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Patient-specific chondrolabral contact mechanics in patients with acetabular dysplasia following treatment with peri-acetabular osteotomy

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SUMMARY

Objective: Using a validated, patient-specific finite element (FE) modeling protocol, we evaluated cartilage and labrum (i.e., chondrolabral) mechanics before and after peri-acetabular osteotomy (PAO) to provide insight into the ability of this procedure to improve mechanics in dysplastic hips.

Design: Five patients with acetabular dysplasia were recruited in this case-controlled, prospective study. Models, which included anatomy for bone, cartilage, and labrum, were generated from computed tomography (CT) arthrography scans acquired before and after PAO. Cartilage and labrum contact stress and contact area were quantified overall and regionally. Load supported by the labrum, expressed as a percentage of the total hip force, was analyzed.

Results: Percent cartilage contact area increased post-operatively overall, medially, and superiorly. Peak acetabular contact stress decreased overall, laterally, anteriorly, and superiorly. Average contact stress decreased overall, laterally, anteriorly, and posteriorly. Only average contact stress on the superior labrum and peak labrum stress overall decreased. Load supported by the labrum did not change significantly.

Conclusions: PAO was efficacious at medializing cartilage contact and reducing cartilage contact stresses, and therefore may minimize deleterious loading to focal cartilage lesions, subchondral cysts, and cartilage delaminations often observed in the lateral acetabulum of dysplastic hips. However, the excessively prominent, hypertrophied labrum of dysplastic hips remains in contact with the femoral head, which continues to load the labrum following PAO. The clinical ramifications of continued labral loading following PAO are not known. However, it is plausible that failure to reduce the load experienced by the labrum could result in end-stage hip OA following PAO.

Introduction

The etiology of hip osteoarthritis is multifactorial; both whole-body-level factors (e.g., race, diet, weight, sex, genetics) and joint-level factors (e.g., joint morphology, muscle function) are involved[1]. Nevertheless, most cases of hip OA occur secondary to untreated anatomical deformities, such as acetabular dysplasia and femoroacetabular impingement[1]. There is some disagreement as to which percentage of hip OA cases can be attributed to each deformity. For example, investigators have estimated dysplasia accounts for 50% of hip OA cases[2,3], whereas femoroacetabular impingement has been suggested to be more common, being observed in 80% of cases[4]. Regardless, it is generally agreed that hip OA develops in these hips as a result of abnormal cartilage and labrum (i.e., chondrolabral) mechanics, which induce structural failure and a cascade of molecular and inflammatory responses that typify hip OA[1,5].

In dysplastic hips, the acetabulum is shallow, resulting in a joint with inadequate femoral head coverage[6,7]. Reduced coverage in dysplastic hips is hypothesized to cause chronic overload of

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cartilage, resulting in end-stage OA[8,9]. Periacetabular osteotomy (PAO) aims to prevent OA by reorienting the acetabulum into a position that increases anterolateral coverage[8,10,11]. Medialization of the joint may reduce cartilage stresses along the lateral border of the acetabulum, which is important as many dysplastic hips have cartilage lesions in this region[12]. PAO may also reduce load and stress at the labrum. Normalization of labral mechanics may be critical: recent finite element (FE) modeling research suggests that it is the labrum that may initially experience stress overload in pre-osteoarthritic hips with dysplasia, rather than cartilage, which may lead to an out-to-in progression of OA[13,14].

Clinical studies have demonstrated positive outcomes after PAO[15–17], but 40% of these patients eventually require hip arthroplasty[18]. Measurements of chondrolabral mechanics, including quantification of stress, contact area, and load sharing, before and after PAO, would establish the biomechanical efficacy of this procedure. Chondrolabral mechanics cannot be measured *in vivo*, but they can be predicted from FE models[19,20]. Generating three-dimensional (3D) reconstructions of bone, cartilage, and labrum from medical images to serve as FE model geometry is time-consuming[21]. Thus, hip FE models have often assumed constant thickness for cartilage and concentric bone and cartilage[9,22]. Unfortunately, these simplifications yield unrealistic predictions[23,24]. More recent FE models of native hips include cartilage thickness[13,14,19,20,25–27], but most of these models neglect the labrum. Importantly, simulations that analyzed hip mechanics before and after PAO excluded the labrum and did not incorporate spatially varying cartilage thickness[22,28].

The objective of this study was to predict chondrolabral mechanics before and after PAO using FE models that included patient-specific anatomy. We hypothesized that PAO would: (1) medialize cartilage contact stresses and reduce average and peak cartilage contact stress, and (2) reduce peak and average stress, contact area, as well as load to the labrum.

Methods

Patient recruitment, radiographic evaluation, CT arthrography

All research was performed in accordance with the Helsinki Declaration with informed consent and institutional board approval. Patient demographics were reported as the mean ± standard deviation. Five patients with acetabular dysplasia (1 male and 4 female) aged 29.8 ± 5.8 years and with a body mass index (BMI) of 21.1 ± 3.6 kg m^{-2} underwent PAO by a single surgeon (author CLP) with 20 years of experience managing dysplasia. Each patient was imaged using radiographs and computed tomography (CT) arthrography before and after surgery, at minimum 1 year follow-up (19.3 ± 7.2 months, yielding a post-operative age of 31.5 ± 6.3 years). BMI following surgery was 21.4 ± 2.9 kg m^{-2}. Anteroposterior radiographs evaluated morphology in pre- and post-operative states.

A previously described CT arthrogram protocol[26] was performed to visualize, within a single image sequence, opposing layers of cartilage, the labrum, cortical bone, and trabecular bone. Images were acquired with a 128-section single-source CT scanner (SOMATOM Definition™; Siemens Healthcare) with the following settings: 120 kVp, 100–400 mA, 512 × 512 matrix, 1.0 pitch, 300–400 mm FOV, and 0.7 mm slice thickness. A Hare traction splint was applied during the CT scan to ensure that contrast agent imbibed the joint space[26].

FE model generation

FE models were generated from the CT images using a validated protocol[19,20]. Briefly, CT images were up-sampled to three times their native resolution to reduce stair-case artifact in 3D reconstructions[25]. The CT images were then segmented semi-automatically using commercial software (Amira, v6.0, FEI, Hillsboro, OR) to generate 3D reconstructions of trabecular bone, cortical bone, pelvic and femoral cartilage, and the acetabular labrum.

The 3D reconstructions were smoothed, decimated and discretized into FE meshes (Fig. 1). Here, cortical bone was represented as triangular shell elements with position-dependent thickness, calculated as the geometric distance between the inner and outer cortex[19]. Pelvic and femoral cartilage as well as the acetabular labrum was represented as hexahedral elements[13,19,25,27]. The boundary between cartilage and labrum was assumed to be located where the concave acetabulum transitioned into the convex acetabular rim[13,14]. Element densities were based on previous mesh convergence analyses[19].

Constitutive models for bone, cartilage, and labrum followed other FE studies of the hip[14,19,25]. Here, bone was represented as isotropic linear elastic ($E = 17$ GPa, $\nu = 0.29$)[31]. Cartilage was represented as a nearly incompressible, neo-Hookean hyperelastic material ($G = 13.6$ MPa, $K = 1359$ MPa)[19,32]. The labrum was represented as transversely isotropic hyperelastic[34] with material coefficients ($C_1 = 1.4$ MPa, $C_3 = 0.005$ MPa, $C_4 = 36$, $C_5 = 66$ MPa, $\lambda^* = 1.103$) derived from experimental data of bovine tissue[35]. Here, C_1 referenced the shear modulus; equations describing the behavior of the fibers included material coefficients that scaled the exponential stress (C_3), specified the rate of collagen uncramping (C_4), the modulus of straightened collagen (C_5), and the stretch at which collagen straightened (λ^*).

A range of anatomical positions and loads, expressed in percent body-weight (BW), were applied to each FE model to analyze activities encountered during daily life. These included walking at toe-off (WTO, 205% BW), midstance during walking (WM 203% BW), the transition of heel-strike and midstance for stair descent (DHM 230% BW), and heel-strike during stair ascent (AH, 252% BW) using the Bergmann dataset[36]. During loading, the pubis and sacroiliac joint were held fixed, but the remaining hemipelvis and femur were free to deform. The femur was translated along the loading axis until the desired load was achieved, but was free to translate in the plane normal to the loading axis to achieve equilibrium[13,25,27]. Tied and sliding contact definitions followed previous FE studies[13,14]. All FE models were analyzed with NIKE3D[37].

Measures of chondrolabral mechanics

Peak and average contact stress and contact area were recorded on the surface of the acetabular cartilage and labrum. Only those FE nodes in contact (i.e., >0.0 MPa) were considered in the calculation of the average stress. The load supported by the labrum was reported as a percentage of the total force transferred across the hip. Contact stress and contact area were evaluated in the lateral and medial regions [Fig. 2(a)], and in the anterior, superior, and posterior regions [Fig. 2(b)]. Contact area was presented as a percentage of the total surface area of acetabular cartilage. Fringe plots of contact stress for each subject and activity were generated. Similar plots were created to visualize average stresses at each FE mesh node for the acetabular cartilage. The same number of node and elements were used to represent acetabular cartilage across subjects; nodal connectivity was also preserved across subjects. This one-to-one correspondence made it straightforward to average nodal stresses. However, it was necessary to select a representative mesh to visualize average nodal stress. To select the representative mesh, the articulating surface of acetabular cartilage from each patient mesh was fit to a sphere. Next, the average radius of the sphere fit for subjects was calculated. The single patient-specific