INTRODUCTION: Abnormal mechanical loading of articular cartilage in the hip is thought to be the primary cause of hip osteoarthritids (OA). Prediction of cartilage contact mechanics using noninvasive or minimally-invasive methods may improve the diagnosis and treatment of acetabular dysplasia and femoral acetabular impingement, which are thought to be driving etiologies of hip OA. To understand the relationship between altered hip joint biomechanics and OA, the range and variation of stresses in the normal hip must be known. We previously demonstrated that subject-specific finite element (FE) models of the hip, constructed from high resolution CT image data, predicted hip cartilage contact pressures that were in very good agreement with experimental data [1]. The objective of this study was to determine the variations in magnitude and distribution of cartilage contact pressures in normal, healthy hips as predicted by subject-specific FE models.

METHODS: Five subjects (2 female, 3 male) aged 28 – 31 (mean=29) weighing 49 – 85 kg (mean=66 kg) with no history of hip pain were recruited as age- and sex-matched subjects to patients that are commonly treated in our clinic for hip dysplasia. With IRB approval, CT arthrography was performed on a randomly selected hip. The joint space was filled with approx. 20 ml of contrast agent (2:1 iodocaine to Omnipaque). Volumetric CT scans were obtained (120 kVp, mAs=CARE, Dose=100-400, 512 x 512 acquisition matrix, 1.0 pitch, FOV=300-400 mm, axial slice thickness=1.0 mm), using a harel traction device to distract the leg for improved delineation of cartilage layers.

Subject-specific FE models were generated and assigned material properties based on a previously validated modeling protocol [1] (Fig. 1). Models were loaded under simulated conditions of descending stairs, walking and stair-climbing by displacing the femur into the acetabulum to an orientation, magnitude and direction of the average equivalent joint reaction force reported by Bergmann et al. [2], who measured in-vivo hip forces using instrumented prostheses. The implicitly integrated FE code NIKE3D was used for all analyses.

A contact stress of 0.1 MPa was chosen as a lower threshold for calculation of average pressure and contact area. The five acetabular cartilage meshes were mapped onto a single “average” mesh using previously published methods [3]. Pressures for all 5 models were re-sampled to the average mesh. The average mesh was then divided into 4 anatomic regions [4]. A nonparametric Kruskal-Wallis test was used to test for significant differences in mean pressure and variation of pressure between loading scenarios and anatomical regions. Post hoc comparisons were performed using the Dunn Procedure. Significance was set at p<0.05.

RESULTS: The predicted pressures highlight the variations in contact pattern in normal hips, which can be attributed to both an irregular bone/cartilage boundary and a varying cartilage surface topography present in all hips (Fig. 2). Contact pressures moved posteriorly as the anatomical position and direction of the equivalent joint reaction force moved from a predominantly anterior orientation in descending stairs to a posterior orientation in simulated stair climbing. The pressure distributions for stair climbing and walking were most similar. Inter-subject variance in pressure (not shown) was greatest during stair climbing, but descending stairs also showed considerable variance, predominantly confined to the anterior acetabulum.

Highest average peak pressure (13.1 MPa) was predicted in the descending stairs scenario, which corresponded to the highest applied force (261% bodyweight [2]). This peak occurred at the anterior aspect of the lateral region, in alignment with the primary direction of the loading (Fig. 2). Mean pressures in the anterior region during descending stairs were significantly greater (p<0.03) than those in the posterior region. Lateral pressures were greater than the other regions, but only significantly more than the posterior region (p<0.01). Primary loading was more superior during walking, compared to descending stairs, resulting in a peak pressure (9.1 MPa) located in the lateral region. During walking, medial and lateral mean pressures were both significantly greater than those in the posterior region (p=0.01, 0.02, respectively). Stair climbing peak pressure (12.0 MPa) occurred in the medial region. During stair climbing, average pressures moved slightly more posterior, but were still significantly greater in the medial and lateral sections than both the posterior and anterior section (p=0.02, 0.02, 0.04, respectively), corresponding to the posterior rotation of the femoral head with respect to the pelvis.

Average contact areas for descending stairs, walking, and stair climbing were 582 mm², 732 mm², and 735 mm², respectively. Contact areas were inversely related to peak and average pressures - e.g. descending stairs had the highest peak pressure and lowest contact area.

DISCUSSION: The results demonstrate the need to quantify the variation in contact pressures that can be considered normal within healthy hips. For example, during descending stairs, pressures at the anterior acetabulum had standard deviations near 7 MPa. Likewise, pressure deviated ~6 to 7 MPa in the anterior-medial section during stair-climbing. This demonstrates that the location and magnitude of pressure can vary substantially between subjects despite identical loading conditions. This is likely due to variations in bone and cartilage geometry.

Significant differences in pressure across loading scenarios and anatomical locations demonstrate the importance of analyzing several physiological orientations. The shifting contact pattern followed an expected trend with regard to the changing anatomical orientation and direction of the equivalent joint reaction force (Fig. 2). Interestingly, all of the hips manifested a bicentric contact pattern, regardless of activity.

Models relying on perfectly spherical joints or uniform cartilage thickness could not predict pressures with the degree of specificity used in this study. Prior hip models [e.g., 4] assumed perfectly congruent geometry and predicted simplistic patterns of contact, with peak pressures on the order of 3 MPa. In the present study, models exhibited irregular contact patterns with much higher peak pressures. Since the present predictions of peak and average pressures and contact areas are in better agreement with in vitro experimental data [1, 5], it is likely that the adopted patient-specific modeling approach yields more accurate predictions of cartilage pressure than those that simplify hip geometry.

Future work will apply these modeling techniques to a larger number of normal subjects to develop a comprehensive picture of contact pressure variations in a normal population. Ultimately, the methodologies can be used to study patients with hip diseases such as developmental dysplasia and femoracetablar impingement to elucidate the link between altered hip mechanics and development of hip OA.

ACKNOWLEDGMENTS: Financial support provided by NIH #R01AR53344 and NIH #RO1EB007689.