

3-D DYNAMIC HIP CONTACT PRESSURE DISTRIBUTION IN DAILY ACTIVITIES

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INTRODUCTION

Abnormal mechanical stress upon joint cartilage is one of the main causes of osteoarthritis. It is well recognized that intrinsic pathomechanical changes in articular cartilage depends upon local stress levels rather than upon global joint loading. Therefore, it is essential to estimate the contact pressure distribution and its peak value on the joint surface in order to consider the pathomechanical changes in joint cartilage. In another aspect, the treatments of osteoarthritis of hip joint vary from conservative rehabilitative exercises to surgical treatments such as osteotomies or total hip arthroplasties. Estimation of hip joint contact pressure during daily activities is useful for both preoperative planning and postoperative rehabilitation. In addition, knowing the distribution of the hip contact pressure is helpful in understanding the mechanics of the normal hip and understanding the pathology of the articular cartilage of the hip.

Computer simulation technique is a non invasive and ideal procedure to provide such information about particular condition of disease. FEM is one of the options to calculate contact pressure, however, it is always difficult and time consuming in analyzing a three dimensional model. Therefore this method is not suitable for acquiring information for treatment in individual case.

The purpose of this study is to use Rigid Body Spring Model technique that is basically a discrete finite element model, to analyze hip joint contact pressure distribution during a daily activity.

METHODS

By assuming that the acetabular surface is spherical in shape, a surface geometry model was created from an anteroposterior (A-P) radiograph of normal hip joint. If the quality of radiograph is normal and the hip joint of the subject is not in a progressive stage of osteoarthritis, the anterior and posterior acetabular edge can be clearly distinguished as shown in Figure 1. The A-P image of the hip joint can be obtained directly from the plane AP radiograph using an image scanner. The contour line of the femoral head was digitized within thirty points. The radius and center of the most proper circle were calculated by Simplex algorithms. The acetabular sourcil line was

digitized and the radius of the acetabular circle was calculated by least square method assuming that the center of the acetabular circle was the same as that of femoral head. Then the anterior and posterior edges of acetabulum are digitized on the plane radiograph. Since the acetabular surface was considered spherical, its radius, y and z coordinates of a point on the sphere were determined in A-P radiograph, the remaining x coordinate values were calculated using the equation for spheres: $x^2 + y^2 = r^2$. The sourcil line was used to determine the cartilage coverage along the acetabular surface and the acetabular fovea was not considered as joint surface. The radius of the articular surface was determined as the mean of the radius of femoral head and acetabulum. The surface of the acetabulum was divided into 6600 rectangular mesh elements. The joint contact was assumed to be congruent so that one common meshed surface could represent hip joint contact surface (Fig. 1).

In each rectangular mesh, one compressive spring in normal direction and one share spring in tangential direction were placed to model the joint cartilage (Fig. 1). The stiffness of compressive spring (kd) was determined from cartilage Young's modules and Poisson ratio, 5 MPa [3] and 0.3 respectively. The share springs' stiffness were set 0.001 N/mm to simulate very low frictional coefficient of cartilage. The stiffness of both compressive and share springs were assumed to be equal in whole joint surface. As deformations mainly occur inside the cartilage, the pelvis and the femoral head could be considered rigid bodies. Pelvis was fixed and femoral head was assumed movable in six degree of freedom. An equilibrium equation was formulated between whole spring forces and external force and moment.

If the load is applied through the centroid of the femoral head, springs are under compressive displacement, some may be under tensile displacement. As the cartilage can not bear tensile force, the spring elements carrying tension are eliminated from overall equilibrium equation and the new equilibrium equations are formulated again. The calculation is carried out in an iterative manner until all existing springs bear reasonable forces. The pressure distribution is obtained by calculating each spring force according to its deformation. Biomechanical application of this method has been

published in [1, 5, 6]. The changes of joint contact pressure through out the activities were visualized graphically (Fig.1).

The input data for the simulation, such as the magnitude, orientation of the resultant force through the hip joint, were taken from Bergman's data, which was based on the joint average force applied to an implanted prosthesis in the femoral head for four patients during daily activities (Gait '98 CD-ROM). The patients group consisted of 3 males and one female. The joint force data was transformed from Bergman's thigh coordinate system [2] to the acetabulum coordinate system in order to apply these forces to our model using 3×3 rotation matrix and translation vectors determined by Bergmann et al[2].

RESULTS

During walking, the maximum pressure was 22.4 MPa at midstance(Fig. 2). The high contact pressure in the acetabulum was in a medial-superior location. During climbing stairs the high contact pressure in the acetabulum was in a posterior-superior location. The maximum pressure was 25.4 MPa at a one-legged position, while the opposite leg was in a swing phase. When standing from a chair, the high pressure in the acetabulum was in a posterior-inferior location. The maximum pressure was 13.2 MPa when subjects were about to raise up from a chair and when the upper body was leaning forward. The maximum pressure was during climbing stairs in daily activities. In all three activities the contact areas were smallest when the contact pressure was highest.

DISCUSSION

In normal walking the maximum contact pressure was at mid-stance. This result was consistent with that of previous instrumented implant experiments of the femoral head [4]. In daily activities the contact pressure of the acetabulum during climbing stairs was highest. The peak contact pressures of getting up from a chair were lower than that of walking or climbing stairs. The interesting thing is that in activities of the high loading conditions, such as at mid-stance of walking and climbing stairs, the location of the peak contact pressure was distributed in the edge of the posterior or medial acetabulum. In activities of the low loading conditions the location of the peak contact pressure was distributed in interior of posterior or medial rather than in the edge of the posterior or medial acetabulum. During climbing stairs and standing from a chair, the distribution of the high contact pressure located in the posterior. These may result in hip disposition. Also, we concluded that in activities of the higher loading condition, contracted hip abductor muscles might pull the femoral head externally and the location of the hip contact pressure move in the edge of the acetabulum.

The model developed in this study can be used to preoperative planning of hip osteotomy surgery, assessment of femoral head and acetabulum fractures, and prediction of collapse of the femoral head and evaluation of treatment in avascular necrosis of the femoral head.

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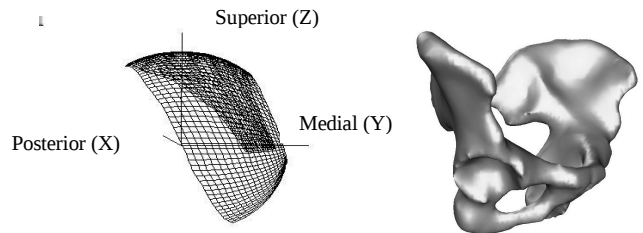


Fig. 1A mesh model

Fig. 1B 3D pelvic model

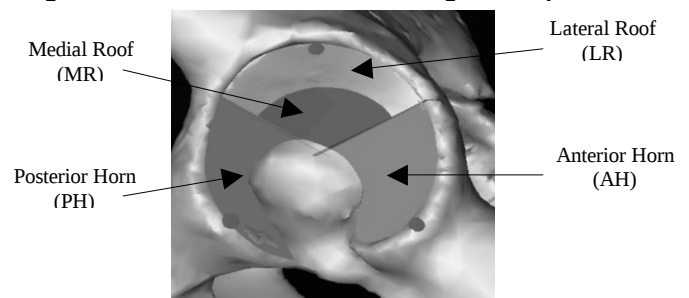


Fig. 1C Geometry of acetabular joint surface

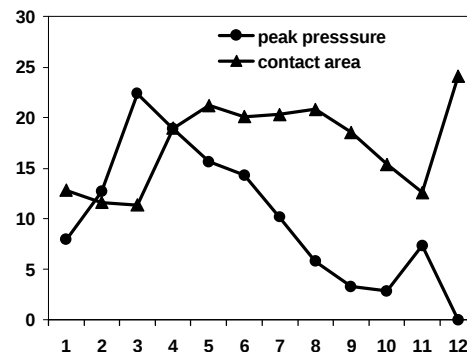


Fig. 2 This graph showing contact pressures during normal walking. The high contact pressure was at mid-stance.

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