Three-dimensional dynamic hip contact area and pressure distribution during activities of daily living

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Abstract

Estimation of the hip joint contact area and pressure distribution during activities of daily living is important in predicting joint degeneration mechanism, prosthetic implant wear, providing biomechanical rationales for preoperative planning and postoperative rehabilitation. These biomechanical data were estimated utilizing a generic hip model, the Discrete Element Analysis technique, and the in vivo hip joint contact force data. The three-dimensional joint potential contact area was obtained from the anteroposterior radiograph of a subject and the actual joint contact area and pressure distribution in eight activities of daily living were calculated. During fast, normal, and slow walking, the peak pressure of moderate magnitude was located at the lateral roof of the acetabulum during mid-stance. In standing up and sitting down, and during knee bending, the peak pressures were located at the edge of the posterior horn and the magnitude of the peak pressure during sitting down was 2.8 times that of normal walking. The peak pressure was found at the lateral roof in climbing up stairs which was higher than that in going down stairs. These results can be used to rationalize rehabilitation protocols, functional restrictions after complex acetabular reconstructions, and prosthetic component wear and fatigue test set up. The same model and analysis can provide further insight to soft tissue loading and pathology such as labral injury. When the pressure distribution on the acetabulum is inverted onto the femoral head, prediction of subchondral bone collapse associated with avascular necrosis can be achieved with improved accuracy.

Keywords: Hip contact pressure; Hip contact area; Discrete element analysis; Activities of daily living; Hip wear

1. Introduction

Abnormal mechanical stress on joint cartilage is one of the main causes of osteoarthritis. It is well recognized that pathologic changes in cartilage and bone depend on local stress levels rather than on global joint loading (Genda et al., 2001; Mavcic et al., 2002). Therefore, it is essential to estimate the contact pressure distribution and its peak value along a joint surface. Knowing the contact pressure distribution on the hip joint helps to understand the mechanics of the normal hip and pathology of the articular cartilage during abnormal loading. In addition, these estimates the contact pressure during activities of daily living provide functional knowledge as a basis for both preoperative planning and postoperative rehabilitation of the hip.

Experimental studies measuring contact pressure of the hip joint have been performed by many authors. Greenwald and O’Connor (1971) used a dye exclusion technique to estimate contact area. The pressure distribution was calculated from this given the known loading conditions. Rushfeldt et al. (1981) utilized an endoprosthesis instrumented with transducers to measure the contact pressure in the acetabulum of a cadaveric specimen. Brown and Shaw (1983) measured...
ex vivo femoral head and acetabulum contact pressures by using piezoresistive transducers in the femoral head cartilage. Hodge et al. (1986) implanted an instrumented femoral head in vivo and acquired data just after the installation and 1 year postoperatively, this study was continued by Krebs et al. (Givens-Heiss et al., 1992; Krebs et al., 1991; Strickland et al., 1992) over a 5-year period to report the effects of various rehabilitation exercises on hip joint contact pressure. In an in vitro study, Afoke et al. (1987) reported on the distribution of the hip contact pressure in the loaded femoral head using a pressure-sensitive film in five cadaver hips. This study demonstrated that peak pressures in the femoral head increased mediolaterally towards the cotyloid fossa in all of the specimens.

Though the aforementioned studies provide insight into hip joint function, estimating the progress of joint disease or conducting a preoperative plan based on individual patient evaluation remains a difficult problem to solve. In addition, these experimental studies are invasive by nature and costly due to the need for specialized implants with a finite lifespan. Handling tissue specimens also involves the inherent risk of disease transmission and the inability to model multiple procedures on a single specimen.

The computer simulation technique pursued in the present study provides information about joint pressure, which is a key element in the progression of joint disease. It has the advantage of being non-invasive, cost-effective, and repeatable. Additionally, individual case studies can be analyzed by modifying various patient-specific geometrical and mass parameters. In general, in order to obtain contact pressure using numerical simulation, inverse dynamic formulations and optimization techniques are necessary (Braune and Fischer, 1987; Chao and Rim, 1973). However, analyses based on these techniques are complex and typically limited to only two points of a given gait cycle. Finite element modeling (Kurtz et al., 1997; Udoia and Jin, 2003) is one of the options to calculate contact pressure, however, when dealing with contact in a three-dimensional model this approach can involve extensive non-linear terms that degrade the reliability of results. The level of computation time and complexities of these simulations are not suitable for acquiring information for treatment in individual cases in a timely manner.

Discrete element analysis (DEA) based on a rigid body spring model is an efficient numerical tool, which provides a near real-time computational output (Genda et al., 1995). The accuracy of the technique was confirmed with other methods, including FEM, in the quantification of the contact stress (Li et al., 1997; Sakamoto et al., 1996; Stone et al., 1996). Recently, instrumented femoral head prostheses have been used to investigate the magnitude, direction, and moment of the hip contact forces during activities of daily living (Bergmann et al., 1993, 2001). These direct measurements of the hip contact forces can provide the accurate loading conditions required by the DEA method. Thus, we utilized the existing dataset of in vivo dynamic joint forces to generate a hip joint model using DEA and analyzed the hip joint contact pressure distribution. The purpose of this study is to use the DEA technique to predict contact area and three-dimensional pressure distribution in the hip during different activities of daily living.

2. Methods

2.1. Development of three-dimensional contact surface

An original three-dimensional surface was generated based on the assumption that the contact surface between pelvis and femur is spherical and articulation between the femoral head and acetabulum is concentric. This assumption is appropriate for normal hips (Genda et al., 1995; Konishi and Mieno, 1993). A surface geometric model was developed from an antero-posterior (A–P) X-ray radiograph of a normal hip joint. The contour line of the femoral head was digitized using C++ software, Simplex algorithms were used to calculate the radius and center of the most proper circle of the femoral head. The acetabular sourcil line was digitized, and the radius of the acetabular circle was calculated using the method of least squares with the assumption that the center of the acetabulum was the same as that of femoral head. The sourcil line was used to determine the cartilage coverage and hence the contactable region of the acetabular surface. The fovea was not considered as joint surface. The joint contact was assumed to be congruent so that only one common mesh-surface would represent the hip joint contact surface. The radius of this articular mid-surface was defined as the mean of the radius of femoral head and acetabulum and was divided into approximately 4000 rectangular mesh elements. This varied, however, depending on the size of the joint surface.

2.2. Three-dimensional generic model for data visualization

The calculated contact area and contact pressure during activities of daily living were visualized by superimposing the image on a generic model of the hip joint using C++ programming based on Virtual Interactive Musculoskeletal System (VIMS), which is a graphic-based simulation software (Chao, 2003). The generic three-dimensional pelvic model was created with the Visible Human Project dataset (National Library of Medicine, Bethesda, MD) using the imaging software analyze (Version 5, Mayo Clinic, Rochester, MN). In
order to describe the location of the pressure result, we defined four sections at the acetabulum anatomic regions (Fig. 1). First, assuming that the acetabulum is spherical, we determined the center of acetabulum by finding the center of a matching sphere. Two lines were drawn to separate the acetabulum into three sections, anterior, middle, and posterior. The first straight line was drawn between the center of acetabulum and antero-inferior iliac spine (AIIS) to delineate the anterior horn (section B in Fig. 1) from the middle section. Then, the first line was flipped about the centerline, which splits the acetabulum surface equally, to create the second line that delineates the posterior horn (section D in Fig. 1) from the middle section. Furthermore, the middle section was divided into two equal-area sections; the lateral roof and the medial roof (sections A and C in Fig. 1, respectively) by drawing a line that ran along the edge of the acetabulum cup model. The pressure distribution was calculated over the contact surface mesh model described earlier. The result of the pressure distribution based on the individual model was superimposed onto the generic pelvic computer graphic model for the visualization purposes only. This provides a user-friendly interface to manipulate the viewpoint and animation of dynamic joint pressure result.

2.3. Acquisition of constraint force data

The input data for the DEA simulation, including the magnitude and orientation of the contact force through the hip joint, was adapted from the complete dataset of the published study conducted at the Free University of Berlin (Bergmann et al., 2001). The average joint force was recorded using an instrumented hip prosthesis from four patients during various activities of daily living. The joint force data were transformed from the native lab-based coordinate system to the hip coordinate system used in the DEA by a $3 \times 3$ rotation matrix and $3 \times 1$ translation vector. DEA calculations were performed for eight activities of daily living: fast walking, normal walking, slow walking, standing up, sitting down, knee bending, climbing up stairs, and walking down stairs. For the purpose of analysis, the average dataset of the subject KW (BW = 702 N) (Bergmann et al., 2001) were used in this study.

2.4. Calculation of contact area and pressure distribution

The meshed region of the acetabulum under normal conditions represents an area of potential contact. This surface is congruent with the actual contact region in the unloaded, undeformed cases. The true functional contact area varied according to the joint loading and was calculated during the analysis (Genda et al., 2001). Within each rectangular mesh element, one compressive spring in the normal direction and one shear spring in the tangential direction were placed to model the deformable joint cartilage between the femoral head and acetabulum. The stiffness of the compressive spring was determined from the mean cartilage Young’s modulus of 11.85 MPa (Kempson, 1980), Poisson’s ratio of 0.45 (Blankevoort et al., 1991) and articular cartilage thickness of 2.66 mm (Athanasiou et al., 1994). The mesh-element shear spring functions to simulate frictional forces. The shear stiffness in the current model was set 0.001 N/mm to represent the low frictional nature of cartilage. The stiffness of both compressive spring and shear spring was assumed to be uniform throughout the entire joint surface. Because consequential deformations occur mainly inside the cartilage, the pelvis and the femur were considered to be rigid bodies. In the context of the DEA, the pelvis was considered fixed and the femoral head was assumed movable in six degrees of freedom.

The external dynamic load was applied through the centroid of the femoral head. The pressure distribution was obtained by dividing each calculating spring force according to its deformation by the respective mesh-element area. Using DEA, it is possible to find not only the peak pressure values, but also the effective contact area as the sum of the active elements. Both peak distribution and contact area were used to decide mesh density in the contact surface. We tested convergence of these values by varying number of the total mesh elements from 100 to 9000. However, there was no obvious effect of the total mesh elements number on the results. Therefore, we decided to represent the possible contact area with approximately 4000 mesh units because it was not too computationally expensive and still allowed us to represent a smooth surface.
3. Animation

Using our VIMS-tools (Chao, 2003), we can animate the result for eight activities of daily living. The screen-captures of the animation at the peak pressure during different activity are included in the result section. We can provide the compact disc containing complete animation file as well as complete dataset of results and flow-chart of DEA.

4. Results

The maximum peak pressure \( P_{\text{max}} \), location of \( P_{\text{max}} \), phase of \( P_{\text{max}} \) occurrence in terms of % of activity cycle, and contact area at the \( P_{\text{max}} \) occurrence in terms of % of maximum possible contact area \( A_{\text{max}} \) are summarized in the Table 1 for each activity of daily living (Table 1).

4.1. Walking activities (normal walking, fast walking, and slow walking)

For the slow, normal, and fast walking, the reference frame of 0% activity cycle was defined at heel-contact of the leg and 100% activity cycle was defined at the instant just before the second heel-contact of the ipsilateral leg.

During normal walking, the bipodal peaks in contact pressure were observed with an absolute maximum of 3.26 MPa at the superior–posterior aspect of the lateral roof of the acetabulum at the beginning of the midstance (16.5% activity cycle) and the corresponding contact area was calculated as 76.3% \( A_{\text{max}} \) (Table 1 and Fig. 2(A)). The peak pressure started rising again towards the end of swing phase of normal walking just before the heel contact and the same phenomenon can be seen in the fast and slow walk scenarios (Fig. 3).

During fast walking, the maximum pressure was 3.28 MPa located at the superior–posterior aspect of the lateral roof at 12.5% activity cycle and the corresponding contact area was 78.7% \( A_{\text{max}} \) (Table 1 and Fig. 2(A)).

During slow walking, the maximum pressure was 2.87 MPa at the superior–posterior aspect of the lateral roof at 16% activity cycle and the corresponding contact area was 81.2% \( A_{\text{max}} \) (Table 1 and Fig. 2(A)).

4.2. Standing up from a chair, sitting down on a chair, and knee bending

During standing up from a chair, the contact pressure increased sharply until 34.5% activity cycle, of which the full cycle was from hip-off to standing. The maximum peak pressure, which occurred at the edge of the posterior horn of the acetabulum, was 8.97 MPa and the corresponding contact area was 19.7% \( A_{\text{max}} \) (Table 1 and Fig. 2(B)).

During sitting down on a chair, the maximum pressure, 9.36 MPa, was developed at the edge of the posterior horn around hip-contact at 50.5% activity cycle, of which the full cycle was from standing to hip contact, and the corresponding contact area was 17.6% \( A_{\text{max}} \) (Table 1 and Fig. 2(B)).

During knee bending (closed chain), the knee joint was straight at the 0% activity cycle, and the subject bent his knee and brought the knee back to straight at the 100% of the activity cycle. The maximum pressure was 3.65 MPa at 53.0% activity cycle at the edge of the posterior horn and the corresponding contact area was 51.6% \( A_{\text{max}} \) (Table 1 and Fig. 2(B)).

4.3. Climbing up stairs and walking down stairs

During climbing up stairs, the definition of activity cycle is the same as of the walking activity, from heel-contact to heel-contact of the ipsilateral. The maximum contact pressure of 5.71 MPa occurred in the superior–posterior aspect of the lateral roof at 11.0% activity cycle and the corresponding contact area was 52.1% \( A_{\text{max}} \) (Table 1 and Fig. 2(C)).

During walking down stairs, the toe-off phase was defined as 0% activity cycle and the time just before toe-off was defined as 100% activity cycle. The maximum contact pressure of 3.77 MPa was found in the superior–posterior aspect of the lateral roof of the acetabulum at 55.0% activity cycle and the contact area was 80.6% \( A_{\text{max}} \) (Table 1 and Fig. 2(C)).

5. Discussion

Maximum joint contact forces and peak contact pressures corresponded to the beginning of the mid-stance of gait. At this point in the gait cycle, the entire body weight is supported by only one leg.
Fig. 2. (A) Maximum peak pressure during activities of walking. (B) Maximum peak pressure during standing up, sitting down, and knee bending. (C) Maximum peak pressure during climbing up stairs and walking down stairs.
Support of the body weight itself is the most important factor during activities of daily living as these activities are carried out with neither external loads applied to the body, nor with any external support. When sitting down the joint contact forces were lower than those of the other three activities, yet the highest pressure was calculated. Consequently, we think that the other important factor is the relative positions of the bones and the direction of the force applied (Fig. 2). The contact pressures were maximal when the coronal plane of the torso was perpendicular to the long axis of the femur (as in sitting down). Therefore, when body weight plays less of a role, muscle contraction and the soft tissue constraints on positioning imposed by the capsuloligamentous structures play an important role in determining the location and magnitude of joint pressure (Fig. 4).

During activities of walking, the contact area involved a large portion of the acetabular surface (about 80% of all potential area) even when the maximum peak pressure was observed. The peak pressures of walking were lower than those of the other groups of activities. Of the walking activities, slow walking produced the lowest peak pressure. This was probably due to the lower demand for stabilizing muscle co-contraction across the hip joint. At these lower speeds, less inertia is generated in the lower extremity segment and less hip muscle contraction is then needed to offset the decreased inertia. Therefore, slow walking is preferable to fast walking in order to minimize the peak pressures generated in the acetabulum. This data can be used to guide postoperative physical therapy regimens that are designed to protect an area susceptible to high-pressure injury. An area of marginal fixation during a complex acetabular reconstruction or a soft tissue labral repair or debride-

With respect to stair travel, the peak pressure while ascending stairs (5.71 MPa at 11.0% activity cycle) was larger than that of descending stairs (3.77 MPa at 55.0% activity cycle). In contrast, the contact forces when the peak pressures were observed were about 250% body weight for climbing up stairs and about 310% body weight for descending stairs (Bergmann et al., 2001). Climbing up stairs generated higher pressures due to the fact that the contact area was smaller than of descending stairs even with the lower contact force. This would suggest that ascending stairs should be avoided to minimize stress concentration in the acetabulum, in favor of descending stairs. The slight change in posture while traveling up and down stairs may play a critical role in the relative positions of the pelvis and femoral
head leading to the higher pressures observed while going up stairs (Fig. 5).

In the pathologic state, such as osteoarthritis, the asymmetry of cartilage degeneration results in greater contact stresses (Mavcic et al., 2002) probably due to decreased contact area and concentrated loading. In this study, the highest pressure appeared at the edge of the posterior horn of the acetabulum during sitting down due to the small area of contact. It is considered that this highly loaded activity reflects the functional difficulties associated with pain in patients of degenerative hip disease. The contact area is one of the most important factors to affect the contact pressure. Therefore, the magnitude and direction of joint constraint forces and the size of the actual contact area in the acetabulum significantly affect the pressure distribution. Postoperative physical therapy regimens discourage patients from bending at the waist, sitting in chairs that are too low, and climbing stair (Clinical Reference Systems Advisor Series, 2003). Avoidance of these activities and movements is intended to minimize loading of any posterior structures excised, or incised and repaired from abnormal stresses, allowing them the opportunity to heal.

During walking, the results showed maximum contact pressure at the beginning of the mid-stance and were comparable to the findings of previous studies (Brand et al., 1994; Hodge et al., 1986; von Eisenhart et al., 1999). The location of the peak pressure when the maximum contact area occurred was found at the anterior–superior aspect of the lateral roof where the cartilage thickness is greatest (von Eisenhart et al., 1999). Knowing that the contact area increases to bear the higher load, this finding could indicate that the thickness of cartilage is directly proportional to the force applied in the hip joint. It is legitimate to believe that walking, which occurs so frequently during daily life, is best optimized anatomically. The coincidence of the peak pressure in relation to the location of the thickest cartilage supports the idea that the distribution of cartilage thickness is related to the long-term frequent loading of the joint (Oberlander, 1977).

During sitting down, standing up, and knee bending, the peak pressures were centered along the edge of the posterior horn when the individual was just beyond the threshold of contact with the seat. The rationale for this finding comes from analysis of the function of the hip abductor muscles, which would be contracted, pulling the femoral head toward the periphery of the joint, thereby moving the femoral head towards the edge of the acetabulum. Thickening of the articular cartilage at the edge of the posterior horn, as shown in previous studies (von Eisenhart et al., 1999) is consistent with our observation that contact pressures are highest in this region during particular activities.

Appearance of peak contact pressure is a stronger function of contact area as opposed to maximums in joint contact force. The direction of joint constraint forces and the actual contact area in the acetabulum significantly affect the pressure distribution. According to Hodge et al. (1989), the maximum contact pressures were 4.0 MPa for walking, 4.5 MPa for stair climbing, and 9.7 MPa for rising from a chair in 36 months post operation. Whereas, the maximum contact pressures that we found were 3.3 MPa for normal walking, 5.7 MPa for climbing up stairs, and 9.0 MPa for rising from a chair. Hodge also found that the highest pressures were localized to the superior and posterior region of the acetabulum, which agrees with our findings well. Specifically we found that the maximum contact pressures were located at superior and posterior region of the acetabulum for normal walking and stair climbing, and at the posterior horn when rising from a chair.

Though our model did not include the soft tissues (muscles, tendons, and ligaments) per se, the aggregate effect of these tissues is accounted for by the use of in vivo endoprosthetic loading conditions. The ligament tension and dynamic muscle forces are independent of the joint contact force used here.

There were several assumptions made to determine the joint surface geometry and material properties (Genda et al., 2001). In order to create the three-dimensional model from a two-dimensional radiograph and have only a single contact surface, the hip joint surface was assumed spherical and congruent. Both the thickness and material properties of cartilage were assumed to be uniform over the contact area to reduce computational time. Using DEA implies that deformations can occur only within the cartilage of the articular surfaces, in reality however the bones of the hip would come into complete contact under high loading conditions (Greenwald and O’Connor, 1971). Because of this analysis was based on nominal forces the DEA technique is a prudent choice. Though this study uses principles of static analysis, the loading conditions used were taken from dynamic situations and calculations were performed at every 0.5% activity cycle, and for these reasons our quasi-static analysis shows realistic results.

In this study, we used a generic DEA model representing the assumed spherical geometry of the femoral head with linear material behavior for the cartilage. If a patient has more contact area in the posterior aspect of the acetabulum, the peak pressure could be reduced even under the demanding activity of sitting and standing from a chair. The potential contact area of the acetabulum should be developed according to the true anatomy of the individual patient for more reliable peak contact pressure prediction. The actual contact mechanics is substantially different from the
assumed method using the DEA and this is one of the limitations. By comparing the results with previous in vivo studies reporting peak pressures (Hodge et al., 1986; von Eisenhart et al., 1999), our DEA model was found to be sufficiently accurate in qualitative analysis. Our model and analysis are also useful in comparing the location and magnitude of peak pressure during ADL on a relative basis. Therefore, our method can be useful for preoperative planning for hip osteotomy. In addition, the data generated from our model could be utilized to design a bearing surface that demonstrated regional differences in the wear characteristics imparting optimal mechanical design and surface material selection. This simulation results could be inverted onto the femoral head during to estimate the subchondral bone stress distribution which would be useful to investigate the treatment options in patients with avascular necrosis in the femoral head.

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Appendix A. Supplementary materials

Supplementary data associated with this article can be found in the online version at doi:10.1016/j.jbiomech.2005.06.026

References


