SCIENTIFIC ARTICLE

Finite Element Modelling for Assessing Effect of Acetabular Component Orientation on the Basic Stress Path above Acetabular Dome

Yong Nie, PhD¹, Fu-xing Pei, MD¹, Zong-ming Li, PhD²

¹Department of Orthopaedics, West China Hospital, Sichuan University, Chengdu, China; and ²Department of Biomedical Engineering, Cleveland Clinic Learner Research Institute, Cleveland, Ohio, USA

Objective: To investigate the effect of acetabular component orientation on the basic stress path above the acetabular dome in the recommended safe zone.

Methods: A subject-specific normal hip finite element model was generated and a convergence study carried out to determine the number of material properties for trabecular bone using a normal hip model. Four abduction angles (35°, 40°, 45° and 50°) and four anteversion angles (10°, 15°, 20° and 25°) from the recommended safe zone of acetabular cup orientation were chosen to simulate acetabular reconstruction. The distribution and level of periacetabular stress was assessed using a normal hip model as a control and 16 reconstructed acetabula in simulated single-legged stances.

Results: The error of the average stress between plans four and five (50 and 100 materials for trabecular bone respectively) was 4.8%, which is less than the previously defined 5% error. The effect of acetabular component orientation on stress distribution in trabecular bone was not pronounced. When the acetabular component was at 15° anteversion and the abduction angle was 40° or 45°, the stress level on posterolateral cortical bone above the acetabular dome was as stable as that in the normal hip model.

Conclusions: Acetabular component orientation affects the basic stress path above the acetabular dome. Thus, orientation should be considered when attempting to restore normal biomechanics in the main load-bearing area.

Key words: Acetabular component orientation; Acetabular reconstruction; Finite element; Stress path

Introduction

Total hip arthroplasty (THA), which is performed to relieve arthritis pain or fix severe physical joint damage as a component of hip fracture treatment, generally involves replacing both the acetabulum and femoral head. Reconstruction of the acetabulum defines a new hip center, which in turn induces changes in hip joint biomechanics, leg length and femoral reconstruction^{1,2}. The mechanism by which the femur responds to the resultant altered stress state is better known

than that by which the pelvis responds to it, because there are few clinical data on the reaction of the acetabulum to a metal press-fit acetabular prosthesis^{3,4}. Several previous studies have shown that after press-fit acetabular reconstruction, some of the stress on trabecular bone is transferred to cortical bone because of stress shielding, resulting in a regional decrease in bone density, which increases the risk of implant migration and loosening⁵⁻⁷. However, long-term follow-up studies have indicated that stress transfer results in more significant bone

Address for correspondence Fu-xing Pei, MD, Department of Orthopaedics, West China Hospital, Sichuan University, 37 Guo Xue Xiang, Chengdu, China 610041 Tel: 0086-28-85422428; Fax: 0086-28-85422570; Email: peifuxing_cd@163.com; xiaodanyong@163.com

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density loss in trabecular bone than in the cortical bone above the acetabular dome. The average amount of bone loss is reportedly small and the implant stable^{8,9}, which suggests that stress transfer caused by stress shielding does not disturb the basic stress path above the acetabular dome. Thus, to ensure stability of the implant, the basic stress path above the acetabular dome should be the primary concern when performing press-fit acetabular reconstruction. Nevertheless, the basic stress path and the factors affecting it remain unclear.

After acetabular reconstruction, acetabular component orientation is a significant factor in pelvic osteolysis, wear, impingement and femoral head dislocation^{10,11}. Although post-operative dislocation rates have been reported to be as high as 2%–11%, the most recent data suggest a rate of less than 1%¹². Correct positioning of the acetabular cup has also been reported to be an important factor in preventing component wear after THA¹³. Lewinnek *et al.* reported that the safe zone for acetabular cup abduction is between 30° and 50°, and for cup anteversion between 5° and 25°¹⁴. Harrison *et al.* highlighted the need for a surgical reference system which can be used to describe the orientation of the acetabular cup intra-operatively¹⁵. Several studies have used a finite element (FE) model and simulated acetabular reconstruction to assess changes in stress distribution in the periacetabular region according to different orientations of the acetabular component^{16,17}. A recent study examined the effects of acetabular cup orientation on pelvic osseous loading and showed that increased cup abduction resulted in a 12% increase along the medial acetabular wall and an 18% decrease in inferior lateral regions in strain¹⁸. Although an experimentally validated FE pelvis model was employed in these studies, a normal hip model was not included as a valid control. In addition, the selected range of angles for placing the acetabular component was not in accordance with the actual situation in acetabular reconstruction.

In the present study, the effect of acetabular component orientation on the basic stress path above the acetabular dome in the recommended safe zone (abduction between 35° and 50° and cup anteversion between 10° and 25°) was analyzed. A subject-specific normal hip FE model and 16 reconstructed acetabula were generated. The number of material properties assigned to trabecular bone was defined through a convergence study. Stress changes in the region of interest (acetabular dome) of the 16 reconstructed acetabula were assessed by comparing the stress distribution of the normal hip model in terms of the principles from external to internal and qualitative to quantitative. We postulated that placement of the acetabular component at a certain angle would result in a stable stress level above the acetabular dome, which is close to the basic stress path.

Materials and Methods

Subject-specific FE Modelling

A male pelvis was scanned (acquisition matrix, 512 × 512; voltage, 120 kV; slice thickness, 0.6 mm) from superior to

Fig. 1 (A) Normal hip model. (B) One example of a reconstructed acetabulum.

inferior, using a Siemens Emotion CT scanner (Siemens Healthcare, Erlangen, Germany). Anatomical acetabular abduction and anteversion angles are 45° and 19° respectively according to published CT measurements¹⁹. A normal hemipelvis 3-D surface model was generated from Digital Imaging and Communications in Medicine (DICOM) data using MIMICS (version 10.01; Materialise NV, Leuven, Belgium). The surface model was then used to generate solid FE models in ANSYS (version 10.0, ANSYS, Canonsburg, PA, USA). A simplified normal hip model comprising a hip bone and a femur hemi-head is shown in Fig. 1A. Based on the normal hip model, the anatomical hip center was determined by experienced joint replacement surgeons with the assistance of SuperImage (version 1.2; Cybermed, Shanghai, China). With the hip center, four abduction angles (35°, 40°, 45° and 50°) and four anteversion angles (10°, 15°, 20° and 25°) from the recommended safe zone of acetabular cup orientation were chosen to simulate acetabular reconstruction in THA, creating 16 (4 × 4) simulations. The 16 simulations had 50 mm diameter reamed cavities in terms of abduction and anteversion angles. The acetabular component was simplified as a hemispherical cup and a hemi-head model (Fig. 1B).

A ten-node tetrahedral element was used for mesh generation and the optimal mesh densities were assessed against a mesh convergence study. In all models, the meshes comprised around 500,000 elements to assure accuracy of calculation^{17,20}. Cortical bone was assumed to be isotropic and homogeneous, with an elastic modulus (17,000 MPa), whereas trabecular bone material was simulated by the method of inhomogeneous material property assignment. In addition, the elastic moduli of the titanium cup and alumina ceramic head were 106,000 MPa and 404,000 MPa, respectively^{21,22}.

A hip joint contact force vector *F* that followed the direction of peak hip contact force during the single-legged stance of normal walking was applied through the rotation center *O* of the hemi-head model²³. Force components in the X, Y and Z

directions (calculated using a body weight of 65 kg) were 325 N, -195 N and 1462.5 N, respectively. Because the interfaces between head-cup and cup-bone were not the main focus of this study, contact behaviors were simplified to a linear problem using constraint equations. For geometrical constraints, all models were rigidly fixed at the sacroiliac joint and pubic symphysis (Fig. 1). This research was approved by the team's Institutional Review Boards.

Inhomogeneous Material Property Assignment and Convergence Study

Before assigning the inhomogeneous material property to trabecular bone, the boundary between cortical and trabecular bone was determined based on the reported relationship²⁰ and range of pelvis trabecular apparent density (from 0.109 to 0.959 g/cm³)²¹. The apparent density (g/cm³) for each tetrahedral element of the hemi-pelvis model was calculated from the Hounsfield unit (HU) values (ranging from -1024 to 3071) of the CT scans according to the following equation.

$$\rho_{\text{app}} = 0.02476 + 0.00128 \times HU \quad (1)$$

A density value of 0.959 g/cm³ was set as the boundary point between cortical and trabecular bone. The elastic modulus (GPa) for each element of trabecular bone was then calculated from the apparent density using the following empirical equation²¹.

$$E = 2.0173 \times (\rho_{\text{app}})^{2.46} \quad (2)$$

Using the normal hip FE model, a convergence study was performed to determine the number of material properties for trabecular bone. Five material assignment plans, including 1, 10, 25, 50 and 100 materials for trabecular bone, were applied in the normal hip model and solved sequentially under the same load and constraint conditions. Because the acetabular dome was the region of interest in this study, the same 10 nodes located in the trabecular bone in the region of the acetabular dome were uniformly for all five assignments. Convergence was achieved when errors of average von Mises stress of the 10 nodes between two sequential results were within 5%.

Stress Analysis

The basic stress path above the acetabular dome was defined as the stress distribution above the intact acetabular dome of the normal hip. According to the principle from qualitative to quantitative, von Mises stress on the nodes of the 16 FE simulations of reconstructed acetabula was compared with that of the normal hip FE model (control). After comparing the overall stress distribution, the maximal stress of each model was extracted. For normalization, the absolute differences between the stress value of the 16 FE simulations of reconstructed acetabula and that of the normal hip FE model were obtained. Variations in the normalized values for each angular position were analyzed. These variations were then used to measure the stability of these models under certain conditions, less variation indicating better model stability.

Fig. 2 10 cross-sections from inferior to superior with 5 mm increments. Sections 6 and 7 are the regions of interest.

In the internal stress analysis starting with the anatomical hip center, 10 cross-sections were acquired in each model from inferior to superior with 5 mm increments (Fig. 2). Stress distribution, both on cortical and trabecular bone, was compared. To assess the stress value on cortical bone quantitatively, 10 nodes within the posterolateral cortical region in two cross-sections above the acetabular dome^{8,9} were selected uniformly in each model, and the average stress of the 10 nodes calculated. Data normalization was performed to measure the effect of acetabular cup orientation on stress level above the acetabular dome.

Results

Convergence Study

A subject-specific FE model of the normal hip and acetabular reconstructions were generated based on an algorithm validated by previous experiments^{17,20}. Fig. 3 shows the results of the convergence study for defining the number of material properties of trabecular bone. Of all the plans, the stress of 10 nodes in plan one (one material for trabecular bone) was the most closely confined to a certain range. Variations in stress for each node and the average stress for 10 nodes decreased in parallel with increasing number of materials. The error of the average stress between plans four and five (50 and 100 materials for trabecular bone, respectively) was 4.8%, which is below the previously defined 5% error.

Periacetabular Stress Changes

The predicted von Mises stress distributions on the cortical bone in a single-legged stance of the normal hip model and 16

Fig. 3 Convergence study for defining the number of material properties assigned to trabecular bone.

simulations of reconstructed acetabula are shown in Fig. 4. In the normal hip model, high values were mainly located in three regions; namely, the acetabular rim, superior part of the lunate surface and the posterosuperior surface of the acetabulum. The stress in these three regions decreased in most of the 16 simulations of reconstructed acetabula. Furthermore, the stress in the region of the acetabular dome was low when the acetabular component was placed at six locations, four of them at 10° anteversion and two at 20° anteversion. The normalized maximum stress of the 16 simulations of reconstructed acetabula is listed in Table 1. The least stress occurred when the components were placed at 15° anteversion and 40° abduction. When the acetabular component was at 45° abduction, a relatively small range of stress was observed, except when there was 45° abduction and 10° anteversion.

Cross Sectional Stress Analysis

The predicted von Mises stress distributions on the 10 cross sections in a single-legged stance of the normal hip model are presented in Fig. 5. In the fourth and fifth sections, the stress was uniformly distributed on the cortical bone of the lunate surface. From the sixth to tenth sections, the highly stressed regions appeared on the posterolateral cortical bone of the acetabulum. The stress distributions on the trabecular bone tended to shift from the dome of the acetabulum to the posterosuperior region. In addition, the differences in stress distribution on trabecular bone between the normal hip and acetabular reconstructed cases decreased with increasing distance from the hip joint center (Fig. A1). The area of stress on the trabecular bone of each simulation of reconstructed acetabula was smaller than that of the normal hip model (Fig. 6). The normalized average stress values of the posterolateral cortical bone of the sixth and seventh sections (5 and

10 mm above the acetabular dome, respectively) in the 16 FE simulations of reconstructed acetabula are listed in Tables 2 and 3, respectively. The values closest to that of the normal hip model were for 15° anteversion.

Discussion

The study included models representing the normal hip and a variety of different potential implantation scenarios. Unlike the usual treatment of cortical bones as shell elements^{16,20,24–26}, both cortical and trabecular bone were meshed with ten-node tetrahedral elements based on a published study¹⁷. An empirical relationship between apparent density of pelvic trabecular bone and elastic modulus was obtained from a sample with an apparent density range of 0.109 g/cm³ to 0.959 g/cm³³², which is consistent with the apparent density range (0.1 g/cm³ to 1.0 g/cm³) of trabecular bone. In this study, a density value of 0.959 g/cm³ was considered as the boundary point between cortical and trabecular bones; this avoided extracting the boundary from the CT data via manual segmentation^{16,17,20}. Considering the inhomogeneous material property assignment for trabecular bone elements, the effect of the number of materials on stress value in the area of trabecular bone above the acetabulum dome was analyzed. The stress solution of plan one, in which trabecular bone was treated as homogeneous, differed greatly from that of the other four material assignments. Therefore, using one material to represent trabecular bone probably fails to predict the correct stress distribution. This finding is in good agreement with those of a previous study¹⁷. Considering that the error of the average stress between plans four (50 materials) and five (100 materials) was below 5%, 100 materials likely reasonably describes the property of trabecular bone. With regard to the application of load and constraints on the

Fig. 4 Predicted von Mises stress (MPa) distributions on cortical bone in a single-legged stance (body weight, 65 kg) of the normal hip and 16 simulations of reconstructed acetabula.

subject-specific models, although previous studies^{17,25} considered the effects of muscle and ligament forces in an FE model, little change in stress distribution was found at the dome of the acetabulum. Hence, loading forces were simplified as hip joint contact force in a single-legged stance.

Acetabular component orientation is an important factor affecting the risk of dislocation, impingement, osteolysis, acetabular migration, wear with polyethylene bearings and squeaking in ceramic-on-ceramic hips²⁷⁻³¹. Thus, understand-

TABLE 1 Normalized maximal stress (absolute difference between the stress value of the 16 FE simulations of reconstructed acetabula and that of the normal hip FE model) (MPa)

Abduction angles	Anteversion angles			
	10°	15°	20°	25°
35°	119.558	79.188	198.049	183.828
40°	118.539	2.584	220.918	199.747
45°	155.995	13.828	37.394	15.792
50°	121.967	143.685	139.01	201.463

ing the impact of acetabular component orientation on stress distribution in periacetabular regions is important for the long-term survival of primary THA implantation¹⁸. To analyze stress changes, a normal hip FE model was used as a control. From the predicted von Mises stress distributions on the cortical bone in the 16 simulations of reconstructed acetabula, the stress distribution above the acetabular dome varies considerably with different abduction or anteversion angles, which differs from the conclusions of a previous study¹⁸ in which component inclination generated minimal and significant changes in strain response in the acetabular dome. This apparent discrepancy is likely attributable to the limited abduction angles concerned in that study. The internal stress around the

Fig. 5 Predicted von Mises stress (MPa) distributions on the 10 cross-sections in a single-legged stance (body weight, 65 kg) in the normal hip model.

Fig. 6 Predicted von Mises stress (MPa) distributions on section 10 mm above the acetabular dome in a single-legged stance (body weight, 65 kg) of the normal hip and 16 simulations of reconstructed acetabula.

acetabular joint surface was used to assess stress distribution on trabecular bone. Because differences in stress distribution on trabecular bone between the normal hip and simulations of reconstructed acetabula decreased with increasing distance from the hip joint center, the sections near the acetabular dome were chosen to analyze stress distribution. Acetabular component orientation did not have obvious effects on stress distribution in the area of trabecular bone.

TABLE 2 Normalized average stress (absolute difference between the stress value of the 16 FE simulations of reconstructed acetabula and that of the normal hip FE model) (MPa) of posterosuperior cortical bone of the sixth section (5 mm above the acetabular dome)

Abduction angles	Anteversion angles			
	10°	15°	20°	25°
35°	16.278	6.748	5.3586	7.655
40°	8.473	0.080	12.030	13.233
45°	7.662	2.787	11.843	3.839
50°	3.570	6.363	4.508	5.543

Given that the stress distribution on cortical bone was more concentrated than that on trabecular bone, further quantitative assessment of stress level on cortical bone was considered both feasible and important. The stress level on posterolateral cortical bone in two sections above the acetabular dome was therefore assessed with different orientations of the acetabular component. Acetabular component orientation significantly affected the stress level on posterolateral cortical bone above the acetabular dome. When the acetabular component was at 15° anteversion and the abduction angle was 40° or 45°, the stress level on posterolateral cortical bone above the acetabular dome was as stable as that in the normal hip model. Combined with the qualitative findings concerning stress distribution in trabecular bone, this finding indicates that stress transfer to cortical bone occurs when the acetabular component is positioned at the correct angles and that the transfer probably originates from the acetabular cup rather than trabecular bone. The stress transfer is beneficial in that it decreases the effects of stress shielding and maintains the basic stress path above the acetabular dome. Therefore, stress shielding may not be the primary problem if the acetabular cup is suitably oriented. This finding supports our hypothesis and extends the findings of previous long-term clinical studies^{8,9}. In addition, when combined with the quantitative results of the normalized maximal stress of the 16 FE simulations of reconstructed acetabula, we found that the acetabular reconstruction model was stable

TABLE 3 Normalized average stress (absolute difference between the stress value of the 16 FE simulations of reconstructed acetabula and that of the normal hip FE model) (MPa) of posterosuperior cortical bone of the seventh section (10 mm above the acetabular dome)

Abduction angles	Anteversion angles			
	10°	15°	20°	25°
35°	14.836	11.771	8.603	7.664
40°	6.195	6.086	15.068	15.869
45°	5.097	1.181	10.353	11.043
50°	9.436	2.703	9.923	5.688

when the component was placed at 15° anteversion and at 40° to 45° abduction, which is in good agreement with reported clinical studies in which acetabular cup anteversion of 15° and abduction of 40° was recommended, and abduction angle of 40° to 45° considered optimal^{32–34}.

A subject-specific FE model of the normal hip and acetabular reconstructions were generated based on an algorithm validated by experiments^{17,20}. The following limitations of our study should be noted. First, all FE cases were generated from one normal hemi-pelvis; however, both the abduction and anteversion angles of the acetabulum are in the normal range¹⁶. Second, muscle forces and ligaments were not included in our FE models. Despite this, stresses in and around the acetabulum have been found to be similar in various models whatever muscle forces and ligaments were considered²⁵. Third, only the contact force during single legged stance was assessed. Future studies including more

loading information during a gait cycle would optimize the stress path. However, our analysis of stress distribution and level explains the mechanisms underlying stress transfer to cortical bone after acetabular reconstruction. Acetabular component orientation affects the basic stress path above the acetabular dome, which suggests the need to consider orientation when attempting to restore normal biomechanics in the region of the main load-bearing area. The results of comparison between the internal stresses of the 16 simulations of reconstructed acetabula with that of the normal hip model can guide clinical practice in the optimal orientation of the acetabular component. In our future research, the orientation of the acetabular component in the Lewinnek's safe zone with bone density above the acetabular dome will be studied in a series of patients after THA to validate its clinical utility in restoration of normal biomechanics in the main load-bearing area.

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Supporting Information

Additional Supporting Information may be found in the online version of this article at the publisher's web-site:

Fig. A1 CT data and stress distribution in 10 sections of all FE cases.