—produces information based on which new, improved designs can be developed. After these new prostheses have been pre-clinically tested, they are manufactured. The process may then go repeatedly through the stages of clinical trials, research and design adaptations, until the new device is considered safe and effective. It is introduced in the market and tried in patients. The proper follow-up of patients produces indications about failure mechanisms and potential improvements. This new information leads to new research, new designs and the cycle starts again. The arrow in the cycle signifies possible barriers to market introduction, such as government regulation, e.g., FDA approval (Faro and Huiskes 1992). I see three principal problems associated with the way in which the cycle runs today. The first one concerns the design process, the second the regulation of market approval and the third the repetition frequency of the cycle as a whole.

Systematic design process

The primary requirement for a new design is that it has an objective—an explicit statement of purpose— indicating which problem it is trying to solve. In a systematic design process, objectives and constraint conditions must be addressed in a systematic way to reach the best compromises relative to conflicting requirements (Figure 11). This process should be based on available clinical and experimental research findings. This, however, is not how it is generally conducted. As we have seen in the previous section, design solutions tend to be unidirectional; incompatible design goals are often neglected. The reason is, possibly, that the design process is governed by different decision-makers. Innovations are usually realized in collaboration between manufacturers and orthopedic surgeons. The former, however, are not responsible for the THR, but merely supply the parts, whereas the latter generally lack the knowledge and experience to handle a systematic design process, and the time to keep up with the scientific literature. They are, in this respect, amateurs.

Market regulations

Unsafe designs are harmless as long as they are not implanted in patients. However, there are no satisfactory *regulatory mechanisms* to prevent them from being sold freely on the market (Faro and Huiskes 1992). Although the U.S.A. has its FDA approval requirements, and a number of other countries have government regulations to monitor market introduction, these hardly consider the long-term endurance of a new THR concept. In our recent publication (Faro and Huiskes 1992), we have shown that disastrous THR products have been marketed in the past, government regulations notwithstanding.

Trial-and-error

The third problem is the trial-and-error culture in orthopedics. It is general practice to convert an idea into a device, and then test it out in patients to see if it works. Although this approach brought success in Charnley's days, one tends to forget that no dependable cure for the severely disabled was available then—and there is one now, whatever its long-term limitations. The trial-and-error culture has several unfortunate consequences. It promotes the clinical

application of devices of which the design presumptions have not been confirmed. An illustrative example is surface replacement, which was introduced with the presumption that this method of reconstruction produces bone stresses closer to normal than could be realized with intramedullary fixation. A simple finite-element analysis would have shown that this is not true. Such a design-evaluation analysis does not produce evidence for the safety of the device in question, but it does clarify whether the theoretical goals of the innovation are attainable. The second consequence is premature market introduction. It is not uncommon to introduce an innovative prosthesis in the general market only a few years after the first clinical trials have been started. As discussed previously, it takes many years of extensive clinical testing before one can decide whether the new design performs significantly better than a conventional one (Herberts et al. 1989). The surgeons who use these new prostheses are not always aware of their participation in what is basically a clinical experiment, nor are their patients. The third consequence is a common lack of systematic scientific analysis of clinical performance and failures. Before the results of the device have been properly analyzed, a new idea has been introduced. The orthopedic community at large thereby misses the opportunity to learn and enlarge its scientific body of knowledge, on which real progress can be based. The trial-and-error culture makes the innovation cycle spin like a merry-go-round; it produces mostly wind.

Deficient innovation process

In summary, shortcomings in the innovation process can be explained by the failure to note incompatibilities in design goals for prosthetic devices, the trial-and-error culture in the quest for progress in orthopedics, and the lack of adequate regulatory mechanisms for market introduction.

Government regulation of market approval cannot be expected to solve the innovation impasse effectively and efficiently (Faro and Huiskes 1992). Self-regulation of the innovation cycle by the orthopedic community as a whole, however, and individual self-control, could change the present culture. One could imagine a set of self-imposed standards to regulate the innovation process effectively. These rules should primarily address three issues. First, to prevent unsafe devices being tried in patients, which is a matter of pre-clinical testing. Secondly, to prevent unsafe devices being sold freely on the market, which is a matter of clinical trials. Thirdly, to prevent unsafe, marketed devices from causing a disaster, which is a matter of post-operative screening and follow-up. As we will see in the next section, recent research in orthopedics and its related sciences has provided excellent opportunities not only to enhance design processes of THR devices, but also to provide effective methods for clinical and pre-clinical testing on which self-imposed rules could be based.

Table 2. An attempt to assess the effectiveness of preclinical and clinical limited-trial-duration test methods in estimating the resistance of THR components against the failure scenarios

Failure scenario	A	B	C	D	E	F
Geometric analysis	–	–	+	–	±	–
FE stress analysis	+	–	±	±	±	±
Remodeling simulation	–	–	–	+	+	–
Micromotion tests	–	–	+	–	±	–
Lab. bench tests	+	±	–	–	–	+
Dyn. hip simulator	–	+	–	–	–	+
In-vivo RSA	±	±	+	–	+	–
Gait analysis	±	±	+	–	±	–
Digitized X-rays	±	–	±	±	±	±
Dual-energy absorpt.	–	–	–	+	+	–

A Accumulated damage
B Particulate reaction
C Failed ingrowth
D Stress shielding
E Stress bypass
F Destructive wear
–not useful; ± gives indication; + useful test.

Pre-clinical testing

Pre-clinical tests of prosthetic devices are required, first of all, for the purpose of design evaluation. Are the presumptions of the design objectives valid and does the actual design fulfil these objectives? This question can only be answered if these objectives and presumptions are explicit and precise. The second purpose is to screen them for adverse safety and endurance effects. Two aspects are important, objectives and methodology. To perform a test of a device concerning safety and endurance, one must know what to test for. Hence, the appreciation of failure scenarios is essential. Secondly, the test methodology must be valid for its purpose. In the following pages a number of pre-clinical test methods are reviewed, regarding their validity and the failure scenarios they can address (Table 2).

Geometric analysis

The shape and dimensions of the prosthetic components largely determine the load-transfer mechanism and stress patterns in THR (Huiskes 1991). In cemented THR, shape determines the thickness profile of the cement mantle. In noncemented THR, the fit depends on it. Shape and dimensions are complex entities in designing THR components, because of natural

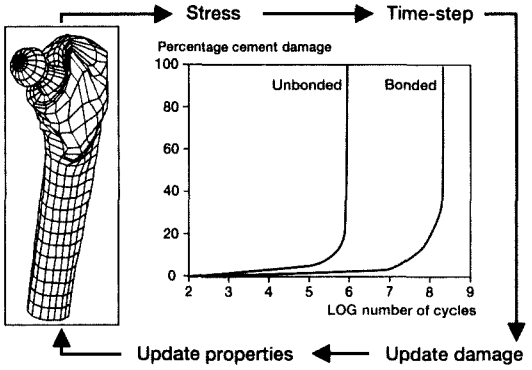


Figure 12. FEA can be applied in the design stage of THR components to document the load-transfer mechanism and stress patterns on a relative basis. This example shows that metal-backed cups (right) tend to emphasize peripheral load transfer and central stress shielding relative to non-metal backed ones (left). High-stressed areas are black (courtesy of M Dalstra, Univ. of Nijmegen, Netherlands).

Figure 13. A FE computer-simulation study of damage accumulation in cement. The iterative simulation model uses experimental data on cement fatigue-failure characteristics. The graph indicates that the process occurs much faster in the cement around a debonded stem than in a bonded one. Analyses of this kind are useful in pre-clinical tests of new designs before they are tried in patients (courtesy of M Verdonschot, Univ. of Nijmegen, Netherlands).

variations in bone geometry in the patients (Noble et al. 1988). Optimal fit is usually a design objective for prosthetic components, and many manufacturers advertise advantages for their products in this respect, such as anatomic shape, optimal fit, proximal fit, canal filling and the like. An excellent and relatively cheap method to check whether these goals are realized in the actual design is geometric analysis. The components, or models thereof, are placed in a series of bones, and the gaps between implant and bone—or the thickness profile of the cement mantle for cemented THR—are evaluated (Huiskes 1991). Although it is likely that these kinds of experiments are often performed in the design stage, their results hardly ever surface. Usually, the above-mentioned claims are made without supporting documentation.

Apart from being an excellent method for design evaluation, geometric analysis is also a versatile tool for pre-clinically testing the likelihood of the failed-ingrowth scenario to be enacted. It can also address the stress-bypass scenario (Table 2).

Finite-element stress analysis (FEA)

This is a design-evaluation and pre-clinical test method which can be applied in an early stage, before prototypes of the THR components are actually made. Based on information about shapes and dimensions of bones and implants, elastic properties of the materials, bonding conditions of the implants and external loads, an FEA computer model can be developed which predicts the stresses and strains in the loaded THR (Huiskes 1991). In this way, the load-transfer mechanism can be documented and compared to other—traditional—designs (Figure 12). The mechanical design presumptions and objectives of a new device can also be checked (compare the discussion about surface replacement above). In addition, stress values can be compared to strength data on implant materials, bone and interfaces, in order to estimate the probability of long-term failure.

FEA of joint-replacement structures took a long time to develop. In the early years, little was known about joint loads, bone properties and damage accumulation processes in the materials concerned. The features of FEA methods themselves and the capacity of computers were not so good as those required for valid FE models. The development in FEA of biological structures during the last decade, however, has been staggering (Huiskes and Hollister 1993). Great progress was made in the analysis of hip-joint loads (Bergmann et al. 1989) and bone properties (Hayes 1991). Three-dimensional FE models can now be used to predict stresses, strains and even interface motions in THR structures, with quite reasonable accuracy. In this way the probability of long-term failure due to accumulated damage can be estimated preclinically. It has been shown that there is a correlation between the predictions of FEA and the relative clinical endurance of different cemented THR designs (Huiskes 1993). Because the damage-accumulation process itself can now also be simulated (Figure 13), the actual endurance limits of reconstructions with new devices can be estimated, at least on a relative basis, in comparison with traditional reconstruction methods (Huiskes and Hollister 1993). FEA also produces information about stress-shielding of the bone, interface motions and stresses at modular prosthetic connections. Hence, the stress-shielding, stress-bypass, failed-ingrowth and

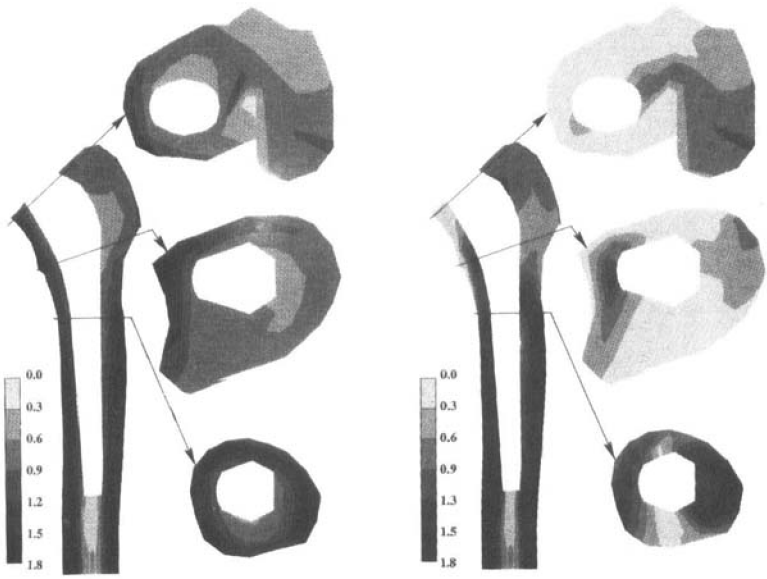


Figure 14. Result of a bone-remodeling computer-simulation study, comparing the long-term (homeostatic) bone density around a noncemented femoral stem (right) to the initial density (left). About 50 percent overall bone loss is predicted with this particular design. A versatile tool for pre-clinical testing (from Huiskes et al. 1992).

destructive-wear scenarios can be addressed as well.

A FEA tests the concept of a design only. It is unable to predict, for example, the stress values in all patients who will receive the new prosthesis in the future. The computer model is an average or typical representation of a THR. For that reason, pre-clinical testing with this method is best applied on a relative basis in standardized bone models.

FE bone-remodeling analysis

Although FEA gives an indication of the amount of stress shielding provoked by a THR design, it does not predict the amount of bone resorption this will eventually cause. In recent years, however, methods of FE bone-remodeling analysis have been developed, which are suitable for simulating the bone-remodeling process in computer models (Huiskes and Hollister 1993). These models are based on strain-adaptive bone-remodeling theory—quantitative formulations of Wolff's law—and they effectively estimate the long-term bone loss around hip stems with regard to shape, material and fit of a particular stem design (Huiskes et al. 1992; Figure 14). Hence, it is a versatile tool for design evaluation and pre-clinical testing of new devices relative to the stress-shielding and bypass scenarios.

Laboratory bench test

Nothing seems more obvious than pre-clinical tests of a THR configuration for strength and endurance in a laboratory bench test. However, no models for this purpose have been developed, with the exception of the ISO standard test for fatigue strength of isolated hip stems. The lack of scientific efforts in this direction may have been enhanced again by the divided responsibilities in THR innovation processes, where manufacturers feel responsible for the prosthesis, but not for the THR, while surgeons acknowledge their responsibility for the THR, but know little about mechanical strength and endurance. Another problem, however, is the difficulty of finding a suitable standardized laboratory bone substitute. In addition, the assessment of damage inside such models is not a trivial matter. Laboratory models could be an alternative—or complement—for FEA stress analysis in testing for the accumulated-damage scenario. Although much more expensive than FEA, they could serve to remove some of the inherent uncertainties of computer models, once a prototype of the prosthesis is available.

Endurance tests of this kind can also be used to test the connections in modular prosthetic components for destructive wear. In view of the many problems with these connections reported recently, pre-clinical testing of this aspect seems rather urgent.

Dynamic hip simulators

Special cases of laboratory bench tests are those with dynamic hip simulators, a number of which have been

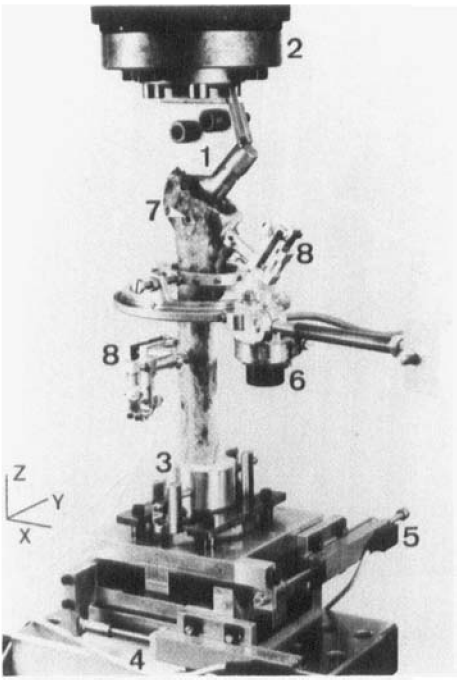


Figure 15. Micromotion bench tests are important to establish the initial stability of a new (noncemented) THR component pre-clinically (from Schneider et al. 1989).

reported in the literature (McKellop and Clarke 1984). In these machines, the motions as well as the forces of the hip joint are simulated in order to measure articular wear (Saikko et al. 1993). They can also be used to evaluate wear at modular connections (Saikko 1993). It is probably the only pre-clinical test method presently available for the particulate-reaction scenario.

Micro-motion bench test

A method that has been developed satisfactorily in recent years, particularly for noncemented THR, is the micro-motion bench test. The purpose of this test is to determine the initial stability of THR devices by measuring the relative motions between implant and bone. Because these motions are small and the three-dimensional motion patterns are complex, refined measuring tools are required. Usually, multiple displacement gauges are positioned between implant and bone specimens to measure the motions (Schneider et al. 1989, Nistor et al. 1991; Figure 15). An alternative method is roentgen stereophotogrammetric analysis (RSA), which uses tantalum pellets connected to implant and bone as markers, and three-dimensional reconstruction of motions from stereo-radiograms (Selvik 1989, Huiskes 1991). These micro-motion tests are useful in particular to assess the likelihood of failed ingrowth.

Animal experiments

Finally, animal experimental models can be used to pre-clinically test devices for all failure scenarios. However, animal THR differs in many respects from its human counterpart—such as shape, dimensions and properties of bones, as well as loads—so the actual devices to be tested cannot be used. Nevertheless, particular innovative aspects of prosthetic components can be tested in this way. Examples are new materials or fixation methods. In recent years, there has been much progress in the development of standardized models for this purpose (Sumner and Galante 1992, Søballe et al. 1992).

Common limitations of test models

The above methods use computer, laboratory or animal models. In this sequence they offer an increasing degree of closeness to reality, but a decreasing degree of experimental control. They also cost more. Yet, they are all models, reductions of reality, and should be used preferably on a relative basis in a standardized fashion. Pre-clinical tests must be performed with regard to particular, well-defined failure modes. Consequently, the tests only provide checks for particular problems. In other words, their results cannot be used to decide whether a design is good enough, but only to determine if it may be deficient in a particular way. It is like a new airplane: everything possible will be done to pre-check its safety in the design and laboratory stages. The next step is to see whether it, indeed, will fly.

Clinical trials and post-operative screening

Clinical trials have not been very effective in the past, because of the trial-and-error culture, which promotes retrospective studies instead of prospective ones, because of ethical objections against double-blind studies, because of imprecise radiographic and subjective patient evaluation methods and because most problems reveal themselves in the long term only. There is, however, reason to be optimistic. If rigorous pre-clinical testing is performed in advance, the ethical objections against double-blind studies diminish, while advanced clinical measurement methods, developed in the recent past, allow for earlier detection of problems.

Roentgen stereophotogrammetry

An example is in vivo roentgen stereophotogrammetric analysis (RSA), as developed by Selvik (1989). By inserting small tantalum pellets in the bone around the prosthesis as radiodense markers, and attaching others to the implants, three-dimensional migration and

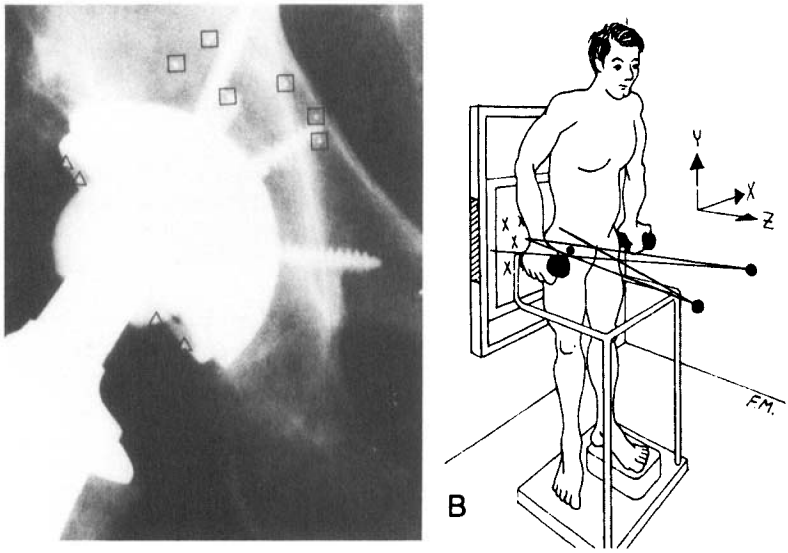


Figure 16. Roentgen stereophotogrammetric analysis (RSA) uses tantalum pellets and stereo-radiogram measurements to determine micro-motions and migrations of THR components in vivo. Although too tedious to be used routinely, it is an excellent method to clinically evaluate new designs in clinical trials of short duration (from Kärrholm and Snorrason 1992 (left) and Ryd 1992 (right)).

micro-motions can be measured very precisely by computer analysis of stereo-radiograms (Kärrholm and Snorrason 1992; Figure 16). In this way, insufficient fixation of the components can be assessed after a very short period of time.

Digital radiographic analysis

A similar function is provided by digital radiographic analysis (Dooley et al. 1992). Digitized radiographs allow for standardized longitudinal computer evaluation of prosthetic migration, in which variations in exposure characteristics can be compensated for. Although much less precise than RSA, it is easier to use routinely. It can also be applied for various other parameters, such as implant position and acrylic cement thickness.

Dual-energy X-ray absorptiometry (DEXA)

Another new radiographic tool is dual-energy X-ray absorptiometry (DEXA), to precisely measure bone density, which is virtually impossible with traditional radiographs. Post-operative bone remodeling can be monitored in this way (Engh et al. 1992, Kiratli et al. 1992).

Gait analysis

Finally, gait analysis has been developed up to a point where it can be applied routinely and provide objective information about function and indications for loosening, where as earlier only subjective indicators could be used (Andriacchi and Mikosz 1991). A loose implant provokes evidence of instability in the hip-joint force patterns during gait, even before the patient has developed clear signs of pain.

Multi-center trials

These clinical measurement tools can be applied to test THR devices for various failure scenarios (Table 2) in a reasonable period of time. After rigorous pre-clinical testing and clinical trials, there is still no guarantee that an innovative device will endure longer than a traditional one. But at least everything feasible will have been done to eliminate errors. Certainty can only be obtained in long-term follow-up studies, coordinated in multi-center trials. An excellent infra-structure for these has been set up in Sweden, where all THR patients can be followed on a national basis (Herberts et al. 1989, Malchau et al. 1993). Particular prosthetic types can be evaluated for their survival rates (Figure 17). Because of the large numbers and the participation of all surgeons, unbiased data can be collected. This is an excellent method for monitoring unsafe devices. Indeed, all prostheses on the market should be included in coordinated multi-center trials like this, until enough data are collected to establish their quality.

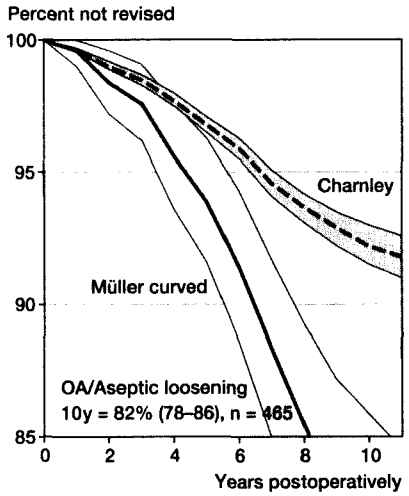


Figure 17. Multi-center clinical trials are well suited to monitor the endurance of new THR designs on the market, relative to traditional ones, in order to limit unnecessary failures. The Swedish trials are conducted in a closed health-care system, hence they limit biases (courtesy of Henry Malchau, MD, Gothenburg).

The individual surgeon and the innovation process

The stages of the innovation cycle discussed here concern primarily those involved in prosthetic development—manufacturers, inventive surgeons and scientists—and, of course, the orthopedic community as a whole. In the selection of prosthetic types, application and post-operative care, the individual surgeon plays the key role. This responsibility requires understanding of the innovation process and its principal mechanisms. For a company, a good prosthesis is one that sells well. A marketing department tends to emphasize the innovative aspects of a new device; a sales person will use all arguments likely to get the prosthesis sold; such are their professional roles. It is up to the customer, the surgeon, to ask the critical questions and request documentation of design-evaluation analyses, pre-clinical tests and clinical trials. When a prosthesis is said to be anatomic, are there reports available in which the fit is documented? If a design is claimed to produce less stress shielding and bone resorption, is this merely an expectation, or is it established in one way or another? And, if so, what about its vulnerability to other failure scenarios? Are the articular dimensions, surface finishes and tolerances such that wear is minimized? If so, can that be documented? It takes knowledge and continuing education to ask the right questions and assess their answers.

It is the orthopedic surgeon who produces a THR; companies merely supply the parts. Success or failure is determined by prosthetic design, surgical and patient factors. Probably, surgical factors are the most important of these. This also implies that the surgeon selects a design which fits his skills and experience, and that he is well-informed about its scientific background, its design concept. It is important to apply the pre- planning procedures designed for the prosthesis used. These are meant to set the technical goals for the actual procedure and to provide the references for the first post-operative radiogram. Thus, they promote technical precision and self-assessment. Failure to use them is like playing golf in the dark.

THR is a clinical service, not a device. Patient selection, preparation and follow-up are as essential as optimized designs and surgical skills. Frequent follow-up is important for many reasons, of which one deserves to be emphasized. It is essential for the eventual success of the THR as a service for each individual patient. In the introduction, I raised the question whether it would be possible at all to create a THR which lasts a lifetime. Frequent monitoring of the patient—with proper appreciation and assessment of signs produced by failure mechanisms—would provide the opportunities for a timely revision, before the loosening process has destroyed the bone (Müller, 1992).

When Coventry was asked to discuss the lessons learned in 30 years of THR, he chose to emphasize awareness and responsibility rather than techniques and design features: "Know and understand the evolution of hip arthroplasty. Selection of the prosthesis must be carefully made from sound clinical and scientific data. Before embarking on a prosthesis program, establish ... (a) prospective protocol for the operation ... and for... (a) long-term follow-up.... Be familiar with the basic biomechanical principles of load and stress about the hip. Learn to use polymethylmethacrylate properly. ... Be wary of the complications .... (relating to) technique, as well as design and even patient selection..... A surgeon should always be able to review his or her experience with unbiased, objective scrutiny" (Coventry 1992). Such lessons are even more pertinent today, when therapeutic and commercial interests do not always coincide.

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