

# Comparison of predicted and measured contact pressures in normal and dysplastic hips

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## ABSTRACT

*Hip dysplasia, a congenital and developmental deformity characterized by malorientation and a reduction of contact area between the femur and acetabulum, is the most common cause of osteoarthritis of the hip. According to current estimates, dysplasia accounts for nearly 76% of all cases of osteoarthritis, and many who are affected require a total hip replacement before the age of 50. It is theorized that in the poorly oriented and deformed pelvis, a reduction in contact area leads to an increase in contact pressure during normal activities. Currently, clinicians attempt to reposition the joint, assuming that improving the position of the existing contact surface will lead to decreased pressures. It is also assumed that improving certain geometric parameters correlates indirectly with decreased contact pressures. Neither these simple estimates nor other non-invasive models have ever been shown to be related to contact pressure. The purpose of this study was to evaluate a computerized method of predicting hip joint contact pressures, which applies known hip joint reaction forces to the three-dimensional surface of the hip joint. To this end, cadaveric and plastic pelvic models were developed to test whether the computer model could predict the magnitude and location of maximum pressure. Mechanical testing revealed that the computer model could be used to predict pressure in cadaveric pelvis at prescribed locations ( $r^2=0.64$ ). The computerized model could also be used to predict the magnitude and location of maximum pressure in a series of plastic models where the load vector and the degree of dysplasia were parametrically varied ( $r^2=0.7$ ). These findings suggest that the computer model may be useful in identifying patients who will fail osteotomy or whether they can be used to select the best osteotomy for each patient. © 1997 Elsevier Science Ltd for IPEM.*

**Keywords:** Hip, dysplasia, pressure, geometry

Med. Eng. Phys., 1997, Vol. 19, 180–186, March

## 1. INTRODUCTION

The hip joint is one of the largest weight bearing surfaces in the human body, and bears as much as 5.5 times body weight during jogging and stumbling<sup>5,15</sup>. Hip joint osteoarthritis is most frequently due to abnormal development, or dysplasia of the joint. As many as 76% of such cases of osteoarthritis can be attributed to untreated hip dysplasia, and affected patients often require a total hip replacement before the age of 50<sup>1,6</sup>. It is believed that dysplasia leads to increased joint contact pressures that result in degenerative changes and eventually osteoarthritis<sup>11,25,36</sup>. Since 120,000 total hip replacements are performed annually in the United States alone<sup>34</sup>, early detection and surgical correction of dysplasia can potentially prevent enormous subsequent morbidity and health-related costs. In young adults, the pelvic oste-

otomy is often used to reorient the acetabulum, which is believed to decrease cartilage pressure and stabilize the joint. The goal of the osteotomy is to correct the deformity before degenerative changes have occurred, thereby delaying or eliminating the need for a total hip replacement.

Currently, the clinical diagnosis of dysplasia is based on simple geometric descriptors (such as the lateral center-edge angle) that are determined from standard radiographs<sup>18,26,27,33</sup>. The lateral center-edge angle is an angle formed between a vertical line and a line from the center of the femoral head to the lateral rim of the acetabulum<sup>39</sup>. These simple geometric parameters measured from plane radiographs are thought to characterize the underdevelopment and malorientation of the joint. Although the material properties of cartilage influence the pressure distribution<sup>14,25</sup>, geometry is assumed to be the most important contributing variable.

While center-edge angles and other plane radiographic parameters are commonly calcu-

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lated, there are several difficulties and potential limitations associated with their use; these parameters: (1) do not account for the shape of articular surface; (2) have not been shown to predict contact pressure; and (3) may not adequately account for contact pressures that occur during all activities of daily living<sup>13</sup>. Furthermore, since most plane radiographic parameters are aimed at indirectly estimating contact pressure, a more direct method of predicting joint contact pressure may prove to be more useful for diagnosis of dysplasia and for planning corrective surgery.

Although several instrumented hip prostheses have been developed<sup>5,12,20,31</sup>, there are no non-invasive models that have been demonstrated to predict contact pressures. Since acetabular dysplasia often presents with vague clinical symptoms, near normal radiographic findings, and subtle physical signs, a sensitive and specific model would be able to aid surgeons in determining which patients would benefit from an osteotomy. Furthermore, because surgeons have shown that improving the orientation of the joint leads to excellent short-term results<sup>11</sup>, the question has largely become what surgical correction will optimize the long-term outcome. The aim of the current study was to determine whether a non-invasive, geometrical model of the pelvis could predict experimentally measured pressures in both plastic models of human pelvis and in cadaveric pelvis.

## 2. MATERIALS AND METHODS

### 2.1. Cadaveric pelvis

Three fresh frozen cadaveric pelvis with intact labra were obtained from 57, 63 and 84 year-old donors. Mild acetabular cartilage degeneration was observed in the 63 year old-pelvis, but the others appeared normal. Measurements were not made in the area of mild cartilage degeneration. The cadaveric pelvis were mounted using polymethylmethacrylate (PMMA) in the center of a hemispherical bowl in a coordinate reference system defined by the pubic symphysis and anterior iliac spines<sup>10</sup>. Medial, lateral, posterior and anterior anatomical directions were marked on the PMMA molds. A radio-opaque marker was embedded in the PMMA to define the origin of the coordinate reference system.

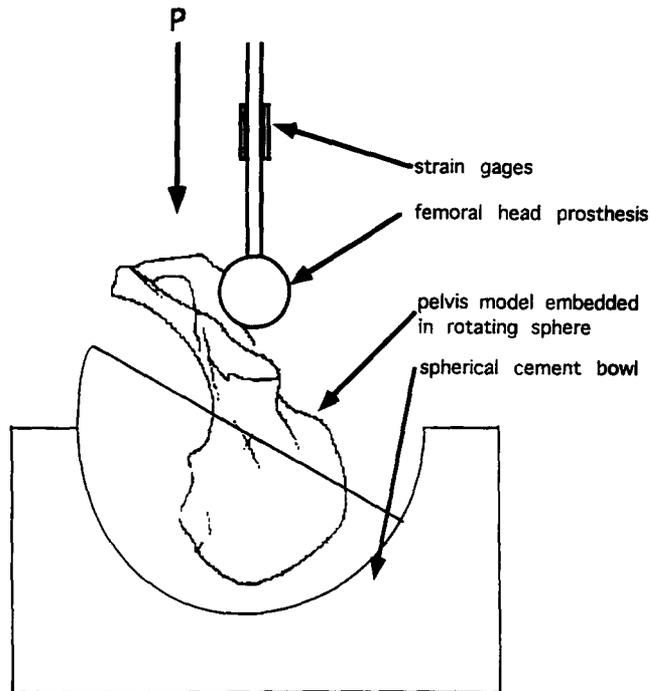
The matching femur for each pelvis was mounted to the crosshead of an Instron servo-hydraulic mechanical testing system (Instron Inc., Canton, MA, USA) with the femoral neck oriented vertically. The potted pelvis was placed in the spherical cement bowl in qualitatively realistic orientations under the constraint that measurements could not be made in areas of cartilage degeneration. A total of nine pressure measurements were made using low-range (2.5–10.0 MPa) Fuji pressure sensitive film (Tokyo, Japan). These nine measurements were scattered within three different pelvis. Each test was performed in compression with a load between 800 and 1200 N. Loads were selected to remain within the working range of the Fuji film and were applied with a

5 s ramp, 5 s hold at the maximum force (800–1200 N), and a 5 s release. The cement bowl was supported by a plate that was free to move in the horizontal plane, thereby eliminating any parasitic forces. The orientation of the potted pelvis, the pressure sensitive film, and the load vector location were determined using a 3D digitizer (Immersion Inc, Santa Clara, CA, USA). Both the pelvis and femur were kept moist during mechanical testing.

### 2.2. Plastic pelvis

Since cadaveric dysplastic joints are difficult to obtain and cannot be used to investigate parametric changes in load vectors and in geometry, a physical model was developed. The physical model was constructed to represent the relative material and geometric characteristics of the hip. A soft RTV (room-temperature vulcanizing rubber), Sawbone pelvis (Pacific Research Inc., Vashon, WA, USA), and femoral prosthesis were chosen to represent cartilage, pelvic bone and the femoral head, respectively. The RTV (HS2 colored catalyst, Dow Corning Co., Midland, MI, USA) was molded between a 47 mm head and a Sawbone acetabulum with an internal diameter of 51 mm. The molding process achieved an effective cartilage layer thickness of 2 mm, which was consistent with the measurements of both Modest *et al.*<sup>24</sup> and Athanasiou *et al.*<sup>4</sup>. Five cylindrical plug samples (each tested twice) had an average compressive modulus value of 1.1 MPa with a coefficient of variation of 0.101. Five identical pelvis with a 51 mm diameter acetabulum were provided by Pacific Research Laboratories, Inc. (Vashon, WA) The modulus value of the pelvic material was reported to be 552 MPa. One pelvis was tested intact, and various degrees of dysplasia in three other pelvis were simulated by removing sectors from the acetabular rim. A 47 mm femoral prosthesis (Osteonics Allendale, NJ, USA) was used to represent the femoral head.

Pelvis were mounted in a hemi-spherical bowl with a rigid resin (DER 324 Dow Corning, Inc., Midland, MI, USA) (*Figure 1*). The position of the mounted pelvis could be adjusted within the concentric spherical bowl to represent activities of daily living (midstance, stair ascent and external rotation). These load cases were adopted from Patriarcho who applied linear optimization methods to the muscular forces in the hip joint<sup>9,28</sup>. The orientation of the pelvis required to represent each load case was determined using the 3-D digitizer. These orientations were developed by defining the frontal plane as the one which contained the anterior superior iliac spines and the pubic symphysis<sup>10</sup>. The femoral prosthesis was attached to the crosshead of an Instron mechanical testing machine. A well-lubricated, simulated cartilage layer was placed on the acetabulum, and pressure measurements were made with super-low range (0.5–2.5 MPa) Fuji film. Strain gages were placed on the flange connected to the prosthesis to insure that it was centered within the acetabulum and that there were no off-axis loads.



**Figure 1** Diagram of testing apparatus with the pelvis mounted in a hemispherical jig, with the acetabulum at its center, so that the pelvis could be rotated to reproduce the hip joint reaction force vectors (P)

Surface pressure measurements were made by using a 5 s ramp to 250 N, holding 250 N for 5 s, and restoring to zero over 5 s. The force was less than applied to the cadaveric pelvis due to the reduced material strength of the physical model. A blunt mark corresponding to a location in the acetabular notch was made on the Fuji film so that the location of maximum pressure could be determined. The normal and dysplastic models were all mechanically tested in external rotation, midstance and stair ascent. Three tests were performed for each loading scenario on each pelvis, and the average maximum pressure for these tests was reported. These measurements were then compared to the maximum pressure predicted by the computer model.

### 2.3. Pressures from Fuji film

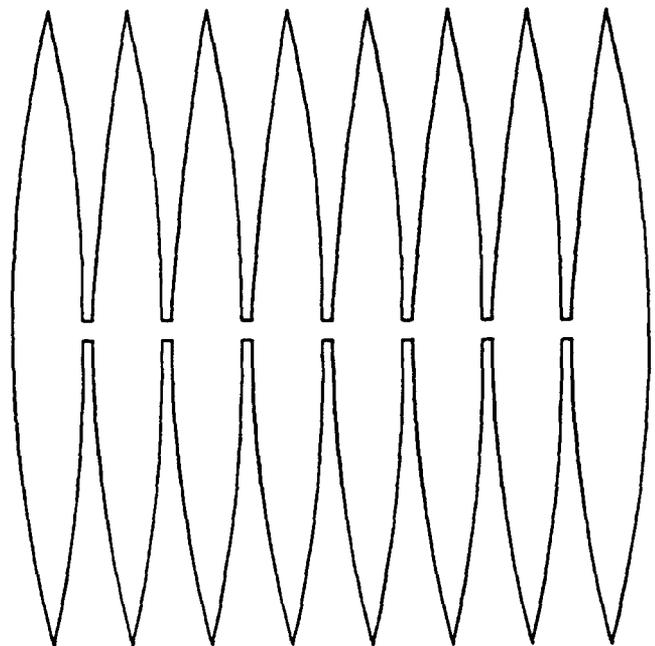
AVS (Waltham, MA, USA) image processing software was used along with calibration films to calculate contact areas and pressures. Eight calibration pressure films were made using known forces on a known circular area in accordance with the methods of Liggins *et al.*<sup>21</sup>. Calibration films were made for both the cadaveric and plastic models to account for their different film ranges. All tests were conducted at 23°C and 61% relative humidity. These films were digitized into 256 grey levels using a HP IICX flatbed scanner (Hewlett Packard, Palo Alto, CA, USA). The average grey level of each calibration chamber was fit to the known pressure using a cubic polynomial. The correlation coefficient for the cubic polynomial was greater than 0.97 in all cases. This function was encoded within a custom image-processing algorithm that calculates pressure from film grey

levels. To reduce artifacts, the pressure map was smoothed with a Gaussian filter.

An algorithm, similar to that used by Afoke *et al.*<sup>3</sup>, was developed for the plastic pelvis to allow the planar film to conform to the spherical geometry of the femoral head (*Figure 2*). The template was designed so that when the tips are closed together, the film fits exactly over the spherical surface. The shape was defined by the length of the major and minor axes of an ellipsoid. The major axis was designed to cover half the circumference of the femoral head. The length of the minor axis was found by subdividing the surface into eight ellipsoids. Struts connected the ellipses to reduce artifacts from overlapping edges of the film. The film was cut using precision laser technology (Laser Machining, Somerset, WI, USA). The computed pressures from the film were remapped to a sphere for clear visual display. Pressure measurements in the cadaveric pelvis were made with 5 mm diameter circular dots of low-range Fuji-PreScale pressure sensitive film sealed in thin tape.

### 2.4. Computer-Based method

Contact pressures were calculated in the cadaver hip joints and plastic replicas of hip joints from computed tomography (CT) examinations. Continuous 3 mm thick slices were obtained through the acetabulum. These data were used to create three-dimensional isosurface reconstructions of the pelvis (AVS 5.0, Advanced Visualization Systems, Waltham, MA, USA). Points along the acetabular rim and along the acetabular notch were digitized from the three-dimensional reconstructions. A sphere was fitted to these data points, and the portion of the sphere that represents the acetabular surface was determined. This surface was



**Figure 2** Pressure sensitive films for the plastic model were cut and folded over the femoral prosthesis to conform to its spherical geometry

discretized into uniform  $0.5 \text{ mm}^2$  patches. At each surface patch, the dot product between the applied load vector and a vector normal to the surface was calculated. The applied load was then divided among the patches based on the dot-products so that the sum of the individual vectors equalled the applied load vector. The pressure at each point was calculated by dividing the force magnitude by the cross-sectional area. Contact pressures were therefore based only on the idealized geometry of the acetabular surface. Material properties are not represented by our model. In addition, this model only represented normal forces across the joint and neglected shear forces.

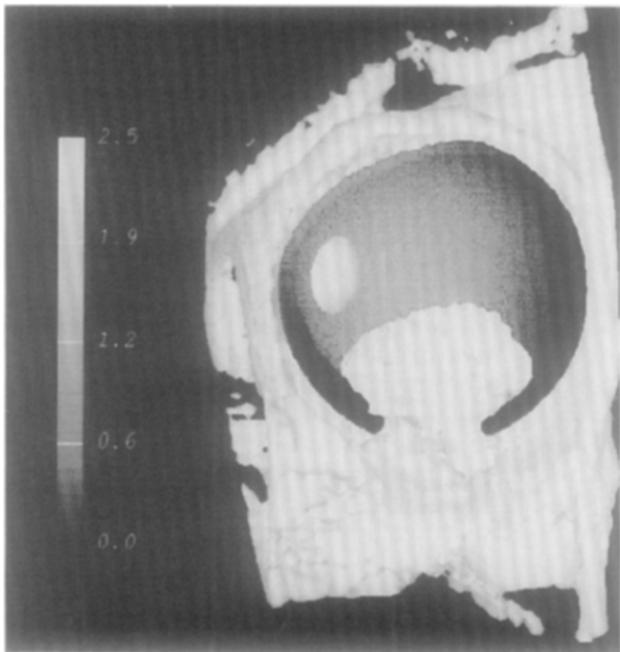
In the cadaveric model, pelvic orientation, the load vector direction, and the location of the pressure sensitive film were reconstructed from the CT exams based on the location of the radio-opaque marker and data obtained from the 3D digitizer. Fuji film pressure measurements were compared to the pressures predicted by the computer model at the specified locations under the given resultant loads (Figure 3).

### 2.5. Statistical analysis

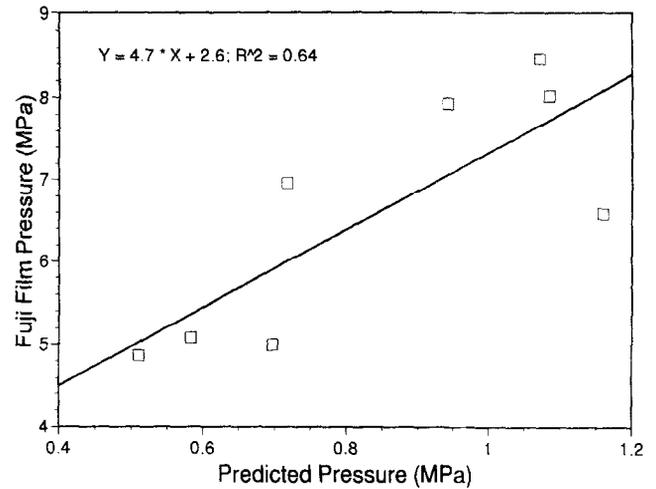
Linear regressions were performed comparing the computed and measured pressures. These analyses were designed to test whether the computer model could be used to predict the measured pressure in both the cadaver and plastic models. The location of maximum pressure for the plastic pelvis was recorded as occurring within one of four zones; antero-lateral, antero-medial, postero-lateral or postero-medial.

## 3. RESULTS

A linear correlation ( $r^2=0.64$ ) was found between the computed and measured pressures in the



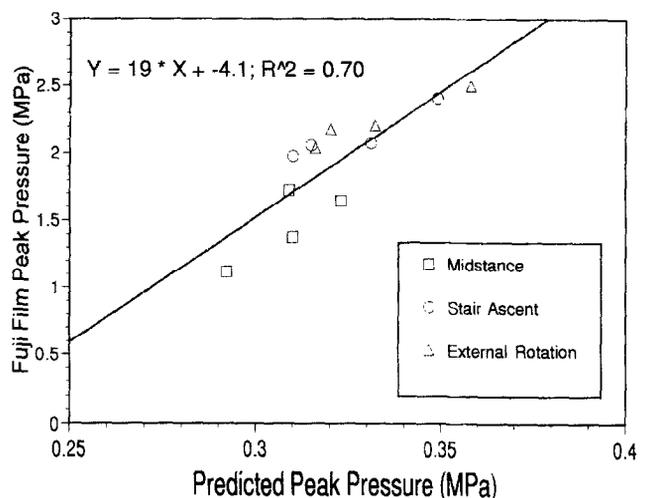
**Figure 3** CT reconstruction of cadaveric pelvis with a greyscale pressure map (in MPa) on the surface and a white dot representing the location of an example pressure measurement



**Figure 4** Comparison of measured versus computed contact pressure in the cadaveric pelvis at prescribed locations. The computed pressures predicted the measured pressures with a correlation coefficient  $r^2=0.64$

cadaver pelvis (Figure 4). A linear correlation ( $r^2=0.70$ ) was also found between the computed and measured pressures in the plastic pelvis (Figure 5). The slope of the regression for the plastic model was four times greater than the slope of the regression for the cadaveric model. The magnitude of pressures predicted in cadaveric and plastic pelvis by the computer model ranged from 0.29 to 1.2 MPa. These values were roughly seven times less than those measured with the Fuji film.

The location of maximum pressure under realistic load vectors in the plastic model for all loading sequences was in the anterolateral rim zone. The exact location varied somewhat depending on the loading scenario, but always remained within the anterolateral quadrant. Furthermore, for nearly all load cases, the pressure measured in the normal hip was lower than in the dysplastic hip. The coefficient of variation for repeated tests in the plastic and cadaveric pelvis



**Figure 5** Comparison of measured versus computed peak contact pressures in a plastic model of normal hips and in hips with dysplasia simulated by resecting sectors of the lateral, anterior or posterior rims. Tests were performed in loading scenarios which simulated midstance, stair ascent and external rotation. The computed pressures predicted the measured pressures with a correlation coefficient  $r^2=0.73$

was 0.06 and 0.11, respectively. Data for the cadaver and plastic pelvis were not pooled since the loads applied to the cadaveric specimens were roughly three times greater so that the combined data were not normally distributed.

#### 4. DISCUSSION

The aim of this study was to determine whether a CT-based model of acetabular geometry could be used to predict experimentally measured joint contact pressure in both plastic and cadaver models. The computer model can be used to predict the relative magnitude of pressure in cadaveric acetabula at prescribed locations. The computer model can also predict the relative magnitude and location of maximum pressure for *in-vivo* loading scenarios in plastic models of both normal and dysplastic hips. Because the two models represented different materials (femoral head and cadaver vs plastic foams and prosthesis) and measured different quantities (maximum pressure vs pressure at a prescribed location), the slopes corresponding to the two models were expected to be different.

Among the strengths of this investigation were that both the full three-dimensional geometry of the hip and its complex loading scenario were represented, parameters that are critical to the evaluation of coverage. Our data support the long-held assumption that dysplastic hip joints experience higher peak contact pressures than normal hip joints. These data also support the notion that acetabular geometry is an important variable in determining contact pressure in the hip joint.

This study has not directly proven that the computer model can be used to predict pressures in a dysplastic hip joint. Cadaveric dysplastic hips were unavailable, and furthermore cannot be used to investigate parametric acetabular rim deficiencies and varying *in-vivo* load vectors. Instead, we have shown that our computer model can be used to predict the relative magnitude of pressure in a normal hip and the location and relative magnitude of maximum pressure in a simulated normal and dysplastic physical model. Although the physical model does not consider the complex behavior of cartilage<sup>25</sup>, or its uneven distribution<sup>4</sup>, the hypothesis is that cartilage pressure leads to mechanical failure<sup>25,30</sup>.

While the physical model did not account for absolute material properties, their relative properties were approximated. The compressive modulus of cortical pelvic bone has been measured to be 5 GPa by Vasu *et al.*<sup>7,29,35</sup>. Although the creep aggregate modulus of cartilage is roughly 1 MPa<sup>4</sup>, the impact modulus of cartilage was chosen to be 10 MPa since *in-vivo* activities produce greater pressures for loading rates that represent moments of greatest pelvic trauma<sup>14</sup>. Given these values, the relative stiffness between pelvic bone and cartilage moduli were maintained by the Sawbone pelvis and RTV rubber. Even with these adjustments, however, the slopes of the regression between computed vs measured pressure for the *ex-vivo* and plastic models were statistically differ-

ent. This may be partially due to the fact that the stiff femoral prosthesis elevated contact pressures, compared to a normal femoral head. If a femoral cartilage layer were added, the slopes may have been more similar; however, two simulated cartilage layers seemed to produce excessive shear. Consistent with our findings, we note that the pressures measured in patients with endoprostheses<sup>14</sup> are greater than those predicted by Maxian *et al.* from plane film radiographs<sup>22</sup>.

This model is also based on the assumption that the femur is spherical and concentrically reduced. While this is a reasonable assumption for many patients, the assumption may not be appropriate for patients with severely subluxated hips or patients with cerebral palsy<sup>2</sup>. A further limitation is that all contact pressures are assumed to be due to normal loads with no shear force contribution. In addition, our geometry-based model may not be appropriate in cases of severe cartilage degeneration, subchondral bone thickening or abnormal gait cycles.

Fuji pressure-sensitive film has been used extensively in biomechanics, and the limitations associated with its use have been well documented<sup>8,21,32,37</sup>. The small working range of the film (2.5–10.0 MPa for low-range film), tendency to produce artifacts and inability to measure shear forces were all considered. Furthermore, the film's accuracy, which is dependent upon the size of the sample-area used to capture the data and the number of pressure-intervals identified<sup>21</sup>, is reported to be roughly 10%<sup>32</sup>. For each experiment (*ex-vivo* vs physical model), the appropriate film range was selected carefully in order to avoid saturation. To reduce artifacts, a Gaussian filter was incorporated within the image-processing sequence. An acceptable coefficient of variation for repeated film measurements insured that the measurements were precise. Finally, since experiments were conducted in acetabula with the freedom to resolve all non-normally directed forces, neglecting shear forces is reasonable.

While there are no similar studies in the literature, our predicted pressures (after adjusting for a force magnitude of 2.7 times body weight) of 0.96–2.12 MPa in the plastic and cadaveric models compare favorably both with a study by Maxian *et al.*<sup>22</sup>, who calculated that the hip joint can chronically tolerate approximately 2.0 MPa of pressure, and with instrumented prosthesis studies. Although we found that the magnitude of pressures measured using Fuji film was higher than that predicted by the model, the pressures are consistent with another study of pressure-sensitive film used in the hip<sup>3</sup>.

This geometric model may be better than center-edge angles or other measurements made from plane radiographs. Clinicians and researchers currently rely on the radiographic parameters as a diagnostic measure and tool for surgical planning<sup>17,26,27,33,38</sup>. Although the measurements of center-edge angles, acetabular index and uncovering of the femoral head are predictive of the long-term outcome of dysplastic hips<sup>26</sup>, they have not been shown to be related to

contact pressure. Some have developed radiographic methods to compute coverage<sup>13,19</sup>; however, in general, radiographic assessment of coverage is not as accurate as CT-based methods<sup>15</sup>, and there are many necessary assumptions. Others have proposed CT-based models<sup>16,18,23</sup>, but none of these models has been shown to be predictive of contact pressure.

This study has evaluated a geometry-based computer model that can predict contact pressure in the normal and dysplastic hip. Prospective clinical studies are needed to determine whether these predicted peak contact pressures can identify patients who will fail osteotomy or whether these methods can be used to select the best osteotomy for each patient.

## ACKNOWLEDGEMENTS

This study was supported by a grant from the AO/ASIF Research Foundation, by The Joint Center of Radiation Therapy at Harvard Medical School, the Maurice E. Mueller Professorship in Biomechanics at Harvard Medical School, and the Harry and Evelyn Indursky Orthopedic Biomechanics Laboratory at Beth Israel Hospital.

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