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LOAD TRANSFER ACROSS THE PELVIC BONE*

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Abstract—Earlier experimental and finite element studies notwithstanding, the load transfer and stress distribution in the pelvic bone and the acetabulum in normal conditions are not well understood. This hampers the development of orthopaedic reconstruction methods. The present study deals with more precise finite element analyses of the pelvic bone, which are used to investigate its basic load transfer and stress distributions under physiological loading conditions. The analyses show that the major part of the load is transferred through the cortical shell. Although the magnitude of the hip joint force varies considerably, its direction during normal walking remains pointed into the anterior/superior quadrant of the acetabulum. Combined with the fact that the principal areas of support for the pelvic bone are the sacro-iliac joint and the pubic symphysis, this caused the primary areas of load transfer to be found in the superior acetabular rim. the incisura ischiadaca region and, to a lesser extent, the pubic bone. Due to the 'sandwich' behavior of the pelvic bone, stresses in the cortical shell are about 50 times higher than in the underlying trabecular bone (15 to 20 MPa vs 0.3-0.4 MPa at one-legged stance). Highest intraarticular pressures are found to occur during one-legged stance and measured about 9 MPa. During the swing phase, these pressures decrease less than linearly with the magnitude of the hip joint force. Muscle forces have a stabilizing effect on the pelvic load transfer. Analysis without muscle forces show that at some locations stresses are actually higher than when muscle forces are included.

INTRODUCTION

A mature pelvic bone is an osseous integration of three separate parts, the iliac, the ischial and the pubic bones. These three merge, forming the acetabulum, the socket of the hip joint, through which the pelvic bone interacts with the femoral head. The primary task of the pelvic bone in this interaction is to support the weight of the upper body and transfer it onto the lower extremities. In doing so, the pelvic bones have to withstand forces which are a multiple of that weight. Within the limits of the anatomical boundary conditions, the pelvic bone has evolved into a very efficient structure, which is well able to carry these large forces. Consisting mainly of low-density trabecular bone (Dalstra et al., 1993), which by itself is not strong enough by far to withstand such high loads, it is totally covered by a thin layer of cortical bone. In this way, it resembles a so-called 'sandwich construction', used in engineering to combine high strength and low weight (Jacob et al., 1976). However, besides this 'sandwichbehavior', little is known about the basic mechanics of the pelvic bone. Strain gage techniques (Finlay et al., 1986; Jacob et al., 1976; Lionberger et al., 1985; Petty et al., 1980; Ries et al., 1989) and finite element (FE) analyses (Carter et al., 1982; Dalstra and Huiskes, 1990; Goel et al., 1978; Huiskes, 1987; Koeneman et al., 1989; Landjerit et al., 1992; Oonishi et al., 1986; Oonishi et al., 1983; Pedersen et al., 1982; Rapperport et al., 1985; Renaudin et al., 1992; Vasu et al., 1982) are the two methods most frequently used for studying pelvic mechanics. Yet, these earlier experimental and FE studies notwithstanding, the load-transfer mechanism and the stress patterns of the pelvic bone under normal physiological conditions are still not well understood: partly because it fell beyond the scope of these studies, partly because the models used were not suitable to properly describe it.

The purpose of the present study was to evaluate the basic mechanics of the natural pelvic bone, using an experimentally validated sophisticated three-dimensional finite element model (Dalstra *et al.*, 1994). By incorporating new information about hip joint and muscle loads, it was our intention to prescribe external loading as realistically as possible. For once it is known how the pelvic bone behaves under normal loading conditions, we will be able to develop a better understanding of possible differences due to acetabular reconstructions in future studies. By varying the external loads, we also wanted to establish to what extent muscle forces are really important in describing the mechanics of the pelvic bone.

METHOD

In this three-dimensional FE analysis, a bilateral pelvic mesh was used consisting of a total of 2662 elements and 1982 nodes (Fig. 1). For one hemipelvis the subdivision into specific element groups was as follows: 365 isoparametric 8-node brick elements were used to represent the trabecular and the subchondral bone and 632 4-node membrane elements were used for the thin cortical shell. The spherical part of the

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Fig. 1. Frontal view of the three-dimensional mesh of the pelvic bone.

femoral head, interacting with the pelvic bone, was also modeled to ensure a smooth and realistic introduction of the hip joint force into the acetabulum (Huiskes, 1987). In this submesh 176 8-node brick elements were used for the trabecular bone and 128 4node membrane elements for the cortex. Finally, contact between the femoral head and the acetabulum was modeled by 60 gap elements, ensuring that only compressive forces could be transmitted from the femoral head onto the acetabulum. This contact was assumed to be frictionless, but articular cartilage was not included in this model, so that the femoral head was in direct contact with the subchondral bone of the acetabulum.

Based on the results of an earlier study (Dalstra *et al.*, 1994), the thickness of the membrane elements ranged from 0.7 to 3.2 mm with an average value of about 1.5 mm. Young's modulus and Poisson's ratio for these elements were assumed to be 17 GPa and 0.3, respectively. Based on the same study, the Young's modulus allocated to the pelvic trabecular bone ranged from 1 to 132 MPa; for the subchondral bone, the range was 186 to 2155 MPa. Examination of the

material properties of pelvic trabecular bone has shown that it is not highly anisotropic (Dalstra *et al.*, 1993). Therefore, assuming isotropy for these elements seems justified. The same study showed that 0.2 is a good approximation for the Poisson's ratio of pelvic trabecular bone. For the femoral submesh, the values for the cortical and the trabecular Young's modulus and Poisson's ratio were 17 GPa and 0.3, and 800 MPa and 0.2 respectively.

As kinematic boundary conditions for the FE model, the nodes situated in the sacro-iliac joint areas on both pelvic bones were kept fixed to simulate sacral support. Loading was applied to only the left hemipelvis. The role of the contralateral bone was to supply a more realistic elastic boundary condition at the pubic symphysis. The loading conditions for the model were based on walking. The hip joint force and the forces of 21 muscles attached to the pelvic bone were taken into account at eight characteristic phases of a normal walking cycle (Table 1). The values and directions of the hip joint force at these eight phases were based on data by Bergmann et al. (1990). By means of prostheses fitted with telemetry devices, they performed in vivo measurements of the hip joint force during all kinds of activities. The direction of the hip force in their measurements was given in a coordinate system relative to the femur and had to be transformed accordingly into a direction relative to the coordinate system of the present pelvic model. The relative position of the pelvic bone and the femur changes during walking. For the transformation calculations, a fixed adduction angle of 15° for the femur was assumed, while the angle between the pelvic bone (longitudinal body axis) and the femur in the A/P-plane (flexion/extension) was variable. Values for this angle were measured with a SELSPOT motion-analysis system (Selspot AB, Mölndal, Sweden) on one of the authors (M.D.) and are also given in Table 1. Furthermore, Bergmann and coworkers express the magnitude of the hip joint force as a percentage of the total body weight. In our particular case, a body weight of 650 N was assumed. The magnitudes of the hip joint forces in which this resulted are given in Table II. In the FE model, the hip joint force was applied as a distributed load on the head/neck section of the femoral head.

Table 1. Description of the load cases with respect to their occurrence within a walking cycle and the flexion/extension angle between the pelvic bone and the femur

Case	Description	Percentage walking cycle	Flexion angle
1	Double support, beginning left stance phase	2	22° (fl.)
2	Beginning left single support phase	13	18° (fl.)
3	Halfway left single support phase	35	4° (ext.)
4	End left single support phase	48	12° (ext.)
5	Double support, end left stance phase	52	14° (ext.)
6	Beginning left swing phase	63	2° (fl.)
7	Halfway left swing phase	85	31° (fl.)
8	End left swing phase	98	21° (fl.)

	Loading phase during gait								
	1	2	3	4	5	6	7	8	
Hip joint force	426	2158	1876	1651	1180	187	87	379	
Gluteus maximus	842	930	167	377	456	491	114	482	
Gluteus medius	1018	1053	1474	1509	1412	982	105	421	
Gluteus minimus	228	140	263	228	175	123	114	219	
Tensor fasciae latae	0	132	88	158	149	88	70	96	
Iliacus	0	0	0	228	307	272	0	0	
Psoas	149	0	316	175	88	175	105	140	
Gracilis	0	0	0	0	88	158	70	140	
Sartorius	0	88	0	0	35	158	88	88	
Semimembranosus	579	368	333	368	421	298	61	421	
Semitendinosus	0	140	105	246	316	368	105	0	
Biceps femoris longus	298	202	88	70	123	114	79	377	
Adductor longus	0	88	0	0	88	158	70	140	
Adductor magnus	0	0	0	0	132	263	0	0	
Adductor brevis	0	114	0	0	0	202	0	114	
Obturator externus	0	0	0	0	123	167	132	123	
Obturator internus	167	123	0	61	61	149	123	0	
Pectineus	0	0	175	96	0	149	0	0	
Piriformis	202	275	0	0	0	0	123	228	
Quadratus femoris	61	96	0	0	88	184	0	0	
Superior gemellus	140	88	123	79	0	0	158	202	
Inferior gemellus	0	0	0	0	0	140	79	149	
Rectus femoris	0	123	0	0	0	175	105	96	

Table 2. Magnitudes (in Newton) of the hip joint force and muscle forces at the considered eight load cases

Apart from the hip joint force, 21 muscles inserting onto the pelvic bone were incorporated in the model (Fig. 2). The directions of the muscles were found by subtracting the coordinates of their distal and prox-



Fig. 2. Identification of the attachment areas of the various muscles used in the model.

imal insertions (Dostal and Andrews, 1981), whereby the same rotation of the pelvic bone relative to the femur in the A/P-plane as mentioned above was taken into account. Because of their multiple lines of action, it was necessary to make a differentiation in ventral, central and dorsal parts for the gluteus minimus, the gluteus medius and the adductor magnus muscle. The magnitudes of the muscle forces were based on data by Crowninshield and Brand (1981). A mapping of the physiological areas of insertion of each of the muscles was made onto the finite element mesh and muscle forces were applied as distributed loads on the surfaces of the brick elements (due to the out-of-plane character of the loading, the loads could not be applied directly to the membrane elements) which were located in these respective areas of insertion. The magnitudes of the muscle forces during the eight considered phases of the walking cycle are given in Table 2 as well.

The various stress components and the von Mises stresses in the various materials were calculated. Strain rates were calculated by subtracting the von Mises strains (defined as the square root of two-thirds of the sum of the squared principal strains) from two consecutive load cases and dividing the result by the elapsed time between these phases (assuming one complete step to last 1.1 s). Because the load cases are static and no accelerations are taken into account, these strain rates are only very rough approximations, but they do give some indication where large changes in stresses are to be expected during walking.

To determine the influence of the muscle forces on the stress distributions in the pelvic bone, two additional cases were analyzed. Firstly, all muscle forces were omitted, leaving the hip joint force as the only external load. Secondly, the assumed body weight was increased from 650 to 800 N (resulting in a higher hip joint force), while the muscle forces were kept at their original level. These two analyses should give insight in how the hip joint force is transferred through the pelvic bone and it will disclose whether including muscle forces is necessary at all when analyzing implants in the future.

For the analyses, the MARC/MENTAT FEM and pre- and post-processing codes (MARC Analysis Corporation, Palo Alto, CA, U.S.A.) running on the EX-60 mainframe computer (Hitachi Data Systems, Bells Hill, Bucks, U.K.) of the University of Nijmegen were used.

RESULTS

The pelvic bone acts as a sandwich construction, which means that the major part of the load in the pelvic bone is transferred through the cortical shell. The stresses here are about 50 times higher than in the underlying trabecular bone. The locations of the highest stresses in the cortical shell and the underlying trabecular bone do not coincide. In the cortical shell, the highest stresses are found in the attachment area of the gluteus major muscle and the incisura ischiadaca major region (Fig. 3), while in the trabecular bone, the highest stresses occur in the thin central area of the iliac wing and in the acetabulum (Fig. 4). The accompanying strain rates were found to be between the orders of 0.001-0.1 s⁻¹. The highest strain rate occurred in the trabecular bone between the phases 8 and 1, and had a value of 0.4 s^{-1} (Fig. 5). At the end of the single leg stance (between phases 5 and 6) high strain rates occurred also at the anterior acetabular rim in the subchondral bone (0.35 s^{-1}) . In general, strain rates in the cortical shell were found to be lower than in the subchondral and trabecular bone.

In and closely around the acetabulum, the highest stresses occur in the superior acetabular wall and from there they are transferred to either the sacro-iliac joint or the pubic symphysis. As can be seen in Figs 3 and 4, the pubic bone is loaded most heavily at the beginning of the swing phase of the leg (phase 6). At this particular moment, the reaction force at the pubic symphysis reaches its maximal value of 750 N (115% BW). The reaction force at the sacro-iliac joint, however, is still more than four times as high. The hip joint force is pointing into the anterior/superior quadrant of the acetabulum in all eight loading cases considered, while the main area of support (the sacro-iliac joint) is located more posteriorly. Due to this, the cortical bone at the anterior/superior wall is still considerably stressed, while in the underlying trabecular bone the areas of high stresses have shifted more to both posterior and anterior.

The hip joint force itself is not distributed evenly over the acetabulum. In Fig. 6 those areas are identified, where during the eight phases considered compressive normal stresses in the subchondral bone occurred persistently (load-transferring contact) and where normal stresses were persistently zero (no loadtransferring contact). From this it can be concluded that during walking the load transfer between femoral head and acetabulum takes place predominantly along the anterior/superior edge of the acetabulum. The highest (compressive) normal stress occurs during one-legged stance (phase 3) and has a magnitude of 8.7 MPa, but even for low values of the hip joint force relatively high compressive normal stresses in the subchondral bone may occur (Fig. 7).

The muscle forces were found to have a considerable influence on the stress patterns in the pelvic bone. The analysis without muscle forces showed that the load transfer is now entirely directed along the axis from the sacro-iliac joint to the pubic symphysis (Fig. 8). The ischial bone and the superior part of the iliac bone remain virtually unloaded, while the pubic bone is more highly stressed than in the case that muscle forces were included (Fig. 3, phase 2). Increasing the hip joint force relative to the muscle forces showed that this affected only the stresses in the direct vicinity of the acetabulum. The stress in the subchondral bone responded almost linearly to the change (Fig. 9). The peak value of the von Mises stress increased from 7.0 to 9.3 MPa though, which is more than the 23% rise in the hip joint force. These analyses have shown that muscles forces are important with respect to the overall loading and deformation modes of the pelvic bone, yet unlike the hip joint force, their exact magnitudes are less important for studying the stresses in and around the acetabulum.

DISCUSSION

Finite element stress analyses of the normal pelvic bone have been described in only a few cases. Vasu et al. (1982) and Rapperport et al. (1985) based their respective models on two-dimensional sections through the pelvic bone. These kind of models, however, lack the ability to describe the three-dimensional aspects of pelvic mechanics adequately. Obviously, two-dimensional models are restricted to the plane of modeling, which in case of the pelvic bone is usually a cross-section through the pubic bone, the acetabulum and the sacro-iliac joint. Because of this, a twodimensional model lacks the re-enforcement of the out-of-plane part of the acetabular wall, thus making these models inherently too flexible. Loading is also restricted to the plane of modeling, which is a serious shortcoming. Due to its circumferential geometry, an axisymmetric pelvic model has to be restricted to the immediate vicinity of the acetabulum. Furthermore, it assumes that the acetabular wall is present around the full 360°, which is not the case in reality. Therefore, axisymmetric models may be able to indicate certain

Load transfer across the pelvic bone



Fig. 3. Lateral views of the von Mises stress distribution in the cortical shell for the eight load cases. Highest stresses mainly occur in the lower iliac bone and the public bone during the first six phases of the walking cycle.

trends, but for more detailed evaluations, a threedimensional model will be required as already demonstrated by Koeneman *et al.* (1989). Goel *et al.* (1978) and Oonishi *et al.* (1983) did use three-dimensional models, but since then additional data concerning pelvic loading and material properties of the pelvic bone have become available. We have aimed at making the present model as sophisticated as possible by using a three-dimensional mesh with material properties taken from quantitative computer tomography



Fig. 4. Lateral views of the von Mises stress distribution in the trabecular bone for the eight load cases. The highest stresses are found in the anterior/superior part of the acetabulum and the central iliac bone.

measurements, and applying realistic loads, which not only consisted of the hip joint force, but also included 21 muscle forces. With this model, the load transfer in the normal pelvic bone was evaluated and for this purpose, several phases in a walking cycle were considered as loading conditions. As models are an abstraction of reality, the model's results should always be interpreted in the light of the assumptions and limitations. In our model articular cartilage was not incorporated. This will certainly have an effect on the stresses within the acetabulum. However, the damping role of cartilage is of no



Fig. 5. Distribution of the momentary strain rates in the trabecular bone between cases 8 and 1.



Fig. 6. Identification of the areas on the articular surface of the subchondral bone, where stress transfer between femoral head and acetabulum is either continuous, intermittent or completely absent during a full gait cycle. Most of the load transfer takes place in the anterior/superior quadrant of the acetabulum.

consequence in quasi-static calculations like the present one, but it is the cartilage's task of distributing the load over a wider surface which is more important. In our model this has been taken care of by the use of gap elements and by assuming congruency between the femoral head and the acetabulum. Both measures will result in a local load transfer which should be a fair approximation of the real situation. Another limitation might be the use of membrane elements for the cortical shell. Membrane elements typically support only in-plane loading, which means that part of the structural response of the shell may be missing. However, in this particular case, where we are dealing with



Fig. 7. Peak values of the compressive normal stresses in the subchondral bone layer versus the respective value of the hip joint force. Even for low values of the hip joint force still considerable compressive stresses are found.



Fig. 8. Lateral view of the von Mises stress distribution during one-legged stance in the cortical shell if only the hip joint force is applied. The iliac bone remains largely unloaded, while loading of the pubic bone is exaggerated compared to Fig. 3 (2).

an extremely curved geometry, the overall deformation patterns happen to be not that much affected: benchtesting of the membrane elements versus thin bricks in a similarly curved (yet geometrically more simple) structure showed less than 10% change between the two. Furthermore, the external forces were taken from two different sources: the hip joint force data from Bergmann et al. (1990) and the muscle force data from Crowinshield and Brand (1981). This leaves open the possibility that these two loading regimes do not 'fit' together properly. However, the hip joint force is by far the most important one where it concerns the stress patterns around the acetabulum. We believe, therefore, that of the available choices, the best one is to work with a precise hip joint force and with muscle forces that may be somewhat off. Finally, the fact that only walking was considered restricts to some extent the scope of the results. However, walking is the most frequent of the more strenuous activities for the hip



Fig. 9. Compressive normal stress at the articular surface of the subchondral bone during one-legged stance for the basic model (top) and the model with the increased hip joint force (bottom); in the white areas, normal stresses are zero. Increasing the hip joint force does not change the basic stress distribution within the acetabulum.

joint and by considering various phases during the walking cycle, we believe that we covered quite a physiological range of loading situations. Rising from a chair might also have been interesting to study, but unfortunately for this activity no muscle force data were available from literature. Besides that, Bergmann *et al.* (1989) showed that the direction of the hip joint force during rising from a chair is not much different than during one-legged stance, while in magnitude it is less than half.

The pelvic bone is usually characterized as a socalled sandwich construction, in which the bulk of the load is carried by thin shells of a high-modulus material, while a low-weight core material acts as a spacer (Jacob *et al.*, 1976; Dalstra and Huiskes, 1990). In the present study, this phenomenon has been confirmed; the stress levels in the cortical shell were found to be about 50 times higher than in the underlying trabecular bone. For the load cases in the first half of the walking cycle (the stance phase), the average von Mises stress in the cortical shell lies between 15 and 20 MPa, while in the underlying trabecular bone, this value lies between 0.3 and 0.4 MPa. Goel and coworkers (1978) found values up to 40 MPa for the principal stresses (both tensile and compressive) in the acetabular region. These values are probably too high, because with the strength of cortical bone around 120 MPa (Carter et al., 1981), this would make the pelvic bone very vulnerable to fatigue failure. Stresses within the acetabulum, found by Oonishi and coworkers (1983) with their pelvic FE model, had a magnitude of around 0.01 MPa. Compared to the present results, this is improbably low, and therefore we fear an error has been made in their calculations or conversions. In their model, the highest stresses in the subchondral bone are found near the bottom of the acetabulum, while in our case the stress peaks occur near the edge of the acetabulum. When comparing the stresses calculated in the rest of the pelvic bone, Oonishi and co-workers also report high stresses in the ilium, above the superior edge of the acetabulum, extending to the incisura ischiadaca major region. The area of high stresses at the posterior part of the ilium due to the gluteus major muscle did not occur in their model.

Distributions of the strain rates indicate that during walking the highest gradients of the stresses occur in the pubic bone, the subchondral bone in the acetabulum and in the posterior part of the iliac bone. In general, strain rates were lower in the cortical shell than in the underlying trabecular bone. The magnitudes of these stain rates were found in the range of 10^{-3} - 10^{-1} s⁻¹. According to Carter and Hayes (1977) no significant hydraulic effect of the marrow will be found for strain rates lower than 10 s^{-1} . Linde et al. (1991) measured the strength and stiffness of trabecular bone for a wide range of strain rates $(10^{-4}-10 \text{ s}^{-1})$. Based on their findings, in the present range of strain rates increases up to 40% for the strength and 20% for the stiffness of the bone may be expected due to viscoelastic effects.

The hip joint force is the most important force for the load transfer across the pelvic bone. During normal walking, it remains directed towards a relatively small area in the anterior/superior quadrant of the acetabulum, according to the measurements of Bergmann et al. (1990). Its line of action does not intersect the line between the iliac and the pubic support areas and therefore, the hip joint force tends to tilt the acetabulum forward and upward. This is countered by the muscle forces acting on the iliac and the ischial bones and it is because of this muscle action that the pelvic bone is stress-relieved in the cases with full loading assumed compared to the cases with only the hip-joint force included (Figs 3 and 8). Due to the muscle forces, the stress distributions in the bone remain fairly constant during a walking cycle (Figs 3 and 4), even though the hip joint force varies considerably (from almost 200-2200 N). Only halfway through the swing phase the pelvic bone is clearly less stressed. So, apparently the muscle forces help to keep changes in the stress distribution to a minimum, which is supposedly favorable with regard to fatigue failure of the bone material.

The transfer of the hip force takes place predominantly in a narrow strip along the anterior/superior edge of the acetabulum. Depending on its precise direction, deeper areas in the acetabulum also transfer a part of the hip joint force. Because of this load transfer at the edge of the acetabulum, the lateral shell of the iliac cortex, just above the acetabulum and extending towards the incisura ischiadaca major region, is heavily stressed. To withstand these loads, the most dense trabecular bone and the thickest cortical shell are found in these areas (Dalstra et al., 1994). The high stresses in this region are also qualitatively confirmed by results from strain gage measurements by Finlay et al. (1986). The absolute values of the stresses which they reported were smaller, due to the lower value they assumed for the Young's modulus of cortical bone. Pauwels (1973) argued that for a normal configuration of the hip joint the stress distribution in the acetabulum is uniform. This, however, is based on the assumption that the acetabulum can transfer loads in all directions. Our results indicate however that loads are mainly transferred from the acetabulum through the lateral cortical shell to the sacro-iliac joint and the pubic symphysis. The actual stress distribution in the acetabulum is affected by this load-transfer mechanism, whereby the deeper parts of the acetabulum are stress-shielded. With their three-dimensional model, Koeneman et al. (1989) did find higher stresses in the deeper acetabulum, but this difference might be explained by the fact that they did include cartilage elements, but did not use the femoral head to introduce the load onto the acetabulum, so that the stress distribution at the articular surface of the acetabulum between their model and our are not the same. The high stresses at the superior acetabular wall demonstrate its importance in the natural load transfer mechanism of the hip joint. In dysplastic acetabuli, where this part of the wall is underdeveloped or 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 acetabuli. Therefore, a dysplastic acetabulum can definitely be considered as a considerable risk factor for wear of the hip joint.

The stress component which actually transfers the hip joint force onto the pelvic bone, is the normal or radially directed component of the contact stress between acetabulum and femoral head. Its highest value was found to have a magnitude of around 9 MPa and occurred during the one-legged stance phase. It is worth noting that even during the swing phase, when the hip joint force has dropped to less than 10% of its value during one legged stance, the compressive normal stress still has a peak value of nearly 2 MPa (Fig. 7). Due to the absence of cartilage, these stress values may be exaggerated and it is therefore interesting to compare these values to experimental results. Hodge and co-workers (1989) reported on pressures between prosthesis and acetabular cartilage in the acetabulum for walking, stair

climbing and rising from a chair, which had been measured in vivo by pressure transducers in the head of a telemetrically instrumented femoral prosthesis. During walking, they found peak values of 5.5 MPa shortly after the operation, but when gait had normalized after two or three years, these peaks reduced to 4 MPa. The superiorly directed stress peak during the stance phase of walking, found by Hodge and coworkers, corresponds well to our findings. Contact stresses within the acetabulum have also been measured by Brown and Shaw (1983) and they reported values of 8.8 MPa for local stress peaks in the region of the acetabular dome. This indicates that the magnitude of the contact stresses predicted by the present finite element model are indeed somewhat high, but still lie within a realistic range.

We may conclude that the pelvic bone behaves like a sandwich construction. The most important force for the pelvic bone, the hip joint force, is predominantly transferred along the superior edge of the acetabulum onto the rest of the pelvic bone towards the sacroiliac joint and the pubic symphysis. Transformation of the data by Bergmann and coworkers (1990) into an acetabular orientation, showed that the hip joint force remains pointed into the anterior/superior quadrant of the acetabulum during walking. When external loading only includes the hip joint force, very high stresses are found in the pubic bone. The muscles forces have a stabilizing effect on the pelvic load transfer and largely compensate for changes in the magnitude of the hip joint force. Because of this the stress distributions in the pelvic bone are not subject to large variations during a walking cycle. Within the acetabulum itself, changes are more substantial, as here the stress distributions are more directly dependent on the magnitude of the hip joint force.

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