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ON SOME OF THE BIOMECHANICAL ASPECTS OF THE PELVIC BONE

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The pelvic bone forms the connection between the spine and the leg. This literally crucial location poses quite some demands on this bone from a mechanical point of view. It has a complex, warped shape; partly to provide effective insertion areas for muscles and ligaments, partly to provide the necessary contact areas for its neighboring bones in the skeleton. Like for all bones, its shape has evolved from the mechanical function and the demands, which come with it. Yet, little is actually known about the mechanics of the pelvic bone. In the following, we will take a closer look at some of the mechanical apects of the pelvic bone. Topics like structure and mechanical properties of pelvic trabecular bone, load transfer across the pelvic bone and the mechanics of acetabular components of a total hip arthroplasty will be discussed.

THE BONE STRUCTURE OF THE PELVIC BONE

A mature pelvic bone is an osseous integration of three separate parts: the iliac bone, the ischial bone and the pubic bone. These three merge, forming the acetabulum, the socket of the hip joint, through which the pelvic bone interacts with the femoral head. The pelvic bone is also connected to the sacrum by the sacroiliac joint and to the contralateral pelvic bone by the pubic symphysis. The relative rigidity of the latter two joints provides mechanical stability to this closed link of bony structures, referred to as the pelvic ring. The pelvic bone consists mainly of trabecular bone. Only a thin layer of cortical bone covers its surface. The thickness of this layer varies locally and ranges from less than 1 mm, for example found at the sacroiliac joint, to around 3 mm, found at the incisura ischiadaca major (Figure 1). With this thin cortex and the trabecular core the pelvic bone resembles a so–called sandwich construction, which is used in structural engineering to combine high strength and low weight. The thin (but strong) cortical shells are able to provide resistence against bending, because they are kept apart by the trabecular core, acting like a spacer. We will come back to this later on, when discussing the load transfer across the pelvic bone.

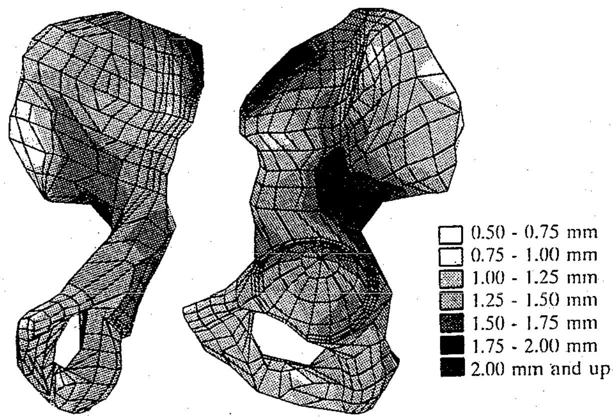


Figure 1: Distribution of the thickness of the pelvic cortex (frontal and lateral view). Values are measured from a series of CT–scans of a pelvic bone of a 89 year old male donor and are subsequently transposed onto a finite element model of a pelvic bone.

Trabecular bone is known to have many different appearances throughout various locations in the skeleton. With the apparent density and the orientation of the trabeculae as main parameters, it can adjust its local architecture (and thereby its stiffness and strength) to the mechanical circumstances it is subject to. When looking at the material properties of the trabecular bone in the pelvic bone, it can be said that it resembles more the trabecular bone in vertebral bodies than the trabecular bone in the femoral head or in the proximal tibia. Values for the Ca–equivalent density of pelvic trabecular bone, measured with dual energy quantitative CT scanning on a series of bones, ranged from 0.04 to 0.22 g/cm^3 (Dalstra *et al.*, 1993). These values correspond well to values of vertebral Ca–equivalent densities, reported by Lang et al. (1988) and Kalender et al. (1989). The density of femoral and tibia trabecular bone is usually higher, and consequently also the Young's (elastic) modulus and the strength, which are both highly dependent on the density. Values for the Young's modulus of pelvic trabecular bone hardly exceeds 100 MPa (Dalstra et al., 1993), whereas the Young's modulus of tibial trabecular bone can easily reach values of around 1500 MPa. So this means, that pelvic trabecular bone is less stiff and less strong than femoral and tibial trabecular bone. The reason for this is the fact, that pelvic trabecular bone has not a direct weight–bearing function like the trabecular bone in the femoral head or in the proximal tibia. Its principal loading mode as core material in the above mentioned sandwich construction is shear loading and it is therefore, that a predominantly plate–like architecture of the trabeculae (with the plates orientated perpendicular to the cortex) is observed in the pelvic bone (Figure 2). One single particular principal direction of the trabeculae, like for instance the longitudinal direction in the tibial plateau or in vertebral bodies, does not exist for pelvic trabecular bone.

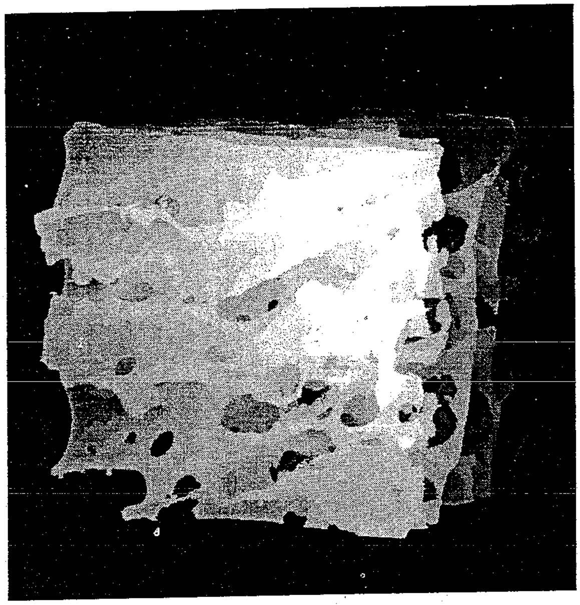


Figure 2: The typical 'lasagna-like' appearance of parallel plate-like trabeculae in pelvic trabecular bone (cube size approximately 6.5mm).

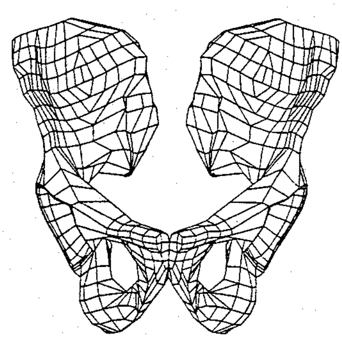
LOAD TRANSFER ACROSS THE PELVIC BONE

The primary task of the pelvic bone in the skeleton is to support the weight of the upper body and to transfer this load onto the lower extremities. While performing this task, the pelvic bone is subjected to relatively large loads: powerful muscles are attached to the pelvic bone and the magnitude of the hip joint force is quite high as well. Even during normal daily activities, like walking, the hip joint force can easily reach a multiple times body weight (e.g.: Bergmann et al., 1990). Interest in the load transfer across the pelvic bone has been growing the last number of years, especially with regard to the design of the acetabular component of a total hip arthroplasty. Due to its 'freakish' shape, analytical methods to calculate stresses and/or strains in the pelvic bone can not be used. Therefore, either experimental methods, which make either use of strain gages (e.g.: Jacob et al., 1976; Finlay et al., 1986) or photoelastic techniques (Yoshioka and Shiba, 1981), or computer simulations of the mechanical situation, using the Finite Element Method (FEM), have to be used to supply information about internal stresses and strains. In the following, we will describe the use of the latter method to study pelvic load transfer.

Since its first introduction into the field of biomechanics in the early 1970's, the FEM has become a well established tool to study the mechanics of bone and bone/prosthesis configurations. The FEM is basically a mathematical approach, whereby a structure is divided into small geometrical entities, the so-called elements. For each of these elements there exists an analytical stress-strain relationship, given their geometry, material properties, boundary conditions and external loading. However, the sheer amount of elements simply prevents a 'manual' solution and therefore, these problems are solved with a computer. Initially, geometrically regular bones, like the femur, were analyzed this way. The first pelvic model appeared in 1978 (Goel et al., 1978). Surprisingly, this, and another early pelvic model by Oonishi et al. (1983), were already three-dimensional. Surprisingly, because these type of models demand much more pre- and postprocessing (making the model and handling the data after calculation) and more computer power than needed for two-dimensional models. Pelvic FE models in later studies, in which the focuss was put on the stress situation in the immediate surrounding of an acetabular cup, were two-dimensional under the assumptions that either the acetabular region could be considered to be axisymmetric (Pedersen et al., 1982) or that the main load transfer would take place in only one plane anyway (Carter et al., 1982; Vasu et al., 1982).

To be able to study the full three-dimensional stress situation around acetabular implants, we recently developed a three-dimensional FE model of a pelvic bone (Dalstra, 1993; Dalstra et al., 1995). To study load transfer across the normal pelvic bone, the contralateral bone was also taken into account (Figure 3). The total number of elements in this model was 2,602. Local distributions of the apparent density of the trabecular bone and the thickness of the cortex were directly measured from CT-scans and transposed (the apparent densities were first calculated into Young's moduli) to the appropriate elements in the model. The model was assumed to be supported at both sacroiliac joints.

Figure 3: frontal view of the FE model of the pelvis



External loading of the model included the hip joint force and twenty two muscle forces. Eight phases in a normal walking cycle were simulated. The directions and the magnitudes of the hip joint force for those phases were based on telemetry data by Bergmann et al. (1990). The maximal value of the hip joint force occurred at the beginning of the one–legged stance phase and was in this case 2,158 N. To ensure a smooth introduction of the hip joint force onto the acetabular surface, the femoral head was partially modelled as well and the hip joint force was applied to this part of the model. Contact between the femoral head and the acetabulum was then described by so–called gap elements, which only allow the transfer of compressive forces. Furthermore, this contact was assumed to be frictionless to simulate the (non–present) articular cartilage in the acetabulum. The directions of the muscle forces were found by subtracting the coordinates of the distal and proximal insertions (Dostal and Andrews, 1981), while the magnitudes of these forces were based on data reported by Crowninshield and Brand (1981). The single highest value in this array of muscle forces was the force in the gluteus medius at the end of the one–legged stance phase and numbered 1,509 N. A mapping of the physiological insertion areas of each of the muscles was made onto the FE model and muscle forces were applied as distributed loads on the surfaces of those elements, which were located in these respective areas of insertion.

Although primarily developed to study load transfer after acetabular reconstruction, the developed FE model also opened the possibility to study load transfer in a normal pelvic bone. In Figure 4 the stress intensity (the so–called von Mises stresses) in the cortical shell is shown for the eight considered phases in a walking cycle.

As might be expected stress levels are highest during the stance phase of the affected leg. The highest stresses occur in the attachment area of the gluteus major (even though the magnitude of the muscle force of the gluteus medius is larger than that of the gluteus major) and are in the order of 30 MPa. It is during the stance phase, that the hip joint force becomes so large, that the pelvic bone has a tendency to be tilted internally upward around an axis running from the pubic symphysis to the sacroiliac joint. This motion is countered predominantly by the powerful gluteus muscles stabilizing the iliac wing.

Stresses in the pubic bone are also relatively high, due to the connection with the contralateral pelvic bone. In particular, stresses here are high at the beginning of the swing phase, which coincides with maximal activity levels of the muscles inserting in this region: the three adductor (magnus, longus and brevis) muscles, the pectineus and the gracilis. Due to the absence of powerful muscles and other connections to the skeleton the ischial bone remains comparatively low–stressed. Also the cortical (subchondral) bone in the acetabulum is relatively low–stressed: even though the hip joint force reaches nearly four times body weight, it is distributed over a relatively large articular surface.

When we think the entire cortex 'peeled off' and we take a look at the stresses in the underlying trabecular bone (Figure 5), then the nature of the pelvic 'sandwich construction' becomes clear: the trabecular core is being 'stress–shielded' by the stiff cortex, resulting in stresses here, which are about 50 times as low as in the cortex.

Figure 4: Lateral views of the stress intensity (von Mises stresses) in the cortical shell of the left pelvic bone for eight phases during a walking cycle: 1) double support, beginning left stance phase, 2) beginning left single support phase, 3) halfway left single support phase, 4) end left single support phase, 5) double support, end left stance phase, 6) beginning left swing phase, 7) halfway left swing phase, 8) end left swing phase.

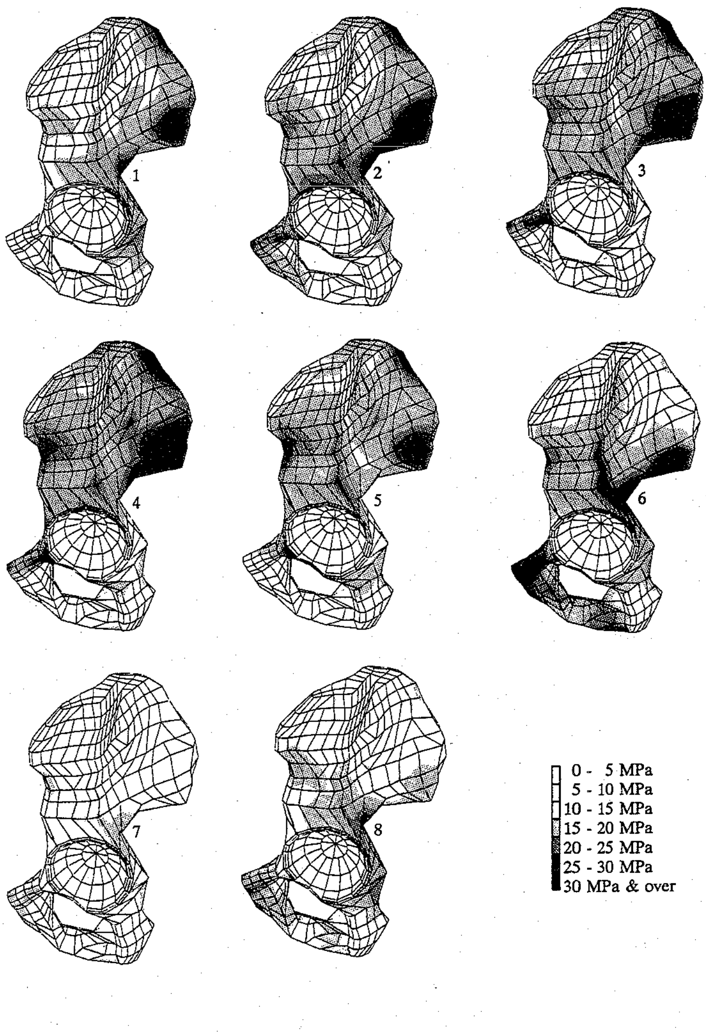
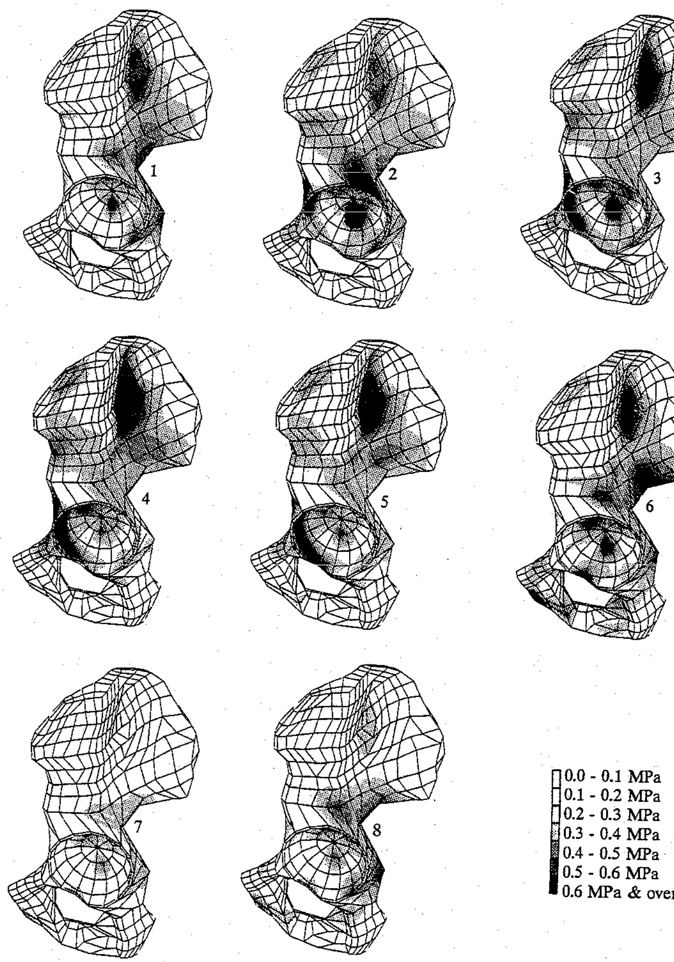


Figure 5: Lateral views of the stress intensity (von Mises stresses) in the trabecular core of the left pelvic bone for the same eight phases as in Fig. 4.



Furthermore, we can see that the locations of the stress 'hotspots' have changed as well. Throughout the entire walking cycle the highest stresses are still found during the stance phases, but in the trabecular core it is the centre of the iliac wing, where the highest stresses occur. Also at the anterosuperior acetabular rim and at the bottom of the acetabulum trabecular stresses are relatively high. In fact, it is in these very locations, where the sandwich construction analogon plays a lesser role: in the centre of the iliac wing, because here the medial and the lateral cortical shell practically merge into one and in the acetabulum, because here the principal loading mode is direct load–bearing rather than bending.

When we look in more detail at the contact in the acetabulum between the femoral head and the pelvic bone, than it becomes clear that this is far from uniform. During the eight considered loading phases, there is constantly contact along most of the acetabular rim, the central part of the anterosuperior quadrant of the acetabulum and the bottom of the posteroinferior quadrant, while a large part of the posterosuperior quadrant of the acetabulum is not in contact with the femoral head at all throughout the entire loading cycle (Figure 6).

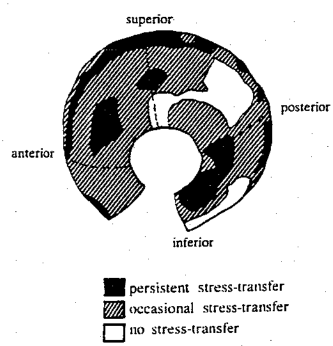


Figure 6: The nature of stress transfer (contact) between the femoral head and the pelvic bone during a complete walking cycle

Pauwels (1973) argued that for a normal configuration of the hip joint, the stress distribution in the acetabulum should be uniform. This idea, however, is based on the assumption, that the acetabulum can transfer loads in all directions, while the present results indicate, that loads are mainly transferred from the acetabulum through the lateral cortical shell to the sacroiliac joint and to the pubic symphysis. The actual stress distribution in the acetabulum is quite affected by this load transfer mechanism and the deeper parts of the acetabulum are stressed much less. The high stresses at the superior acetabular wall demonstrate its importance in the natural load transfer mechanism of the hip joint. In dysplastic acetabula, where this part of the wall is underdeveloped or sometimes even lacking, an alternative load transfer mechanism with higher stresses to compensate for this will be the result, which is shown by Schüller et al. (1993) in case of reconstructed acetabula. Therefore, a dysplastic acetabulum can definitely be considered to be a considerable risk factor for progressive wear of the hip joint.

ACETABULAR PROSTHESES AND PELVIC MECHANICS

The reconstruction of a hip joint with a total hip arthroplasty is primarily meant to relieve pain and restore functionability of the joint. Yet, at the same time it creates a rather unnatural situation from a mechanical point of view. At the femoral side a metal stem is inserted into the medullary canal. Due to the high stiffness of this stem compared to the stiffness of the bone shaft, the surrounding cortex becomes stress–shielded (it does not transfer so much load anymore as before). This phenomenon carries the potential danger of a local reduction of bone mass (Wolff's Law: bone in disuse will gradually disappear), which may eventually lead to a loosening of the implant due to the lack of supporting bone stock. At the pelvic side the acetabulum is reamed to allow the placement of an acetabular implant. In general these implants are cups, which are made from high density polyethylene, but individual designs may differ in shape (hemispherical or conical), type of fixation (cemented or non–cemented) or the presence of a metal backing. The insertion of such a cup creates also an unnatural situation, but its consequences are not so directly apparent like on the femoral side. In this last part, we will take a more detailed look, how the mechanical situation in the pelvic bone changes due to the presence of an acetabular implant.

We used the same model as described before, with some additions to account for an acetabular implant. Loading conditions were kept the same as before. Figure 7 shows a comparison between the stresses in the cortical shell of both a normal pelvic bone and a pelvic bone reconstructed with a traditional hemispherically shaped cemented cup.

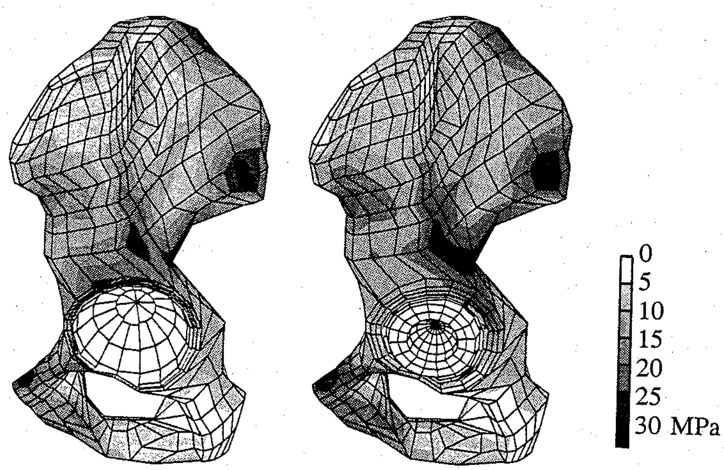


Figure 7: Comparison between the stress intensity (von Mises stresses) in the trabecular core of the left pelvic bone for the same eight phases as in Fig. 4

Differences between both stress distributions are only marginal and are restricted to the immediate vicinity of the acetabulum. In the superior acetabular rim the load transfer has shifted slightly to posterior after reconstruction. For both the subchondral and the trabecular bone in the acetabular region, the changes are more substantial. In the normal case the highest stress in the subchondral bone occurs in the anterosuperior quadrant of the acetabulum. In the reconstructed case stresses have not only reduced considerably, but have also shifted from the dome area towards the edges (in particular the posterosuperior edge) as can be seen in Figure 8.

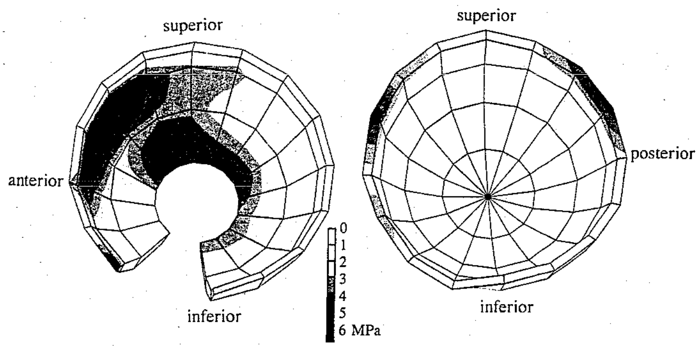


Figure 8: Comparison between the stress intensity (von Mises stresses) in the subchondral bone layer of a normal pelvic bone (left) and a reconstructed pelvic bone (right) during one-legged stance

This leaves the deeper areas in the acetabulum stress-shielded. Two effects seem to play a role in this. Firstly the addition of material in the acetabulum makes the pelvic bone more stiff, which leads to a general stress reduction in those locations, where the deformations used to be highest (stiffening effect). Secondly the transfer of the anterosuperiorly directed hip joint force to the posterosuperiorly directed support area of the sacroiliac joint takes already place within the cup and the cement mantle instead of in the subchondral bone layer and the lower ilium as in the normal case (load-diverting effect). In the underlying trabecular bone this leads to an increase of the stresses in the superior acetabular rim area (Figure 9). At the lower anterior side of the ilium and in the pubic bone trabecular stresses decrease slightly after reconstruction.

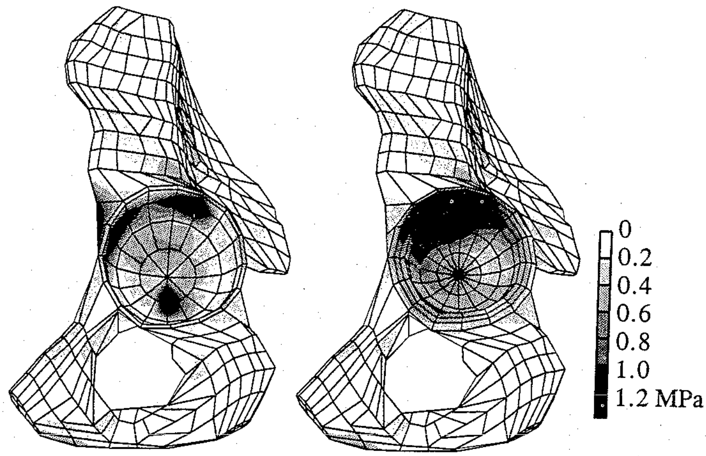
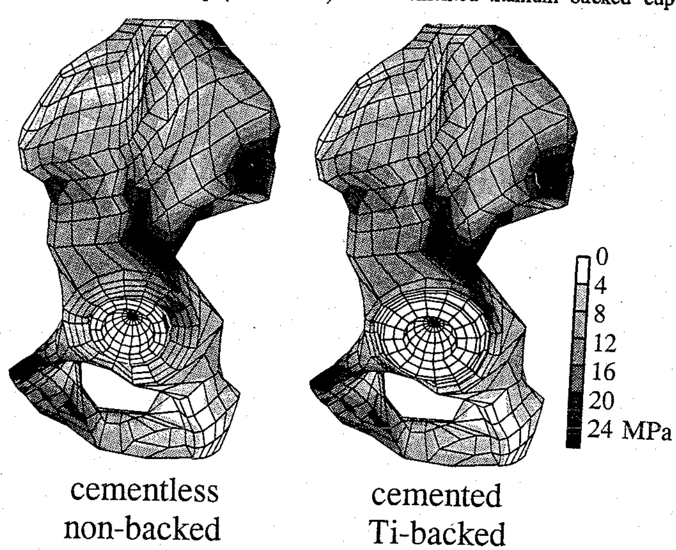


Figure 9: Comparison between the stress intensity (von Mises stresses) in the trabecular bone of a normal pelvic bone (left) and a reconstructed pelvic bone (right) during one-legged stance

As might be expected after the description above, the effect of a change in the design of an acetabular implant is also just a local affair. Yet, it may be important for a good understanding of the mechanics of acetabular implants. This is best illustrated by looking at the stresses in the subchondral bone layer. Figure 10 shows these stress patterns for four different types of acetabular cups (cemented/uncemented and metal-backed/non-backed).

Figure 10: Comparison between the stress intensity (von Mises stresses) in the suchondral bone layer of pelvic bones, reconstructed with a cementless full-polyethylene cup (top left), a cementless titanium-backed cup (top right), a cemented full-polyethylene cup (bottom left) and a cemented titanium-backed cup (bottom right).



In all four cases stresses are highest at the anterosuperior acetabular rim. The principal effect of adding a cement layer is a shift of the stresses from the anterosuperior rim to both the posterosuperior and the anterosuperior rim. This phenomenon is comparable to the load-diverting effect mentioned above. The principal effect of a metal backing on the other hand is more comparable to the above mentioned stiffening effect, and causes a shift of the load transfer from the deeper areas of the acetabulum to the acetabular rim. The clinical consequences of this may be detrimental in two ways: the reduction of the stresses underneath the cup might lead to unwanted local bone resorption and the increase of the stresses around the rim might lead to an overstressing of the prothesis/bone interface in this area (Schmalzried et al. (1992) actually demonstrate, that the prosthesis/bone interface of cemented metal-backed cups starts to fail along the acetabular rim). The reduction of the stresses underneath the cup in case of metal backing was also observed in earlier pelvic models (Carter et al., 1982; Pedersen et al., 1982), but then it was seen as a positive aspect of metal backed acetabular cups. In fact, to some extent the results of these studies were used to promote metal backing for acetabular cups. However, clinically metal-backed cups have not quite lived up to their expectation (e.g.: Ritter et al., 1990). So, even though in the 'real-life' situation additional effects, like wear particles (Schmalzried et al., 1992), play an important role in the failure mechanism of acetabular implants, our analyses have shown, that it is possible to point out possible causes for failure with a mechanical background.

CONCLUSION

The pelvic bone is a relatively high-loaded bone. As a structure, it is able to deal with these high loads effectively by having evolved into a natural analogon of a sandwich construction, whereby the cortical shells bear the bulk of the load. The main load transfer takes place between acetabulum and sacroiliac joint and a minor part goes from the acetabulum to the pubic symphysis. This overall load transfer is not much changed by the reconstruction of the acetabulum by an acetabular implant. Changes in the stresses in the bone after such a reconstruction occur only in the immediate vicinity of the acetabulum. Despite this local effect, however, some of these changes, like stress-shielding of the bone in the dome of the acetabulum, may be of great importance to determine the ultimate success of such an implant.

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