From structure to process, from organ to cell : recent developments of FE-analysis in orthopaedic biomechanics

From Structure to Process, From Organ to Cell: Recent Developments of FE-Analysis in Orthopaedic Biomechanics

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The introduction of finite element analysis (FEA) into orthopaedic biomechanics allowed continuum structural analysis of bone and bone-implant composites of complicated shapes (Huiskes and Chao, J. Biomechanics, Vol. 16, 1983, pp. 385–409). However, besides having complicated shapes, musculoskeletal tissues are hierarchical composites with multiple structural levels that adapt to their mechanical environment. Mechanical adaptation influences the success of many orthopaedic treatments, especiallly total joint replacements. Recent advances in FEA applications have begun to address questions concerning the optimality of bone structure, the processes of bone remodeling, the mechanics of soft hydrated tissues, and the mechanics of tissues down to the microstructural and cell levels. Advances in each of these areas, which have brought FEA from a continuum stress analysis tool to a tool which plays an ever-increasing role in the scientific understanding of tissue structure, adaptation, and the optimal design of orthopaedic implants, are reviewed.

Introduction

In view of the orthopaedic goals, it is not surprising that biomechanics has developed a prominent role as a basic science in orthopaedic teaching and research, since it provides the theories and methods for functional mechanical analyses (Mow and Hayes, 1991). Finite Element Analysis (FEA) was first introduced in the orthopaedic literature 20-years ago (Brekelmans et al., 1972), and developed as a tool to analyze load transfer in bones and artificial-joint replacements (Huiskes and Chao, 1983).

The application of FEA in this field has never been trivial. Apart from the variable, dynamic loads to which bones and joints are subjected, the constitutive properties of their tissues are also complex. The structures to be analyzed are always composites at various structural levels. Combinations of bones and biomaterials often occur, with the bone itself a noncontinuous composite from macro to submicroscopic levels. Collagenous tissues like articular cartilage, intervertebral discs and ligaments are even more complex in their mechanical properties, as a result of interactions between solid and fluid phases. Test specimens are not readily available and the determination of constitutive structure-function relationships has become a science by itself.

Many time-dependent processes play a role as well. These can be of mechanical origin, for instance fluid consolidation, measured in seconds, but also of chemical origin, like osmotic processes, measured in hours, and even of biological origin, due to adaptive remodeling, measured in years. Because these processes are substantially influenced by mechanical variables, complex feed-back interactions occur within tissues, which effect their mechanical behavior as well as their normal physiology, failure mechanisms, and pathology.

The traditional role of the FE-analyst in engineering mechanics has been to assess structural failure probabilities, given the applied loads, and the elastic and failure properties of the structural materials. This was also basically the scope of FEA during the first ten years of its application in orthopaedic biomechanics (Huiskes and Chao, 1983). During the last decade, however, the scope has widened, both in a technological and philosophical sense. Increased computer capacities and more sophisticated (nonlinear and 3-D) FE program packages have allowed for more realistic modeling and the application of iterative procedures to describe time-dependent mechanical behavior and perform structural optimization. These new capacities have extended the scope on one side to the study of constitutive tissue behavior, and on the other side to the study of mechanical failure mechanisms, biological processes and pathology. The FE-method has developed in biomechanics from a tool of technology to a tool of science.

The purpose of this article is to discuss these recent developments. Rather than a historical review or an explicit discussion of goals and methods, we have opted for an implicit approach, discussing contemporary work in three advanced

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areas of FEA technology, emphasizing applications of structural optimization, process analysis, and micro-modeling to bones and implants.

Structural Optimization

By using structural optimization with the FE-method, its role is in effect reversed. FEA produces field variables (i.e., stresses, strains) from given external loads, geometry and constitutive properties. Conversely, a numerical optimization process (usually) produces a geometry or a set of constitutive parameters to satisfy a given, desired field-variable distribution. Central in its use is the definition of an optimization target, described by an *objective function* which is minimized in an iterative process. Secondary goals may be included as *constraint functions*. During the iterative process, a finite number of *design parameters* are varied within the design space, of which the boundaries are represented by constraint conditions. Structural optimization is beginning to attract attention in two areas of bone and implant research.

Optimization of Implant Design. The design problem for orthopaedic implants is to determine the combination of design parameters which maximizes the longevity of the artificial reconstruction, given the surgical and biological constraints of applicability and biocompatibility. Such a problem is well suited for a numerical optimization approach.

Huiskes and Boeklagen (1988, 1989) and Yoon et al. (1989) applied numerical shape optimization in combination with simplified one and two-dimensional FE-models of the femoral-stem reconstruction in cemented hip replacement. The objectives were to find an optimal stem shape to minimize the chances for loosening. For this purpose objective functions were formulated to minimize stress peaks in cement and at the cement-bone and cement-stem interfaces, while varying design parameters representing the stem contour in the frontal plane. The maximal available space in the bone and the maximal allowable stress in the stem were used as constraint conditions. Shapes were found which indeed reduced stresses dramatically—at least in theory. The nature of these shapes was such that they would probably not have been found in traditional parametric analyses, thus illustrating the attractiveness of the method.

FE optimization studies for noncemented hip stems were reported as well. Blake et al. (1992) used a quasi-3-D laminated-plate model to minimize stresses at the stem-bone interface, by variation of stem shape and material properties. Kuiper and Huiskes (1992) applied a 2-D (side-plated) FEM model to find an optimal elastic modulus distribution for a nonhomogeneous, composite stem. Minimal chances for interface loosening was the optimization target, constrained by a maximally allowed reduction of normal (natural) bone stresses, in order to prevent excessive adaptive bone resorption (Fig. 1). In this case the use of optimization is particularly attractive, because these two criteria contradict, in that prevention of loosening requires a rigid stem, and the maintenance of bone a flexible one (Huiskes et al., 1992). The result of the optimization process was a stem with a high modulus proximally, gradually reducing towards distal.

The usefulness of FE-based optimization studies of implant design has certainly been demonstrated, but only a few studies have been published as yet. Most of these used simplified FE-models to accommodate limitations in computer capacity. However, efficient software for analytical determinations of search directions, using the adjoint variable method, is now available (Haug et al. 1986), and computer capacity has also improved greatly in recent years. The application of full, "Anatomic" FE-models is no longer prohibitive (Kuiper, 1993). A much greater problem is that failure mechanisms of implant fixation are still poorly understood, hence it is difficult to define sensible objective functions. Of course, the "optimal implant" is only that for one particular bone, namely, the one

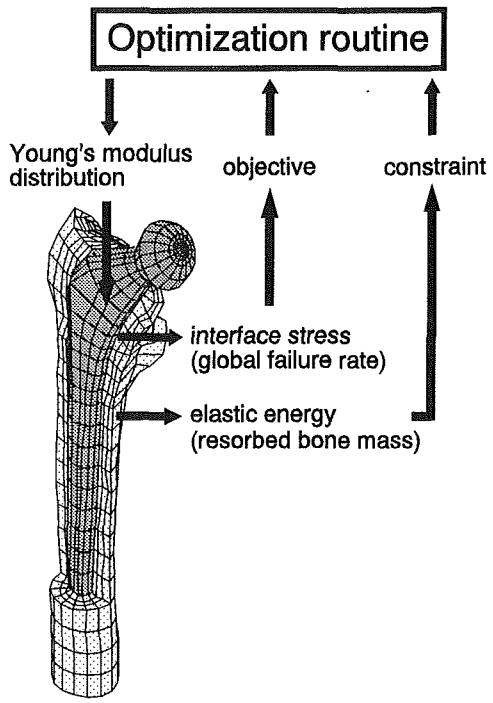

Fig. 1 Schematic depiction of a numerical optimization procedure for implant design (Kuiper, 1993)

represented in the FE-model. Because of variations in shape and properties, it will be suboptimal in most other bones. For that reason, the optimization results can as yet only be seen as general guidelines for a design process. As a consequence of this, very refined FE-models may overshoot the realistic goals of optimization studies.

Bone Shape and Architecture. Scientists have always assumed that bones are optimal constructions relative to their mechanical, load carrying function (Wolff, 1892; Thompson, 1917; Currey, 1984). However, it is still unclear for what, precisely, they are optimal structures. Is it minimal weight or minimal stress, or deformation? Are they not optimal structures, but only adequate ones? To paraphrase Nobel laureate Leon Lederman: if bone is the answer, then what is the question? The application of numerical optimization methods could provide answers to these questions. In a systematic fashion, different optimization goals can be tried until one reproduces their shapes and architecture. In studies of this kind, the question what governs the self-organizational *processes* of bone growth, maintenance and remodeling from the question of what kind of structure it eventually *produces* for the bone as a whole are separated, although process and product are obviously related.

A good example of such a (global) optimization study is the one from Luo (1992). In a 2-D FE model of the proximal femur he varied the apparent density of the bone and its trabecular anisotropy, as characterized experimentally by the fabric tensor (Cowin, 1985), subject to the goal of minimal overall compliance under the constraint of given overall bone mass. He showed a reasonable resemblance between the optimal result and the apparent density distribution and trabecular orientation in the frontal plane of the real femur. He also showed that the apparent density distribution provided more influential design parameters for the minimizing process than trabecular orientation, although the overall compliance was still 40 percent reduced when both aspects were considered, relative to only considering apparent density.

Fyhrie and Carter (1986) formulated a local optimization goal as a particular relationship between the local apparent density of bone and an "effective stress" to which it is sub-

jected. Carter et al. (1989) used this criterion for optimization of the density distribution in a 2-D model of the proximal femur. The result was not unlike the real density distribution. They showed that multiple loading cases must be used to obtain the most realistic results. Fyhrie and Carter (1990) studied a 3-D model of the femoral head, again using the same method. They showed that a criterion accounting for all stress components (like strain-energy density) was more likely to produce realistic bone density patterns than criteria that only included a part of the components Orr et al. (1990) used this method to explain bone remodeling patterns around implants. The optimization routine used in these studies cannot be readily classified as a global optimization routine with convergence to a specified global objective nor can it be classified strictly as a local self optimization process.

Other optimization analyses have concentrated on local bone architecture. Hollister et al. (1993a) studied the interface bone around porous implants to test the hypothesis that a trabecular structure with minimal overall strain energy is formed. For that purpose they applied a topology optimization program, using homogenization theory, developed by Bendsoe and Kikuchi (1988), in a local FE model. Hollister et al. (1993a) found that the distribution of the material, predicted by the optimization routine, coincided with the bone distribution at the interface measured experimentally (Fig. 2). However, the predicted bone was more consolidated than that found experimentally. They concluded that interface bone may try to satisfy other physiological constraints in addition to being a mechanically efficient structure.

Whether bones are optimal structures relative to a particular global mechanical criterion, or merely adequate ones, is as yet uncertain. But the question remains intriguing and the methods presently available present exciting prospects for further research.

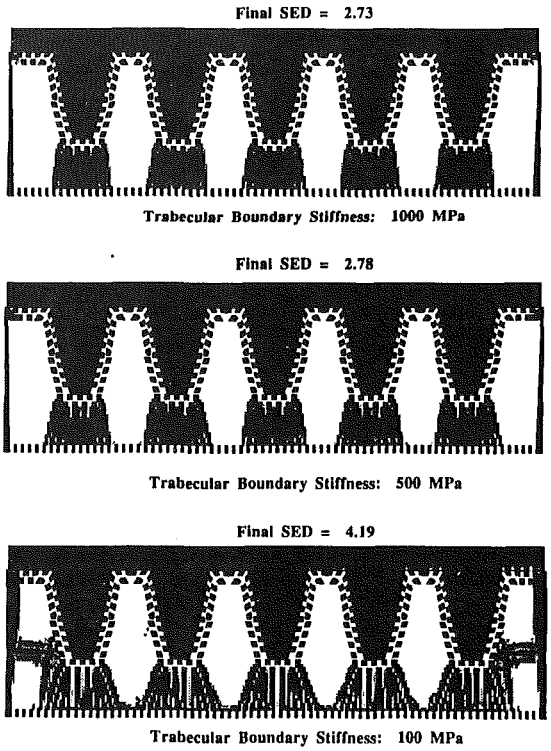

Fig. 2 Material distribution in a localized interface zone of a tibial component predicted using topology optimization for three surrounding bone stiffnesses (reproduced from Hollister, S. J., Kikuchi, N., and Goldstein, S. A., (1993) "Do bone ingrowth processes produce a globally optimized structure?", *J. Biomechanics*, Vol. 26, pp. 391-407; with permission)

Process Simulations

Most scientific problems to which FEA is applied do not just require information about the values of mechanical variables, but also about their consequences. It is nice to know the stress patterns in a joint-replacement structure, but the question is really how and when it will fail. Or how the bone, as a biological structure, handles mechanical stimuli. Most of the answers we seek depend on time-dependent processes. Be it mechanical failure mechanisms, biological processes, transient constitutive behavior of fluid-solid interactions, or steady-state dynamics. Without consideration and analysis of these phenomena, biomechanics remains a technology, rather than a science.

Much progress was made in recent years in the analysis of time dependent biphasic collagenous structures with FE-methods, such as articular cartilage (Suh et al., 1991; Spilker et al., 1992) and intervertebral discs (Yuan and Simon, 1992). Recently, triphasic theories were even developed to account for the swelling behavior of hydrated proteoglycans in these tissues (Lai et al., 1991; Snijders et al., 1992; Setton et al., 1993). These studies are certainly leading in providing realistic prospects for FEA of soft tissues in general.

Although of paramount importance for orthopaedics, hardly any FE-studies of implant failure mechanisms have been reported. One usually evaluates stress distributions and leaves it at that. Continuum damage theory (Chaboche, 1988a, 1988b) provides an excellent tool to investigate time-dependent failure mechanisms in correlation with clinical information of implant-fixation endurance. Verdonschot and Huiskes (1992) used this method in 2-D FE models of total hip replacement in order to estimate cement endurance depending on cement-stem bonding characteristics. Culleton et al. (1993) used crack-propagation theory to reproduce a cement fracture found in vivo in a 3-D FE model. Weinans et al. (1993a) recently formulated a theory for implant-bone interface resorption due to interface motion and used it to predict resportion patterns around several kinds of implants.

A FEA process simulation area which has seen much activity in recent years is that of bone adaptive behavior, which we will review more extensively below.

Bone Growth, Maintenance, Repair and Remodeling. The most intriguing—and for orthopaedics probably the most important—biological processes are the emergence and maintenance of bone structure and shape under the influence of mechanical stimuli. Roux (1881) hypothesized these processes to be regulated by cells, activated by local mechanical stimuli; a hypothesis later adopted by Wolff (1892) for his remodeling "law" (Roesler, 1987). However, it was not until the 1970's that a consistent mathematical theory for bone remodeling was developed, known as the theory of adaptive elasticity (Cowin and Hegedus, 1976). According to this theory, a bone has a characteristic equilibrium configuration, represented by a particular shape and apparent-density distribution; the latter representing its homogenized porous internal structure. This equilibrium configuration is assumed to produce an equilibrium (or reference) strain field in response to a typical external load. A deviation in the strain field (for example due to a change in external loads) becomes the driving force for an adaptation of shape (external, or surface remodeling) or apparent density (internal remodeling) towards the equilibrium strain field. The theory and its early applications to surface remodeling of long bones are reviewed in this issue (Cowin, 1993). Later, the external remodeling theory of Cowin was adapted for application in conjunction with 3-D FE-models (Hart et al., 1984a, 1984b). They were able to predict surface-remodeling phenomena in the sheep tibia, after removal of the fibula, similar to an experiment reported by Goodship et al. (1979). Hart (1990) also used this model to demonstrate that

maturation in bone mineralization may be an efficient mechanism for reducing overload strain.

Huiskes et al. (1987) used a similar theory to study the effects of orthopaedic implants on long-term adaptive bone remodeling in 2-D FE-models. Only surface remodeling was considered and the local deviation in the strain-energy was used as the remodeling signal, instead of strain. It was shown that long-term bone loss around intramedullary hip stems in animal experiments could be reproduced by the theory, provided that a region of reduced sensitivity around the equilibrium strain energy is accounted for (a "lazy" or "dead" zone), thus confirming theories of Frost (1983) and Carter (1984). A similar formulation for internal remodeling—i.e., adaptation of bone apparent density—was used in 2-D and 3-D FE models of femoral hip replacement, to study the effects of patient parameters and prosthetic design parameters such as shape, materials and bonding conditions on long-term post-operative bone loss due to adaptive remodeling (Huiskes et al., 1989, 1992; Weinans et al., 1992a, 1993b).

Huiskes and associates generalized their adaptive model by combining internal and surface remodeling in the same iterative process, using the pore-surface theory of Martin (1972). The theory of Carter et al. (1987a, 1987b) was applied to normalize strain energy to apparent density, in order to approximate the tissue strain energy rather than the apparent one, and to apply an integrated series of external loading cases to represent the loading history of daily activity. The generalized theory was applied in validation studies to simulate various series of animal experiments with canine hip replacements (Sumner et al., 1992). The similarity between actual animal bone morphology after 2 years and the predictions by the model were excellent, in individual detail as well as in terms of statistical averages (Weinans et al., 1993c; van Rietbergen et al., 1993a).

A different approach was taken by Prendergast et al. (1992), who developed a remodeling theory based on damage accumulation in bone. An advantage of this theory is that it automatically accounts for dynamic loading history as the driving force for the remodeling process. They simulated the same sheep experiments as Hart et al. (1984b), with good results as well. They also studied bone remodeling around femoral stems, in a FE-model similar to that of Huiskes et al. (1987), again with very similar results. These findings suggest that alternatives to strain energy or strain as mechanical signals may well produce equally realistic results. Mattheck and Burkhardt (1990), for example, use von Mises stress as the driving force behind surface remodeling with reasonable results as well.

The empirical models discussed above are limited to studies of adaptive bone remodeling, from one equilibrium configuration to another, due to a change in external loads or due to the implantation of a prosthesis. In order to determine a reference distribution of the signal value, a FE-analysis of the initial equilibrium configuration is required. However, the equilibrium configuration is in reality also maintained in a dynamic process of bone turnover, and formed during growth and repair processes. It is likely that all these processes use the same adaptive mechanism, the local biological control process of bone regulation suggested by Roux (1881) and Wolff (1892). Such a process is depicted schematically in Fig. 3. The actors in this control scheme are osteoblasts for bone formation, and osteoclasts for resorption. The sensor, which evaluates the mechanical signal and mediates the actors in its environment, is assumed to be the osteocyte (Cowin et al., 1991). In the adaptive models discussed above, it is assumed that the target of each sensor is to reach the mechanical signal value to which it is normally used (a site-specific reference signal). However, in a generalized regulation process, it is more likely that the target is the same for all sensors. This assumption touches upon the local self-optimization criterion of Fyhrie and Carter (1986), discussed earlier. Beaupré et al. (1990a) incorporated this theory in a time-dependent process formulation to be used

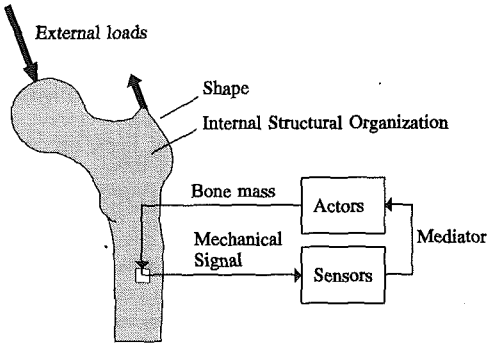

Fig. 3 Scheme of a self-organizational biological control process of local bone remodeling, in accordance with Roux's hypothesis (see text)

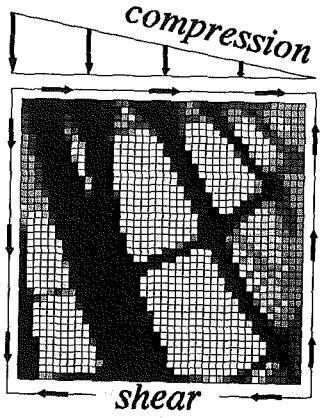

Fig. 4 Trabecular morphology generated in a local FE-model from an initial uniform density, using a self-optimization scheme representing the control process in Fig. 3, with particular sensor density, sensor domain, and external load (Mullender et al., 1993)

in conjunction with FE-models. They compared this method to the optimization scheme applied earlier by Carter et al. (1989) to predict the density distribution in the proximal femur, discussed above. The results were very similar (Beaupré et al., 1990b). This method was applied to a 2-D FE-model of the acetabulum, reconfirming that multiple loading cases must be considered in order to generate realistic density patterns (Levenston et al., 1992). It was also used to simulate surface remodeling to study cortical bone growth (van der Meulen and Beaupré, 1993).

Weinans et al. (1990, 1992b), applying their bone-regulation theory to study local remodeling of trabecular bone structure, using constant strain energy as the local remodeling target, discovered an interesting paradox. The structure converged to a "checker-board" pattern of fully filled or empty elements in a chaotic process with self-organizational qualities. Although on the one hand the solution violates the continuum assumption on which the solution process is based, on the other hand the emerging morphology is not unlike bone itself. This paradox was further investigated by Harrigan and Hamilton (1992) and by Jacobs et al. (1992), who suggested smoothing procedures in the FE formulation to repair the continuum violation, while maintaining the predictive power of the model. Mullender et al. (1993) formulated a new theory, based on unrestricted sensor influence in its domain. This is a more physiological assumption when compared to the earlier restriction of sensor effect per element (see Cowin, 1993). It effectively separates the sensor distribution from the FE-mesh in the iterative solution process—making the solution FE-mesh independent—and produces results in accordance with continuum theory, while maintaining a trabecular morphology, depending on biological and physical parameters such as the sensor density and the dimension of its domain (Fig. 4).

This area of FEA applications is very exciting indeed, and it is likely to spread to adaptive processes in other tissues. Much work is still to be done, however, in stability and convergence studies of the process formulations applied, and in the area of experimental validation. Although iterative analyses of growth and repair processes have not been reported, certainly the innovative work of Carter and Wong (1988), Wong and Carter (1990), and Blenman et al. (1989) relating stress patterns to ossification during both growth and fracture healing provides a good starting point.

Micro-Structural Modeling

The above discussion of bone-remodeling analyses illustrates the problems encountered when trabecular bone is modeled as a continuous (homogenized) material (see also Cowin, 1993). In addition, it seems obvious that when cells regulate bone (and other tissues as well) by mechanical stimuli, it is the cell load or deformation itself that we ultimately wish to measure. Although a full-scale FE-model of a whole bone which produces stress and strain data on the submicroscopic level seems a rather utopian goal, recent developments in that direction have created fascinating prospects.

Full Scale Models of Trabecular Bone. Recognition that trabecular and cortical bone effective stiffness depends on microstructural architecture and that continuum analyses of bone are inherently limited (Harrigan et al., 1988) has led to increased development of finite element models of bone microstructure. Previous idealized trabecular bone microstructural models have provided qualitative insight into trabecular bone mechanics (Pugh et al., 1973; Gibson, 1985; Beaupé and Hayes, 1985; Williams and Lewis, 1982). However, only recently have two critical developments made it possible to approach full three-dimensional modeling of realistic trabecular bone microstructure. First, two digital imaging techniques, a nondestructive 3-D micro-computed tomography scanner (micro-CT; Feldkamp et al., 1989) and a 3-D serial reconstruction technique using microtome slices (Odgaard et al., 1990; Dalstra et al., 1993), have made it possible to construct precise 3-D finite element models of trabecular bone microstructure. Second, the development of element-by-element preconditioned conjugate gradient (EBE-PCG) techniques (Hughes et al., 1983; Carey and Jiang, 1986; Ferencz, 1990) have made it possible to analyze very large 3-D finite element meshes, because no global stiffness matrix is handled, which greatly reduces memory requirements and time.

Three groups, Fyhrie et al. (1992), van Rietbergen et al. (1993b), and Hollister and Kikuchi (1993), have combined the 3-D digital imaging techniques with EBE-PCG solvers to analyze bone specimens containing from approximately 12,000 bone elements in a volume of $1.75 \times 1.75 \times 1.75$ mm (Fyhrie et al., 1992) to 216,000 bone and marrow elements in a volume of $3 \times 3 \times 3$ mm (Hollister and Kikuchi, 1993) to over 296,000 bone elements in a volume of $5 \times 5 \times 3.5$ mm (van Rietbergen et al., 1993b) illustrated in Fig. 5. The results showed that tissue level strain energy density (SED) may range between 1 and 268 times continuum SED (van Rietbergen et al., 1993b), and that tissue level strains may range between -0.1 to 2.5 times continuum strains (Hollister and Kikuchi, 1993). Large scale trabecular bone models may be used in conjunction with mechanical testing to better understand failure mechanisms, remodeling and damage development at the trabecular level.

Microstructural Models of Cells. Perhaps the ultimate goal of microstructural modeling is to calculate the mechanical stress environment around connective tissue cells and relate this with cellular activity. No FEA work to that effect has been done yet in bone. Guilak et al. (1990) and Guilak and Mow (1992) analyzed chondrocytes within an articular cartilage explant using local-global two dimensional modeling (Fig. 6). Using

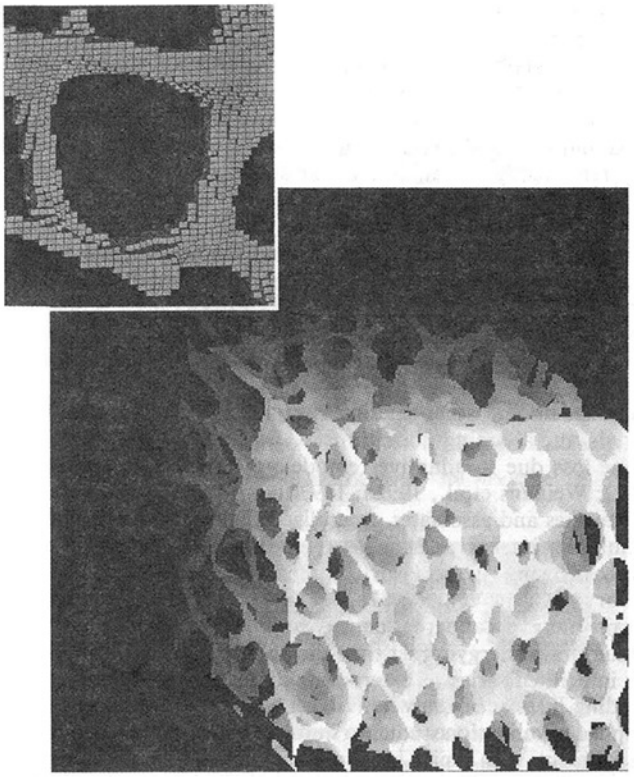

Fig. 5 Full-scale FE-model of an about $5 \times 5 \times 5$ mm trabecular bone specimen, containing over 296,000 elements. The full reconstruction and a detail are shown; every voxels is an 8-noded brick element (van Rietbergen et al., 1993b).

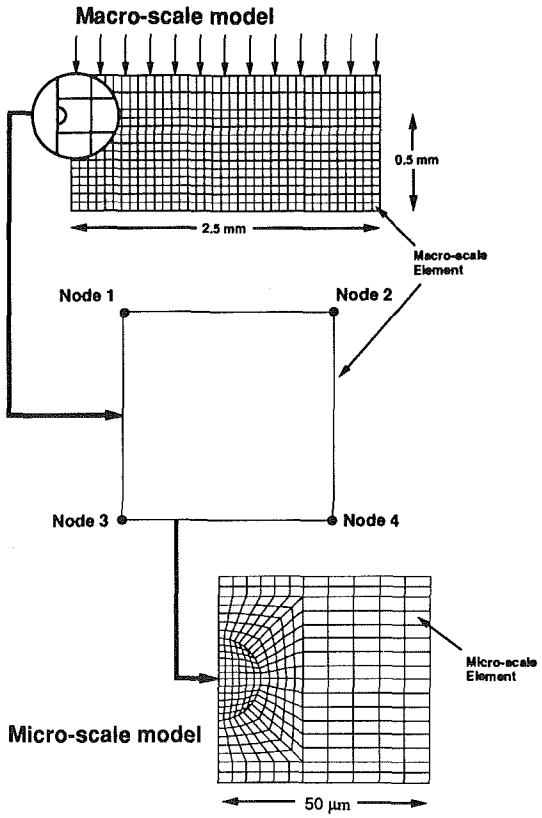

Fig. 6 Schematic illustration of the local global analysis procedure used to model chondrocytes within a cartilage explant (Guilak and Mow, 1992)

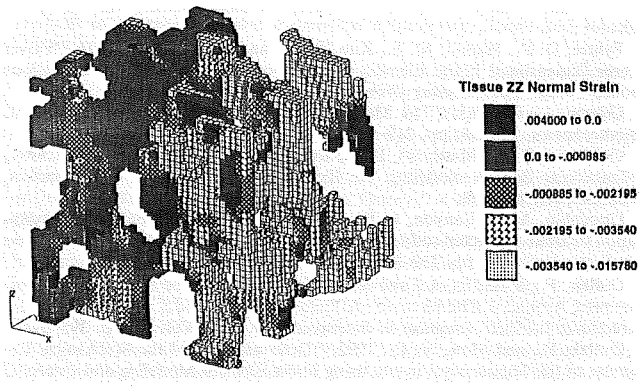

Fig. 7 Tissue strain distribution within a 2 x 2 x 2 mm volume of canine trabecular bone after 38 weeks of controlled implant load computed using homogenization sampling analysis (Hollister and Goldstein, 1993)

this model, Guilak and Mow (1992) parametrically studied the effect of cell elastic properties, cell shape, and pericellular matrix permeability on stress, strain and fluid velocity in and around the chondrocyte. When the chondrocyte was stiffer than the surrounding extracellular matrix (ECM), higher stresses occurred within the cell. If the ECM was stiffer, higher stresses occurred at the cell ECM junction.

Homogenization Analysis of Bone and Bone-Implant Interfaces. The primary difficulty with microstructural modeling is relating global to local level quantities. Relating global and local quantities is done through a Representative Volume Element (RVE) aproach (see Hill, 1963; Hashin, 1983). A recently developed, mathematically rigorous RVE method is homogenization theory (Sanchez-Palencia, 1980; Bakhvalov and Panasenko, 1989). Crolet et al. (1988; 1993) applied homogenization theory to analyze idealized models of two cortical bone microstructural levels including haversian systems and lamellae. They calculated the transversely isotropic material constants for cortical bone, finding results similar to experimental values. Hollister et al. (1991) applied homogenization theory to analyze idealized models of trabecular bone. Finding effective stiffness estimates inconsistent with experimental results. Hollister and co-workers (1992; 1993b) developed a technique for homogenization analysis of trabecular bone micromodels based on precise three-dimensional digitized images of trabecular bone architecture and a specially written EBE-PCG homogenization finite element solver. This technique, denoted the homogenization sampling method, gave effective siffness estimates within 30 to 40 percent of experimental values on average.

The homogenization sampling method can estimate tissue strains throughout whole joints for arbitrary global loads. Hollister and Goldstein (1993) calculated trabecular tissue strain estimates under controlled implant loads in a canine porous coated implant model (Goldstein et al., 1991). A significant portion of trabecular tissue strains ranges were greater than 15,000 microstrain in compression (Fig. 7). Goldstein et al. (1991) observed continued remodeling and woven bone in that particular animal suggesting that a damage repair response was initiated.

To better understand local implant bone interface mechanics, Ko et al. (1992) and Kohn et al. (1992a; 1992b, 1993) applied homogenization theory to analyze both screw-thread dental implant-bone interfaces and porous-coated implant-bone interfaces. Kohn et al. (1992a) found the porous-coated interface region to have transversely isotropic effective stiffness with a modulus of 2.3 GPa normal to the interface and a modulus of 25.7 GPa tangential along the interface. Tissue stresses at the bone-implant interface were generally concentrated in areas where the outer edge of the bead contacted bone, similar to results obtained by Pedersen et al. (1991) using a direct FE analysis.

Discussion

FEA has come to fill three important roles in orthopaedic research. First, it can be used as a *method of data-evaluation*, to interrelate experimental data from different time periods or experiments and integrate them into explanatory models for biological phenomena. Due to its analytic bases, it can be much more powerful than a statistic model in this respect. Second, it can also be applied as a *method of data-extrapolation.* With FEA, variables can be estimated which could not have been determined in any other way. The studies on micro-mechanical models of trabecular bone and cells, in which stress information from a higher level of tissue organization is extrapolated to a lower one, are good examples of this role of FEA. Much too often, FEA is still seen as an alternative to experimental mechanical analyses, with each having inherent advantages and disadvantages (Lee, 1992). The combination of FEA and experimental analysis, however, is many times more powerful than the sum of their individual applications. Another example of this role of FEA is the extrapolation of animal-experimental data to human cases.

The third role of FEA is as a *method of numerical experimentation*, very similar in concept to other methods for laboratory experiments in orthopaedic research. Experimental models can be categorized according to their relationship with reality on the one hand, and the degree of control over the experimental parameters on the other. A patient is very real, but many factors affect the outcome of clinical research over which the investigator has little control. On the other side of the scale, a numerical model does not include the multiple factors that affect clinical results, but it does provide the advantage of maximal experimental control. It offers the opportunity, for instance, to isolate one single factor of a system and evaluate its individual effect on a particular process. Obviously, this only provides clues where it concerns the reality one wishes to unravel, but providing clues is also a characteristic of laboratory experiments. Validating how well an experiment mimics reality is a problem of science in general, not just of FEA. Furthermore, numerical experiments have generated a number of useful hypotheses. It is interesting to note, for instance, that Wolff (1892) derived his famous "Law" directly from the results of a mathematical model; a graphic-statics model very similar in concept, but much more primitive than a FE-model.

Where practical aspects of FEA in orthopaedics are concerned, a number of applications are obvious. It is a useful tool in the search for design criteria of joint replacements or other methods of surgical reconstruction. It does not provide the bottom line, but may provide a good starting point. The cost-effectiveness of FEA is a particularly important asset in this case. FEA is very cheap compared to clinical, animal or laboratory testing methods. This makes it a profitable tool for design evaluations and preclinical testing of implants as well. Finally, it is useful as a tool to investigate cause-effect relationships in clinical failure mechanisms.

In 1983 the first author reviewed applications of FEA in orthopaedic biomechanics ten years after its introduction. He wrote that ". . .FEA. . .has created an exciting environment with potential applications previously undreamed of." The contemporary work discussed here shows that now, precisely ten years later, dreams are being fulfilled.

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