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Discrete Element Analysis is a Valid Method for Computing Joint Contact Stress in the Hip before and after Acetabular Fracture

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Abstract

Evaluation of abnormalities in joint contact stress that develop after inaccurate reduction of an acetabular fracture may provide a potential means for predicting the risk of developing post-traumatic osteoarthritis. Discrete element analysis (DEA) is a computational technique for calculating intra-articular contact stress distributions in a fraction of the time required to obtain the same information using the more commonly employed finite element analysis technique. The goal of this work was to validate the accuracy of DEA-computed contact stress against physical measurements of contact stress made in cadaveric hips using Tekscan sensors. Four static loading tests in a variety of poses from heel-strike to toe-off were performed in two different cadaveric hip specimens with the acetabulum intact and again with an intentionally malreduced posterior wall acetabular fracture. DEA-computed contact stress was compared on a point-by-point basis to stress measured from the physical experiments. There was good agreement between computed and measured contact stress over the entire contact area (correlation coefficients ranged from 0.88-0.99). DEA-computed peak contact stress was within an average of 0.5 MPa (range 0.2 MPa - 0.8 MPa) of the Tekscan peak stress for intact hips, and within an average of 0.6 MPa (range 0 - 1.6 MPa) for fractured cases. DEA-computed contact areas were within an average of 33% of the Tekscan-measured areas (range: 1.4% - 60%). These results indicate that the DEA methodology is a valid method for accurately estimating contact stress in both intact and fractured hips.

Keywords

Discrete Element Analysis; Acetabular Fracture; Hip; Contact Stress; Tekscan

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Introduction

It has been reported that as many as one in four acetabular fracture patients will rapidly progress to develop post-traumatic osteoarthritis (PTOA) of the hip (Bhandari et al., 2006; Dirschl et al., 2004; Matta, 1996; Saterbak et al., 2000). The likelihood of developing PTOA after an intra-articular fracture has been associated with quality of articular reduction and increases in joint contact stress (Anderson et al., 2011; Kern and Anderson, 2015). Residual incongruity is common in surgically reconstructed acetabular fractures, where incongruities of 1-3 mm are often considered satisfactory reductions (Borrelli et al., 2002; Moed et al., 2003) despite biomechanical studies demonstrating that 1 mm and 2 mm articular step-offs increase joint contact stress by 23% and 48%, respectively (Malkani et al., 2001).

The primary goal of surgical reduction is restoration of an anatomic joint surface, the necessity of which has been illustrated by numerous mechanical studies across both fractured and intact joints showing a direct correlation between elevated joint contact stress and development of osteoarthritis (Anderson et al., 2011; Hadley et al., 1990; Kern and Anderson, 2015; Maxian et al., 1995; Segal et al., 2009; Segal et al., 2012). This association indicates that joint contact stress may be useful as a predictor of a patient's risk for developing PTOA. Though joint contact stress can be obtained *in vivo* (Anderson et al., 2003; Bergmann et al., 2001; Bergmann et al., 1999; Hodge et al., 1989; Hodge et al., 1986), such measurements are highly invasive and are only realistic in small patient cohorts and under extremely controlled conditions.

Joint contact stress is most commonly assessed in living patients using non-invasive computational modeling techniques such as finite element analysis (FEA) (Harris et al., 2012; Henak et al., 2014a; Rhyu et al., 2011). Unfortunately, the technical burden required to develop and run patient-specific FE models is substantial, with mesh generation and establishment of numerically stable contact conditions requiring many days or weeks of effort, and model runtimes that range from hours to days depending on model complexity. Consequently, even purportedly large-scale patient-specific FEA investigations have been limited to less than 15-20 patients (Harris et al., 2012; Henak et al., 2014a; Li et al., 2008).

Discrete element analysis (DEA) is another computational method that has been utilized for determining intra-articular contact stress in many different joints (Anderson et al., 2010b; Chao et al., 2010; Volokh et al., 2007). In DEA, cartilage is modeled as an array of compressive springs between rigid bony surfaces, eliminating the need for development of a continuum mesh and greatly simplifying model generation relative to FEA. Contact stress is computed by balancing applied loads with cartilage spring forces using numerically stable and rapidly executing methods that provide solutions within seconds. While DEA methodology sacrifices continuum mechanics information and the use of advanced material property definitions, it can provide joint contact stress estimations in a fraction of the time required to obtain similar information using FEA, making it an appealing technique for application to larger patient cohorts (Anderson et al., 2010b; Segal et al., 2009; Segal et al., 2012).

DEA models of articular joints can be developed using specimen-specific cartilage surfaces (Abraham et al., 2013; Segal et al., 2009; Segal et al., 2012) or approximated cartilage surfaces – most often generated by prescribing a uniform projection from subchondral bone (Anderson et al., 2007; Kern and Anderson, 2015). In some DEA models of the hip, cartilage surfaces have been defined as spherically congruent projections from the subchondral bone (Genda et al., 2001; Yoshida et al., 2006), which results in calculation of contact stress values lower than those measured in physical experiments (Hodge et al., 1989; Hodge et al., 1986). In contrast, systematic exploration of many of the assumptions of the DEA methodology has shown that projected cartilage surfaces and rigid underlying bone result in calculation of contact stress values higher than those measured physically (Anderson et al., 2010a), and even DEA performed using specimen-specific cartilage data results in contact stress at least 15% higher than that obtained using FEA (Abraham et al., 2013). Given that specimen-specific DEA models tend to over-predict, and spherically approximated DEA models tend to under-predict hip contact stress, there would seem to be some level of model approximation that could yield accurate calculations of contact stress using the DEA methodology.

To obtain hip contact stresses in acetabular fracture patients using DEA, model generation must be from a CT scan, which (without the use of a contrast agent) does not show cartilage. As metal artifact from fracture fixation hardware precludes use of MRI scans, and contrast from a CT arthrogram would leak through fractured surfaces, it is presently not possible to obtain cartilage geometry directly from clinical imaging in acetabular fracture patients. Therefore, cartilage information must be approximated from bony geometry for these patients, an approximation that it is important to validate to determine the accuracy of resulting contact stress calculations. The objective of this work was to determine if DEA estimates of contact stress in both intact and imperfectly reduced fractured hips are valid by comparing DEA-computed contact stress to those physically measured from cadaveric preparations.

Methods

Specimen preparation

Two fresh-frozen cadaveric hemipelves were used for physical loading experiments. Specimen 1 was from a 170 cm, 63.5 kg, 71 year old male, and Specimen 2 was from a 183 cm, 79.4 kg, 37 year old male. Each was carefully dissected free of soft tissues down to the joint capsule and the acetabular labrum. The ilium was cut in an approximately transverse plane 10 cm superior to the acetabulum to fit in a pre-existing potting device. Using previously described instrumentation (Martin et al., 2015), the cut pelvis was oriented so that a vertically directed load along the axis of the femur corresponded with the hip loading vector during maximum load during walking (15 degrees abduction and 25 degrees flexion) (Bergmann et al., 2001). The ilium was embedded in a polymethylmethacrylate (PMMA) block in this position. The specimen was then inverted and the femur was transected mid-shaft and embedded vertically in a 5-cm diameter PMMA cylinder.

Loading apparatus

The experimental loading apparatus (Figure 1) was mounted in an MTS Bionix 858 test machine (MTS Inc., Minneapolis, MN). The lower assembly consisted of a compound sine plate and a set of dual orthogonal linear bearings (an XY-stage) attached to the load cell. The upper assembly was a short metal tube attached to the MTS linear/rotary actuator. For testing, each specimen was inverted and the pelvic PMMA block was bolted to the sine plate, which controlled hip flexion/extension and abduction/adduction. The femoral PMMA cylinder was locked into the upper tube so that the MTS could simultaneously apply axial load and femoral rotation. Once oriented in the testing fixture, the entire joint capsule was excised and the acetabular labrum was partially removed to permit insertion of the Tekscan sensor. Upon opening the joint, we visually confirmed macroscopically normal cartilage in Specimen 1 and noted the focal defect in Specimen 2 was sufficiently medial within the acetabulum so as to not be loaded in any testing poses. Specimens were sprayed copiously with saline throughout the potting process and each loading tests to prevent cartilage dehydration/degeneration.

Testing: Intact specimens

Specimen 1 was tested in a heel-strike pose (16.3° flexion; -5.5° abduction; 5.4° external rotation), and Specimen 2 was positioned to simulate heel-strike, mid-stance (0° flexion; -8.3° abduction; 6.7° external rotation), and toe-off (-6.9° flexion; -4° abduction; 8.7° external rotation) of normal walking gait (Bergmann et al., 2001). At the beginning of each test, the MTS was used to distract the femur from the acetabulum for insertion of a calibrated hip-specific pressure sensor (Model 4402, Tekscan, Inc. Boston, MA) (Rudert et al., 2014). The sensor was lubricated with petroleum jelly to minimize shear damage. Two investigators held the sensor with the outermost ring of the sensor even with the lateral rim of the acetabulum while the MTS actuator pushed the femoral head into contact with the acetabulum with 50 N of force. This compressive force then held the sensor in place for testing. For each test, the actuator applied a ramp load to 1,000 N, and joint contact pressure data were then recorded by the Tekscan sensor.

With the load still applied, a threaded rod was installed to lock the relative positions of the sine plate and femur tube (Figure 1, right) so that the specimen and sensor could be removed from the MTS as a rigid construct and transported for CT scanning (0.39 mm × 0.39 mm × 0.5 mm voxels). The sensing elements of the Tekscan sensor were visible in the CT scans (Figure 2), allowing for the Tekscan sensor location to be spatially registered to the adjacent bony anatomy. After CT scanning, the locking rod was removed, the femur and the Tekscan sensor were removed from the acetabulum, and the specimen was remounted in the MTS for the next gait pose. A newly calibrated Tekscan sensor was inserted and the pressure recording/CT-scan sequence was repeated.

Testing: Mal-reduced fracture specimens

After completion of the experiments in the various gait poses, an acetabular fracture was created in each specimen using a combination of saw and osteotome cuts. Fractures were fixed by an orthopaedic trauma surgeon using two interfragmentary screws and a buttress plate supported by two distal and two proximal screws (Figure 3). In Specimen 1, the

fracture was intentionally reduced with a uniform 2-mm recessed step-off, and in Specimen 2 the fracture was reduced with an intentional inter-fragmentary gap plus a 1-2 mm recessed step-off. These malreductions are surgically realistic and would likely be considered acceptable reductions in a clinical setting (Ebraheim et al., 2007; Matta et al., 1986). The loading/pressure recording/CT-scan sequence was repeated at each gait pose in the fractured specimens.

DEA Model Generation

Two sets of DEA models were generated for comparison to the physical loading experiments. The first set incorporated specimen-specific cartilage thickness information obtained from an MRI scan (dual echo steady-state; 0.5 mm isotropic voxels) (Abraham et al., 2015) acquired prior to testing while the specimen was intact and unloaded. Femoral head and acetabular articular cartilage surfaces and underlying bony anatomy were manually segmented from the MRI scans using OsiriX software (Pixmeo, Geneva, Switzerland) and converted into triangulated surfaces using Geomagic Studio (Geomagic Inc., Research Triangle Park, NC). Cartilage thickness was defined as the perpendicular distance from each triangular facet on the articular surface to the subchondral bone. An iterative closest point algorithm was used to align the MRI-derived bone surfaces with CT-derived bone surfaces obtained with the bones locked in each testing position. Those transformations were then applied to the MRI-derived specimen-specific cartilage surfaces in order to bring them into the orientations used during physical loading experiments.

The second set of DEA models was generated entirely from the CT scans obtained after each physical loading experiment, simulating the most realistic option for DEA model generation in acetabular fracture patients. For these models, the cartilage surface was generated from each post-testing CT scan by isolating femoral head and acetabular subchondral bone from the full bone surfaces, and projecting that subchondral bone geometry 1 mm into the joint space along local surface normals. These projected articular surfaces were then iteratively smoothed towards sphericity using a custom algorithm (Shivanna, 2006). The Euclidean distance of each cartilage surface vertex from the center of a sphere fit to the entire projected surface was calculated. The location of each cartilage surface vertex was adjusted along a ray connecting that vertex to the sphere center by a distance equal to the difference between the vertex's radius and the average radius of the neighboring vertices within a 2 mm neighbor threshold. A 0.05 mm maximum radial change for each of smoothing 5 iterations was allowed, permitting a total maximum radial change of 0.25 mm, approximately half the distance of a CT voxel. The number of smoothing iterations was determined empirically from the intact heel strike model of Specimen 1 by choosing the number of iterations after which the DEA-computed contact stresses were closest to the physically measured contact stress data. As the goal of this work was to validate the results of a standardized modeling approach, these parameters were applied uniformly to generate the other seven projected-cartilage DEA models.

Contact Stress Calculations

The Tekscan sensor was calibrated (Rudert et al., 2014) before and upon completion of testing, and equal weighting was applied to the pre-test and post-test calibration curves when

converting Tekscan raw pressure values (0-255) into contact stress values (MPa). A sensor was used for a maximum of two tests to limit the effects of sensor degradation.

For all DEA models, cartilage was assigned isotropic linear-elastic material properties ($E = 12 \text{ MPa}$, $\nu = 0.42$) (Kern and Anderson, 2015). Boundary and loading conditions applied to the models were identical to the physical loading experiments. DEA contact stress solutions were computed using a Newton's method solver implemented in MATLAB (Mathworks, Natick, MA) to match DEA spring forces with applied boundary conditions (Kern and Anderson, 2015). Run time for each DEA model was approximately 10 seconds in Matlab 2013b running on a desktop computer (IntelCore i7-6700 3.4 Gz; 16384MB RAM; 64-bit operating system).

Contact Stress Comparisons

The piezoresistive wires of the Tekscan sensor were visible in each CT scan (Figure 2), permitting direct segmentation and 3D surfacing of the Tekscan sensor as it was positioned in the joint during testing. Tekscan pressure data were mapped onto the 3D sensor geometry by subdividing the triangulated 3D sensor surface (mesh size $\sim 0.0295 \text{ mm}^2$) into the 21 circumferential rings and 52 radial spokes of the physical sensor (sensel size 1.923 mm^2), which resulted in approximately 65 surface triangles per Tekscan sensel. Every triangular facet within a sensel was assigned the measured value of that sensel. A ray casting technique was then used to identify the surface vertex on the acetabular cartilage surface closest location of each Tekscan sensor vertex (Figure 4). Spatial agreement between Tekscan and DEA data sets was then assessed using point-to-point correlation, omitting locations where both DEA and Tekscan were reporting zero stress. Peak contact stress and contact area were also compared between DEA models and physical loading experiments.

Results

Specimen-Specific Cartilage Models

Average cartilage thickness for Specimen 1 was 0.96 mm on the femoral head and 1.34 mm on the acetabulum. The average cartilage thickness for Specimen 2 was 1.33 mm on the femoral head and 1.67 mm on the acetabulum. Contact stress values computed using specimen-specific cartilage DEA models were highly correlated with Tekscan-measured values for intact hips (correlations greater than 0.94 in all phases of the gait cycle – Table 1). Specimen-specific cartilage DEA models also accurately approximated Tekscan-measured peak contact stress (average difference of 11%; range: 4% - 25%) for the intact state. Contact areas from DEA were within an average of 33% (range: 18% - 48%) of the physical measurements for intact hips.

Contact stress values computed using specimen-specific cartilage DEA models of a malreduced posterior wall acetabular fracture were also highly correlated with Tekscan-measured values (correlations greater than 0.91 – Table 1). Peak contact stress values computed were also within an average of 11% (range: 9%-15%) of the Tekscan-measured values. DEA-computed contact areas were within 30% (range: 16% - 36%) of physical measurements for the malreduced fractures.

Point-by-point absolute difference maps show visually that cartilage-specific DEA predictions had similar contact stress distributions as the Tekscan measurements (Figure 5). Areas of high contact stress in the DEA models appeared to be in similar locations as high contract stresses measured during physical testing. In the malreduced fracture state, both specimens exhibited bimodal contact anterior and posterior to the fragment. The fracture fragment in Specimen #1 was uniformly recessed 2 mm and did not contact the femoral head at heel strike. The fracture fragment for Specimen #2 had a better reduction near the acetabular rim, and thus the fragment did experience load at all instances of gait investigated in this work.

CT-Projected Cartilage Models

DEA models of intact hips with cartilage surfaces that were projected from CT scans and smoothed using our algorithm were also able to accurately approximate physical measurements of contact stress and area. DEA-computed contact stress values were highly correlated with Tekscan-measured values for intact hips (correlations greater than 0.93 in all phases of the gait cycle – Table 1). Peak contact stress in the projected cartilage models was within an average of 12% (range: 4% - 29%) of the Tekscan measurements for the intact hips, and contact areas were within an average of 12% (range: 4% - 29%).

DEA-computed contact stress values for the malreduced posterior wall acetabular fractures also highly correlated with Tekscan-measured values (correlations greater than 0.88 in all phases of the gait cycle – Table 1). As for the specimen-specific cartilage models, peak contact stress computed with projected cartilage DEA models of malreduced posterior wall acetabular fractures had excellent agreement (average within 6%; range: 0%-12%) with the Tekscan-measured contact forces. Contact areas were within 48% (range: 29% - 60%) for the malreduced fractures.

Point-by-point absolute difference maps of the projected cartilage DEA contact stress maps (Figure 6) show agreement between DEA-generated and Tekscan measured contact stress that is very similar (though not identical) to the good agreement found using specimen-specific cartilage models. Areas of high contact stress on the acetabular cartilage appeared in similar locations on the projected-cartilage DEA models and the Tekscan sensor. Models of both malreduced fracture specimens exhibited similar contact stress and loading patterns as those seen with the cartilage-specific models and the Tekscan measurements.

Discussion

The goal of this work was to validate the use of a DEA methodology for estimating contact stress in the context of acetabular fractures. We demonstrated that with the use of specimen-specific cartilage thickness information, there was excellent agreement in terms of peak contact stress, contact area, and spatial correlation between DEA-computed and Tekscan-measured contact stress. Unfortunately, specimen-specific cartilage data is likely to remain unavailable when dealing with acetabular fractures patients, and implementation of DEA for obtaining contact stress in fractured joints will continue to rely on cartilage surface models extrapolated from clinical CT data. This work demonstrated that with an appropriate smoothing algorithm applied to a cartilage surface extrapolated from bony geometry, the

resulting DEA contact stress estimations had excellent agreement with physically measured intra-articular contact stress values ($r = 0.93$ for intact state, $r = 0.88$ for fractured state).

While this work was to the best of our knowledge the first to use a Tekscan sensor to explore contact stress in the hip after an acetabular fracture, a variety of comparative information about joint contact stress in intact hips and in hips with acetabular fractures is available from studies using pressure-sensitive film. The average contact area across all DEA models in our work was 6.3 cm^2 in intact hips and 4.1 cm^2 in fractured hips, values similar to the range of contact areas that have been measured for intact hips (range: $4.3 \text{ cm}^2 - 6.8 \text{ cm}^2$) and fractured hips (range: $2.9 \text{ cm}^2 - 5.3 \text{ cm}^2$) using pressure-sensitive film (Hak et al, 1998; Konrath et al, 1998b; Olson et al, 1995; Olson et al, 1996). Peak contact stress values measured in this work ranged from $2.0 \text{ MPa} - 6.5 \text{ MPa}$ for intact hips and $4.4 \text{ MPa} - 11.3 \text{ MPa}$ for fractured hips. These peak stress values were on the lower end of the range reported for experiments using pressure-sensitive film ($5.62 - 10.54 \text{ MPa}$ for intact; $6.31 - 20.5 \text{ MPa}$ for fractured) (Hak et al, 1998; Konrath et al, 1998b; Olson et al, 1995). This is likely in part due to the higher spatial resolution of pressure-sensitive film compared to the Tekscan sensor permitting measurement of highly localized higher peak pressures and the fact that our loading experiments used approximately half of the applied load of those previous studies.

Despite the excellent agreement between the DEA-computed and the experimentally measured contact stresses in a variety of hip positions, this work has several limitations. Neither the DEA model nor the cadaveric testing included an acetabular labrum. Omitting a labrum was determined acceptable for this work because while the labrum affects intra-articular fluid pressure and joint lubrication (Ferguson et al., 2003; Song et al., 2012), cadaveric and computational studies have indicated that absence of an intact acetabular labrum minimally affect hip stability or intra-articular contact stress in normal hips (Crawford et al., 2007; Henak et al., 2011; Konrath et al., 1998a; Myers et al., 2011). Furthermore, the labrum is likely to be disrupted in a fractured hip.

The 1 mm projected cartilage thickness was thinner than the MRI-measured cartilage thickness in our specimens, and on the very low end of literature values reported for acetabular or femoral head cartilage thickness (Adam et al., 1998; Athanasiou et al., 1994; Shepherd and Seedhom, 1999). This thickness was selected because it resulted in DEA contact patch locations being most consistent with the physically measured contact. Thicker cartilage projections seemed to prevent the femoral head from seating in the acetabulum, and the DEA computed contact stress pattern appeared as rim loading (Figure 7) that did not correspond to the measured spatial pattern of contact stress (Figure 6). Peak contact stress and contact area were most sensitive to changes in cartilage modulus with a 1 mm cartilage projection, yet even then, key measures of contact stress were minimally affected by variation in cartilage Young's modulus (Figure 7). This greater effect of geometry than material properties on stress is similar to findings in other orthopaedic soft tissues (Hansen et al., 2017).

While the loading conditions used in this work were similar to loads seen *in vivo*, some of the loading characteristics and model assumptions limit generalization to a wider clinical

population. While we did vary the position of the hip to simulate multiple instances of the stance phase of gait, we did not vary the applied load (1000 N). Furthermore, hip positions used in this work were based upon the average of the instrumented hip data acquired by Bergmann, et al. (Bergmann et al., 2001; Bergmann et al., 1999), and thus were representative of hip orientation in elderly individuals rather than those typically suffering acetabular fractures. Thus, joint contact data determined in this work should be considered representative examples of contact patterns that may develop in a fractured acetabulum rather than a complete description of contact stress occurring in the acetabulum after a posterior wall fracture.

We elected to report the agreement between DEA and contact stress measurements in positions simulating gait, however we also tested Specimen 1 in several arbitrary abduction poses not representative of gait and found similar agreement between DEA and physically measured contact stresses. Peak contact stress at 11° of abduction was 6.84 MPa and 6.80 MPa for the DEA and Tekscan, respectively. At 5° abduction, DEA predicted a maximum contact stress of 4.88 MPa compared to Tekscan measurement of 4.66 MPa. At 4° adduction, DEA predicted a maximum contact stress of 5.53 MPa and Tekscan measured 5.27 MPa. Testing was performed in the ab/adducted poses only with an intact acetabulum due to concerns about stability of the experimental setup with the fracture fragment location relative to the acetabular loading pattern in these poses.

The use of DEA to assess contact stress in the hip has previously been validated (Abraham et al., 2013) using comparisons to FE models and to pressure-sensitive film measurements of contact stress in cadavers (Anderson et al., 2007; Henak et al., 2014b). That work emphasized the non-spherical nature of the femoral head and encouraged the use of specimen-specific cartilage thickness to obtain realistic results (Abraham et al., 2013). However, in cases of intra-articular fractures, specimen-specific cartilage thickness is unavailable. In this work, we have shown that our DEA methodology is valid for cases of an incongruent fractured acetabulum, and further, if a suitable smoothing algorithm is applied, a cartilage surface projected from underlying subchondral bone will yield valid contact stress information. This allows for implementation of DEA in cases of intra-articular hip fractures in which standard clinical CT imaging is the standard of care, thereby facilitating large, patient-specific population studies of contact stress in studies of acetabular fractures.

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Figure 1.

Photographs of the experimental loading fixture mounted in the MTS. The hip is inverted relative to its physiologic orientation. The individual fixturing components are indicated, and the axes about which anatomic rotations occur are shown (left). Upon completion of loading, a locking rod was added to the fixture (right) to maintain the relative positions of the femoral pot and the compound sine plate, and the portion of the device between the two yellow planes was transferred to the CT scanner for imaging.

Figure 2.
CT scan slice with the conductive wires of the Tekscan sensor visible (arrows).

Figure 3.

Photographs of the two different posterior wall fracture malreductions used for validating DEA-computed contact stress. Specimen 1 had a fairly uniform 2 mm step-off around the entirety of the fragment (arrows), while the majority of the malreduction in Specimen 2 was deeper in the acetabulum (arrows). Along the acetabular rim, the fracture fragment was relatively well aligned (brackets), making this a very clinically realistic pattern of malalignment. The blue lines on the labrum in Specimen 1 were intended to assist returning the specimen to the MTS in the same femoral rotation after fracture. These lines are absent in Specimen 2 because instead of the tissue, the PMMA blocks were marked during testing for this purpose.

Figure 4.

(a) Photograph of the Tekscan sensor positioned inside the acetabulum of Specimen 1 for experimental testing. The circumferential rings of the sensor (black) are visible. (b) Surface model corresponding to Specimen 1 illustrating the segmented acetabular cartilage surface (blue) and the segmented Tekscan sensor (green). (c) Tekscan data were project to the cartilage surface to spatially register the Tekscan pressure data for direct comparison with DEA-generated contact stress distributions.

Figure 5.

Contact stress distributions at heel strike for the intact and fractured states of both specimens obtained from DEA models generated with specimen-specific cartilage thickness. The leftmost column shows the acetabular cartilage in light gray and the Tekscan sensor surface in dark gray with the mapped contact stresses shown as indicated on the colorbar. Dark gray indicates that the Tekscan sensor registered no contact at that location under load. In the center column, the light gray shows all acetabular cartilage in the DEA models with a black line denoting the edge of the Tekscan sensor which is not shown. The rightmost column shows difference maps which are the absolute value of the Tekscan maps minus the DEA maps. Contact stress values above 8 MPa are represented in black, with a maximum contact stress value of 11.3 MPa. The presence of a fracture disrupts the peripheral contact distribution of the intact joints and appears to shift stresses anterosuperiorly and medially within the acetabulum.

Figure 6.

Contact stress distributions at heel strike for the intact and fractured states of both specimens obtained using DEA models generated by projecting cartilage from CT scans. The leftmost column shows the acetabular cartilage in light gray and the Tekscan sensor surface in dark gray with the mapped contact stresses shown as indicated on the colorbar. Dark gray indicates that the Tekscan sensor registered no contact at that location under load. In the center column, the light gray shows all acetabular cartilage in the DEA models with a black line denoting the edge of the Tekscan sensor which is not shown. The rightmost column shows difference maps which are the absolute value of the Tekscan maps minus the DEA maps. Contact stress values above 8 MPa are represented in black, with a maximum contact stress value of 11.3 MPa. The presence of a fracture disrupts the peripheral contact distribution of the intact joints and appeared to shift stresses posteromedially within the acetabulum.

Figure 7.

Sensitivity of computed contact stresses on projected cartilage thickness and cartilage modulus values. All models are of Specimen 2 at heel strike, with the Tekscan measured data shown in the box on the upper left. Contact stress maps on the left are from models with 1 mm cartilage projections and variable Young's Modulus for cartilage. Contact stress maps along the top are for variable thickness cartilage projections and a constant 8 MPa Young's modulus. The data shown in yellow result from the modeling parameters that were validated.

Table 1

Comparative contact area, peak contact stress, and stress correlation data for the intact and malreduced posterior wall fracture experiments. HS – heel strike; MS – mid stance; TO – toe off.

Specimen	State	Position	Contact Area (mm ²)				Peak Contact Stress (MPa)			Stress Correlation	
			Tekscan	Specimen-Specific Cartilage DEA	Projected Cartilage DEA	Tekscan	Specimen-Specific Cartilage DEA	Projected Cartilage DEA	Specimen-Specific Cartilage DEA	Projected Cartilage DEA	Stress Correlation
1	Intact	HS	1562	808	1174	2.8	3.5	2.0	0.95	0.97	
2	Intact	HS	928	651	616	5.7	5.9	6.1	0.95	0.93	
2	Intact	MS	667	419	455	6.8	6.4	6.5	0.94	0.93	
2	Intact	TO	515	421	522	5.1	5.5	5.5	0.99	0.94	
1	Fractured	HS	821	528	580	4.7	5.1	4.4	0.94	0.97	
2	Fractured	HS	627	426	250	11.3	9.7	11.3	0.96	0.88	
2	Fractured	MS	652	421	274	7.1	6.0	6.7	0.91	0.90	
2	Fractured	TO	587	496	330	6.9	6.4	7.7	0.95	0.88	