Computationally Efficient Magnetic Resonance Imaging Based Surface Contact Modeling as a Tool to Evaluate Joint Injuries and Outcomes of Surgical Interventions Compared to Finite Element Modeling

Joshua E. Johnson

Department of Mechanical Engineering,
University of Kansas,
3138 Learned Hall,
Lawrence, KS 66045
e-mail: a2joe@ku.edu

Phil Lee

Hoglund Brain Imaging Center,
University of Kansas Medical Center,
3901 Rainbow Boulevard,
Kansas City, KS 66160
e-mail: plee2@kumc.edu

Terence E. McIlff

Department of Orthopedic Surgery,
University of Kansas Medical Center,
3901 Rainbow Boulevard,
Kansas City, KS 66160
e-mail: tmciff@kumc.edu

E. Bruce Toby

Department of Orthopedic Surgery,
University of Kansas Medical Center,
3901 Rainbow Boulevard,
Kansas City, KS 66160
e-mail: btoby@kumc.edu

Kenneth J. Fischer¹

Department of Mechanical Engineering,
Department of Orthopedic Surgery,
1530 W. 15th St.,
3138 Learned Hall,
Lawrence, Kansas 66045
e-mail: fischer@ku.edu

Joint injuries and the resulting posttraumatic osteoarthritis (OA) are a significant problem. There is still a need for tools to evaluate joint injuries, their effect on joint mechanics, and the relationship between altered mechanics and OA. Better understanding of injuries and their relationship to OA may aid in the development or refinement of treatment methods. This may be partially achieved by monitoring changes in joint mechanics that are a direct consequence of injury. Techniques such as image-based finite element modeling can provide in vivo joint mechanics data but can also be laborious and computationally expensive. Alternate modeling techniques that can provide similar results in a computationally efficient manner are an attractive prospect. It is likely possible to estimate risk of OA due to injury from surface contact mechanics data alone. The objective of this study was to compare joint contact mechanics from image-based surface contact modeling (SCM) and finite element modeling (FEM) in normal, injured (scapholunate ligament tear), and surgically repaired radiocarpal joints. Since FEM is accepted as the gold standard to evaluate joint contact stresses, our assumption was that results obtained using this method would accurately represent the true value. Magnetic resonance images (MRI) of the normal, injured, and postoperative wrists of three subjects were acquired when relaxed and during functional grasp. Surface and volumetric models of the radiolunate and radioscapoid articulations were constructed from the relaxed images for SCM and FEM analyses, respectively. Kinematic boundary conditions were acquired from image registration between the relaxed and grasp images. For the SCM technique, a linear contact relationship was used to estimate contact outcomes based on interactions of the rigid articular surfaces in contact. For FEM, a pressure-overclosure relationship was used to estimate outcomes based on deformable body contact interactions. The SCM technique was able to evaluate variations in contact outcomes arising from scapholunate ligament injury and also the effects of surgical repair, with similar accuracy to the FEM gold standard. At least 80% of contact forces, peak contact pressures, mean contact pressures and contact areas from SCM were within 10 N, 0.5 MPa, 0.2 MPa, and 15 mm², respectively, of the results from FEM, regardless of the state of the wrist. Depending on the application, the MRI-based SCM technique has the potential to provide clinically relevant subject-specific results in a computationally efficient manner compared to FEM. [DOI: 10.1115/1.4026485]

Keywords: finite element modeling, surface-based contact modeling, magnetic resonance imaging, joint mechanics, ligament injury

Introduction

Computational modeling is very useful in biomechanics to simulate normal and pathologic joint function. It is also useful to determine the efficacies of various surgical procedures performed

to treat joint pathologies and simulate their outcomes. Models can be used to estimate in situ measures such as contact pressure distributions that are difficult to acquire through experiments noninvasively. Currently, computational modeling is the only technique available to noninvasively evaluate in vivo joint contact mechanics [1]. However, most models make use of input parameters derived from various general sources such as literature, standards, or experiments and are, therefore, limited for patient-specific applications [2]. Joint injuries, whether ligaments or articular surface, are a significant problem and there is still a need for tools to

¹Corresponding author.

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effectively evaluate joint injuries and associated sequelae [3]. The ability to monitor the initiation and progression of joint instability after injury may aid in determining prognosis, leading to better treatment algorithms. In order to refine or develop treatments that are targeted toward individuals, it is important to focus on subject-specific models.

Several modeling techniques exist to evaluate in vivo joint mechanics. The common techniques include image-based FEM [4–13], rigid body spring modeling/discrete element analysis (RBSM) [14–16], or SCM [17–19]. The models are either displacement driven or force driven. Generally, model geometries are acquired from modalities such as computed tomography (CT) [4–8,14,15,19] or MRI [9–11,17,18]. Kinematics are determined through external (surface markers) or internal (biplanar radiography) measures, while tendon forces are estimated from corresponding musculature electromyography (EMG) and cross-sectional area, and ground reaction forces are measured using force platforms [11,20]. These loads and displacement boundary conditions are input into the model to infer joint kinetics/kinematics and resulting surface and/or volumetric stresses and strains.

FEM is the most common and accurate method to determine stresses [11,20,21]. However, depending on the complexity of the problem, the process of developing the mesh can be laborious and obtaining a converged solution can be computationally intensive [22], which limits its clinical applicability. Depending on the type of problem (for instance, deformable versus rigid), more simplified analyses can be performed based on relevant assumptions to determine appropriate solutions. This is the basis of RBSM and SCM techniques. Using these methods, joint mechanics can be evaluated in a computationally efficient manner compared to FEM [23], which makes them relevant for clinical applications. The underlying question is whether these methods are competent to provide data that are sufficiently accurate for the intended application.

The ability to accurately determine joint mechanics has wide clinical implications, especially in complex joints such as the wrist. It may be possible to sufficiently evaluate changes in joint mechanics as a result of injury or surgical intervention from surface contact mechanics data alone. This can be achieved through the SCM technique, without the need for a complex volumetric analysis [24]. However, the SCM technique has not been extensively used for orthopedic applications.

Computational modeling has been extensively applied to the lower extremity to evaluate in vivo joint mechanics [4–11,18]. In the wrist, studies have evaluated in vivo joint mechanics during functional activities [12,14,16,17] and have also simulated the effects of some carpal fractures and limited fusions [13,15,19]. Scapholunate (SL) ligament injury is a commonly occurring wrist ligament injury that can lead to SL joint instability and progressive degenerative changes [25–28]. Prior modeling work on the in vivo effects of SL ligament injury or surgical repair appears to be limited [29,30].

Hence, we investigated differences in radiocarpal in vivo joint mechanics obtained from SCM for normal wrists, after SL ligament injury, and after surgical repair to results from the FEM “gold standard.” We did not intend to make comparisons between the normal, injured, and postoperative states. Our goal was to show that contact outcomes obtained from SCM would be comparable to those obtained from a similar FEM analysis regardless of wrist state and to demonstrate the feasibility and applicability of the SCM technique.

Methods

Three human subjects, diagnosed with unilateral SL ligament injury by a board certified physician, were enrolled for the study. Protocols were approved by the local Institutional Review Board. The subjects were all male with a mean age of 36 years (55, 23, and 31 for subjects 1, 2, and 3, respectively). Preoperative images were first acquired of the injured and contralateral (normal) wrists.

The injured wrists were imaged again 12 to 15 weeks post surgical repair (postoperative).

MR images were acquired using a constructive interference steady state (CISS) or a dual echo steady (DESS) state sequence. High resolution *frontal plane* images were first acquired with the wrists in a relaxed position (unloaded) at 0.15×0.15 mm in-plane pixels for CISS sequence (field of view (FOV) = 95×61.8 mm, slice thickness = 0.5 mm, number of slices = 104, repetition time (TR) = 14.8 ms, echo time (TE) = 7.4 ms, flip angle = 60 deg, number of average = 1) and 0.22×0.2 mm in-plane pixels for DESS sequence (FOV = 100×48 mm, slice thickness = 0.5 mm, number of slices = 110, TR = 30 ms, TE = 5.7 ms, flip angle = 25 deg, number of average = 1). Considering the small size of the bones and comparatively thin cartilage, the in-plane resolution and slice thickness gave us very accurate models. Then, images were acquired during active light grasp (loaded) at a lower resolution (CISS: voxel = 0.3×0.3 mm, slice thickness = 1.0 mm, number of slices = 52, TR = 12 ms, TE = 6 ms and DESS: voxel = 0.31×0.31 mm, FOV = 100×100 mm, slice thickness = 1.0 mm, number of slices = 52). The loaded scans were acquired at lower resolution to reduce scan time. This minimized subject fatigue and related motion artifacts during scanning, while maintaining adequate resolution for the modeling process. The grasp task involved gripping to a specified target of 22 KPa (3.125 psi) with the help of a visual feedback system. This target grip was maintained for the duration of scan. The wrists were also braced in 20 deg extension (neutral radioulnar deviation with forearm pronated) for the grasp activity to maintain a consistent loaded position between wrists.

Both the SCM and FEM techniques required model geometries, kinematics (displacement driven), and a contact relationship. Geometries and kinematics were acquired for the normal, injured, and postoperative wrists of each subject. Figure 1 shows a schematic of the modeling process. For SCM, the radius and carpal bones (scaphoid and lunate) together with their cartilage surfaces were segmented from the unloaded images using ScanIP[®] (Simpleware, Exeter, UK). These segmentations were used to generate surface models of the radiocarpal articulations with triangular elements (Fig. 2 left).

The bones were assumed as rigid, that is, undergoing negligible deformation under the relatively light loading activity. Hence, for FEM (Fig. 2 right), only cartilage segmentations with similar surfaces as SCM were used to generate four-node tetrahedral meshed volumetric models using ScanFE[®] (Simpleware, Exeter, UK). Lunate, scaphoid, and radius mesh refinement resulted in a range of 10,805, 9691, and 17,740 nodes, respectively, to 19,301, 18,286, and 28,635 nodes, based on convergence analysis.

Kinematics were acquired from normalized mutual image volume intensity registrations between the unloaded and loaded image volumes with subvoxel accuracy using Analyze 5.0[®] (Analyze Direct, Overland Park, KS). Since the bones were assumed as rigid, only the bone tissue of the radius and carpal bones was segmented from the two image sets. Briefly, the radius was used as a fixed reference to align the loaded and unloaded image sets. This reference transformation was used to reposition the loaded carpal bones into the unloaded coordinate system. Then, carpal bone transformations were determined between the unloaded carpal bones and repositioned loaded carpal bones, both in the same unloaded image coordinate system. This final transformation placed the unloaded carpal bone models in the functionally loaded position. Since the models were displacement driven, the final loaded pose was used to determine contact outcomes. As such the initial configuration of the surfaces was not taken into consideration and should not influence the contact results.

Surface models of the radiocarpal articulations with the final kinematics were implemented in the SCM software Joint_Model [24]. A uniform thickness of 1 mm was assigned to each articular surface based on typical measured values. Uniform thickness was

Fig. 1 Schematic showing the magnetic resonance imaging (MRI)-based surface contact modeling (SCM) and finite element modeling (FEM) processes

Fig. 2 Radiocarpal surface (left) and volumetric (right) models of the normal wrist of subject 2 used for surface contact modeling (bone and cartilage geometry) and finite element modeling (only cartilage geometry), respectively. Shown from a dorsal perspective.

assumed to simplify the analysis and to reduce number of modeling steps (eliminating the determination of spatially varying subject-specific cartilage thicknesses). Though the thickness was uniform for analyses, the model articular surface was the actual anatomical surface segmented from the images. Since the loaded images were acquired over a relatively long period of time, the cartilage was assumed to have had sufficient time to relax or approach equilibrium. Hence, an effective compressive relaxation modulus of 4 MPa was assigned to the articular surfaces [24]. Contact measures were determined from interactions of the rigid surfaces in contact, based on a linear contact rule. Contact area was calculated from overlapping nodes of the surface elements in contact. The depth of interpenetration/overlap was used to

determine a first order estimate of normal compressive strain at each node in contact, based on the thicknesses of the cartilage surfaces. These “deformations” were distributed to the nodes on each surface, based on the ratios of their thickness and modulus values, which were the same for each surface in contact. Nodal contact pressures were determined from the local strains and material properties. Contact pressures were numerically integrated over the entire contact region to obtain the total compressive contact force on the articulation.

Volumetric models of the radiocarpal articulations together with kinematics were also implemented in the FEM software Abaqus® (Simulia, Providence, RI), on a Windows® personal computer (PC) with 12 processors. Since the cartilage was

Fig. 3 Contact pressure distributions (MPa) of the normal (N), injured (I), and postoperative (P) wrists of subject 1 from radioscaphoid (a) and radiolunate (b) articulations. For comparison, surface contact modeling (SCM) and finite element modeling (FEM) results are shown side by side. For surface contact modeling the scale varies linearly from white (minimum) to dark red (maximum). For maximum values see Table 2.

assumed to have substantially relaxed, time- and flow-independent material characteristics were assumed. Hence, the cartilage was modeled as an elastic, homogeneous, and isotropic deformable solid. Similarly, the value of effective modulus was the same as for the SCM analysis. Due to the dual phase interstitial fluid-solid matrix interaction, the instantaneous response of articular cartilage is to behave as an incompressible material, where the hydrostatic forces support the loads applied to the joint surface. With time, fluid efflux and matrix compaction result in cartilage starting to become more compressible. Hence, Poisson's ratio $\nu = 0.20$ was chosen to model the long-term response of the hydrated cartilage matrix [31–33]. The subchondral nodes of the cartilage volumes were rigidly constrained by the kinematic boundary conditions. The linear penalty constraint method was used to enforce the pressure-overclosure relationship between the surfaces in contact. Frictionless contact was assumed between the articular surfaces (neglecting tangential behavior) and a finite sliding formulation was used to model the normal interaction between the bodies. Contact areas, contact forces, peak contact pressures, and pressure distributions were determined. The solutions were accepted as converged after a less than 5% variation in contact force with mesh refinement.

To verify modeling accuracy, contact area results from the simulations were compared to contact areas directly measured from the loaded MR images [17,23]. The regions where articular surfaces were visibly in contact were segmented from each image. The lengths of these contact arcs were multiplied by slice thickness and summed to determine the effective "direct" contact areas for each articulation. Contact forces, contact areas, peak and mean contact pressures, and contact pressure distributions were compared between the SCM and FEM techniques for the radiolunate (RL) and radioscaphoid (RS) articulations from normal, injured, and postoperative wrists.

Results

Qualitatively, the location of contact pressure distribution matched well between SCM and FEM for the normal, injured, and postoperative wrists of all subjects (Figs. 3–5). This was expected since the same kinematics were implemented in both modeling approaches.

Due to the low number of subjects, no tests for statistical significance were performed. Absolute differences between SCM and FEM contact outcomes for RL and RS articulations of each

subject were compared for all three states, resulting in 18 total comparisons per measure. Absolute differences between SCM and FEM contact outcomes for both articulations averaged across subjects for each state were also compared, resulting in six total comparisons per measure.

Table 1 shows a comparison of contact forces between SCM and FEM and also the value of the difference for the normal, injured, and postoperative wrists of all subjects. For all subjects, absolute differences were within 10 N for 15 out of the 18 results. Taking the average across the three subjects, absolute differences of RL and RS contact force in all wrist states were all within 6 N (Fig. 6). The overall average (wrist states and articulations) signed difference between SCM and FEM was 1 N. Contact force signed differences for normal, injured, and postoperative wrists were similar between SCM and FEM for the radiolunate articulation (Fig. 6).

Table 2 shows comparison of peak contact pressures between SCM and FEM for the normal, injured, and postoperative wrists of all subjects. The absolute differences were within 0.5 MPa for 13 out of the 18 results. The differences of the average of peak contact pressures across the subjects between SCM and FEM were under 0.5 MPa (five out of six results), except for radioscaphoid postoperative wrists (Fig. 7). The overall average signed difference between SCM and FEM showed SCM peak pressure to be 0.34 MPa lower than FEM. For the radiolunate articulation, signed differences were similar between SCM and FEM for normal, injured, and postoperative wrists (Fig. 7).

Table 3 shows mean contact pressures compared between the SCM and FEM techniques for the normal, injured, and postoperative wrists of all subjects. The absolute differences were within 0.2 MPa for 15 out of the 18 results and under 0.5 MPa for all results. Except for radioscaphoid postoperative wrists (Fig. 8), the differences of the average of mean contact pressures across the three subjects were within 0.2 MPa (five out of six results). The overall average signed difference between SCM and FEM showed SCM mean pressure to be 0.14 MPa lower than FEM. Signed differences in mean contact pressures were similar between SCM and FEM for normal, injured, and postoperative wrists in both articulations (Fig. 8).

Table 4 shows comparison of normal, injured, and postoperative contact areas between SCM and FEM and also direct contact areas for all subjects. For 15 out of the 18 results, absolute differences between SCM and FEM were within 15 mm². Taking the average of contact areas across the three subjects (Fig. 9), the difference between SCM and FEM was over 15 mm² only for

Fig. 4 Contact pressure distributions (MPa) of the normal (N), injured (I), and postoperative (P) wrists of subject 2 from radioscaphoid (a) and radiolunate (b) articulations. For comparison, surface contact modeling (SCM) and finite element modeling (FEM) results are shown side by side. For surface contact modeling the scale varies linearly from white (minimum) to dark red (maximum). For maximum values see Table 2.

Fig. 5 Contact pressure distributions (MPa) of the normal (N), injured (I), and postoperative (P) wrists of subject 3 from radioscaphoid (a) and radiolunate (b) articulations. For comparison, surface contact modeling (SCM) and finite element modeling (FEM) results are shown side by side. For surface contact modeling the scale varies linearly from white (minimum) to dark red (maximum). For maximum values see Table 2.

Table 1 Comparison of radiolunate (RL) and radioscaphoid (RS) contact forces (N) from MRI-based surface contact modeling (SCM) and finite element modeling (FEM) and differences (DIFF) between SCM and FEM for normal, injured, and postoperative wrists of all three subjects

		Normal			Injured			Postoperative		
		SCM	FEM	DIFF	SCM	FEM	DIFF	SCM	FEM	DIFF
RL	1	13	16	3	50	27	-23	7	9	2
	2	46	53	7	39	41	2	9	10	1
	3	6	5	-1	9	12	3	16	19	3
RS	1	22	13	-9	34	10	-24	12	13	1
	2	15	12	-3	3	5	2	5	5	0
	3	11	17	6	11	18	7	15	26	11

Fig. 6 Average contact forces across the three subjects from surface contact modeling (SCM) and finite element modeling (FEM) for the three states

Table 2. Comparison of radiolunate (RL) and radioscapoid (RS) peak contact pressures (MPa) from MRI-based surface contact modeling (SCM) and finite element modeling (FEM) and differences (DIFF) between SCM and FEM for normal, injured, and postoperative wrists of all three subjects

		Normal			Injured			Postoperative		
		SCM	FEM	DIFF	SCM	FEM	DIFF	SCM	FEM	DIFF
RL	1	0.88	1.17	0.29	1.39	1.12	-0.27	0.42	0.77	0.35
	2	0.91	1.23	0.32	1.30	1.42	0.12	0.73	1.09	0.36
	3	0.72	0.85	0.13	0.98	1.15	0.17	0.79	1.14	0.35
RS	1	1.15	1.24	0.09	2.18	2.20	0.02	1.59	2.55	0.96
	2	1.04	1.27	0.23	0.54	1.09	0.55	0.66	1.17	0.51
	3	0.69	1.11	0.42	0.87	1.48	0.61	0.60	1.42	0.82

Fig. 7 Average peak contact pressures across the three subjects from surface contact modeling (SCM) and finite element modeling (FEM) for the three states

Table 3 Comparison of radiolunate (RL) and radioscapoid (RS) mean contact pressures (MPa) from MRI-based surface contact modeling (SCM) and finite element modeling (FEM) and differences (DIFF) between SCM and FEM for normal, injured, and postoperative wrists of all three subjects

		Normal			Injured			Postoperative		
		SCM	FEM	DIFF	SCM	FEM	DIFF	SCM	FEM	DIFF
RL	1	0.36	0.48	0.12	0.62	0.56	-0.06	0.20	0.33	0.13
	2	0.46	0.63	0.17	0.48	0.56	0.08	0.21	0.31	0.10
	3	0.34	0.44	0.10	0.35	0.45	0.10	0.35	0.50	0.15
RS	1	0.40	0.48	0.08	0.55	0.57	0.02	0.83	1.30	0.47
	2	0.40	0.43	0.03	0.33	0.52	0.19	0.23	0.39	0.16
	3	0.20	0.40	0.20	0.30	0.60	0.30	0.25	0.50	0.25

radioscaphoid normal and injured wrists (four out of six results). The overall average signed difference between SCM and FEM showed FEM contact area to be 12 mm² lower than SCM. Signed differences for normal, injured, and postoperative wrists observed between SCM and FEM were of similar magnitude only in the radiolunate articulation (Fig. 9). Except for radioscapoid postoperative wrists, average contact areas from SCM were within 15 mm² of the average direct contact area results (Fig. 9). SCM and FEM contact areas tended to match direct contact areas more closely for the radiolunate articulation (Fig. 9).

Discussion

We compared radiocarpal contact mechanics as a result of scapholunate dissociation and subsequent surgical intervention determined using the SCM technique to results of a parallel analysis

Fig. 8 Average mean contact pressures across the three subjects from surface contact modeling (SCM) and finite element modeling (FEM) for the three states

Table 4 Comparison of radiolunate (RL) and radioscapoid (RS) contact areas (mm²) from MRI-based surface contact modeling (SCM) and finite element modeling (FEM) and differences (DIFF) between SCM and FEM, and also direct contact area measurements (DA) for normal, injured, and postoperative wrists of all three subjects

		Normal				Injured				Postoperative			
		SCM	FEM	DA	DIFF	SCM	FEM	DA	DIFF	SCM	FEM	DA	DIFF
RL	1	37	33	40	-4	79	49	87	-30	34	27	31	-7
	2	99	85	99	-14	81	72	119	-9	42	31	60	-11
	3	18	11	17	-7	25	28	19	3	45	38	54	-7
RS	1	57	26	62	-31	62	18	49	-44	15	10	65	-5
	2	38	28	25	-10	10	10	35	0	22	14	50	-8
	3	57	43	77	-14	38	30	64	-8	59	53	71	-6

Fig. 9 Average contact areas across the three subjects from surface contact modeling (SCM) and finite element modeling (FEM) and also from direct contact area (DA) measurements for the three states

using FEM. At least 80% of contact forces, peak contact pressures, mean contact pressures, and contact areas from SCM were within reasonable limits (10 N, 0.5 MPa, 0.2 MPa, and 15 mm², respectively) of the results from FEM, suggesting viability of the SCM approach. Change in wrist state did not appear to affect SCM accuracy compared to FEM, further affirming the SCM technique.

Mechanical factors are considered important in the etiology of initiation and progression of OA, and peak contact pressures and pressure distributions are considered to be key mechanical factors [34]. Increase in peak contact pressures above a physiological threshold may cause cellular damage, leading to the inability of

cartilage to repair and maintain a healthy extracellular matrix. Peak contact pressures tended to increase with injury, particularly in the radiolunate articulation. Peak contact pressure results from SCM and FEM were in the range reported by prior *in vitro* studies [23,35,36]. Also, the nonuniform/asymmetrical contact distributions observed from both techniques are typical of the incongruent radiocarpal articulations [35]. While radioscapoid peak contact pressure appeared to decrease after injury in some instances, a separation in the location of peak contact pressure and distribution between scaphoid and lunate contact was also observed (Fig. 4). Scapholunate diastasis is clinically observed after ligament disruption with increasing severity of instability [37], and a subsequent shift in contact may also be a potential OA risk factor. Similar trends between SCM and FEM peak contact pressures for normal, injured, and postoperative wrists suggested that SCM accuracy did not appear to be affected by change in wrist state.

Since the measures (based on averages) that were over the criteria were only radioscapoid measures, higher differences in contact outcomes between SCM and FEM observed in the radioscapoid articulation may be due to geometry and kinematics specific to this location. Scaphoid contact was mainly observed at the dorsal ridge, particularly in the injured and postoperative wrists (Figs. 3–5). There are rapid changes in geometry associated with this region, where the radius contact surface transitions from relatively flat to areas of high curvature. As such, deformable interaction of the FEM technique may more accurately model the contact in this region compared to rigid surface interaction of SCM. Also, the lower mesh resolution of the SCM models (Fig. 2 left) may have caused an averaging effect, especially in the regions of high curvature. This may have caused an underestimation of SCM nodal strains (less overlap), resulting in lower estimates of peak and mean contact pressures. Further work is needed to conclusively state whether transition areas are a limitation of the SCM method or whether the averaging effect from using lower resolution models is a factor.

FEM contact area trends corresponded more closely to direct contact area measurements than those from SCM (Fig. 9). This was expected as the volumetric models were deformable, which would represent the apparent deformations of the articular surfaces on the loaded MR images from which the direct contact areas were calculated, while contact areas from SCM were determined from overlapping nodes of the interacting rigid bodies. Lower magnitudes of FEM contact areas may be due to the rigid bone assumption for model definition as shown by previous studies [38]. Direct contact area measurements were generally higher than the model measurements. The reliability of direct contact area measurements has been previously shown using cadaveric specimens [23]. With *in vivo* data, it is possible that direct contact area is being overestimated. For both articulations, SCM and FEM contact areas for the normal wrists most closely matched the direct contact area measurements. With injury, the effects of motion artifacts on the loaded images may be more significant, thereby influencing segmentation accuracy and potentially affecting direct contact area measurements. Larger differences observed in the radioscapoid articulation between model and direct contact area results (Fig. 9) could be a result of the scaphoid being more mobile than the lunate [39]. Currently, direct contact area is the only measurement available to non-invasively verify our *in vivo* modeling accuracy.

Since its introduction, the application of finite element modeling has progressed rapidly in the biomechanics field. The applications are no longer just limited to basic tissue stress analysis. They now include complex multiphasic interactions, biomechanical models that simulate physiologic/pathologic conditions, tissue response at implant interfaces, osteoporotic fracture predictions, etc. [40]. The extent of model complexity is dictated by the level of detail required of the output variables. However, as models get more complicated, as in the case of a nonlinear contact problem, more time and resources are required to prepare the models and complete the analysis [22], which questions their feasibility for

clinical applications. Alternate modeling techniques that can produce similar results to the FEM gold standard could potentially provide clinically relevant results in an efficient manner.

With regards to computational efficiency, SCM and FEM models were both constructed from the same segmented surfaces. As such, the time for performing image segmentation for both procedures was essentially the same. Generating triangular element surfaces from the segmentations for SCM required only a few minutes (less than a minute per surface). However, generating volumetric meshes ensuring high fidelity surface geometry, element shape, and a well-converged meshes for both cartilage surfaces required several hours. On average, processing times for FEM were 58 min and 85 min for radiolunate and radioscapoid contacts, respectively, the longest being over 200 min. The processing times for SCM were approximately 5 s per articulation.

MRI-based surface contact modeling has significant potential as a research and clinical tool. While CT is a common imaging modality, MRI is believed to be the best method to obtain subject-specific geometry [20], though longer acquisition times are a limitation. Osseous and soft tissue geometries can be acquired from the same set of images without the need for coordinate system registration when using multiple imaging modalities or the use of contrast agents. With MRI-based SCM, the actual cartilage surface can be segmented, thereby eliminating the need to replace articular surfaces with a mechanical equivalent such as discrete springs [14–16]. It also eliminates the need for approximating cartilage surfaces using half the distance between bone geometry assuming articular contact, which may not be the case during imaging. Studies have shown these methods to have lower accuracy in predicting contact mechanics [22,38]. Inclusion of joint space with cartilage thickness and additional assumptions at joint edges such as extra thickness are, thus, avoided. Using fat-suppressed T2-weighted sequences like DESS minimizes chemical shift artifacts, providing better delineation between tissue interfaces and allowing for semiautomatic segmentation of the bone, which improves model generation efficiency [41]. Also, with quality control, boundary segmentation errors can be reduced to less than 8% [42]. There is a risk of radiation exposure in using radiographic methods to acquire kinematics [43], such as multiplanar X-ray fluoroscopy techniques. Though this technique provides accurate real-time 3D motion, it is difficult to apply effectively to complex joints with multiple small articulating bones such as the wrist, and accuracy in such applications is not yet verified. Kinematic boundary conditions can be determined from image registration of static/quasi-static images (kinematic MRI) or from dynamic images (cine-PC MRI) acquired during functional loading [2,17], which do not require *a priori* force assumptions. Intensity-based volume registration is less sensitive to segmentation errors [44]. Loading conditions estimated from muscle geometry determined using diffusion tensor MRI [2] and muscle activity from EMG can also be applied to SCM for force-driven models [24].

One limitation was that FEM contact results were not directly validated with experiments. The MRI-based SCM technique has been sufficiently validated to provide reliable joint contact mechanics data [23,45]. Since FEM is accepted as the gold standard, it was used as the standard for comparison. As mentioned earlier, both SCM and FEM results were in reasonable agreement with published data and somewhat in agreement with direct contact area measurements. The low number of subjects was another limitation of the study. While data from three subjects provided a reasonable comparison between SCM and FEM, they were not sufficient (nor intended to be so) to make conclusions on the effects of injury and surgical repair. Using the SCM technique, prior *in vivo* work in our lab has shown significant increases in all contact outcomes in the radioscapoid articulation as a result of injury [29] and a significant decrease in contact pressures in the radiolunate articulation after surgical repair [30]. Another limitation of the SCM modeling approach, as implemented was the uniform thickness assumption for the articular surfaces to simplify

the analysis. Total articular thickness of the contacting surfaces was used to approximate local strains. Assigning higher thickness values to the cartilage surfaces can lead to lower strain estimates and, subsequently, underestimate contact pressure results. Variable thickness cartilage models can also be implemented using SCM. Also, SCM mesh resolutions were comparatively lower than FEM due to limitations of the current modeling software version and the fact that it requires a closed surface model. Lower geometric fidelity may have resulted in an averaging of the nodal contact pressures from a coarser mesh distribution. To further improve SCM accuracy, future work would include comparing results from higher resolution models and also from assigning subject-specific spatially varying cartilage thicknesses determined from the MR images.

The use of absolute error instead of relative error for comparison could also be viewed as a limitation of the study. However, the disadvantage of using relative or percent error in joints such as the wrist where the magnitudes of contact outcomes are substantially lower than weight bearing joints (i.e., hip or knee) is that small absolute values can result in very large relative errors. We have previously shown that using a relative value can exaggerate the error when the absolute value is small, which may not provide a better comparison [23]. Since the number of subjects was too low to perform statistical analyses, we used “cut-off” criteria for comparing the SCM and FEM methods. The 10 N, 0.5 MPa, 0.2 MPa, and 15 mm² criteria for comparison of contact force, peak contact pressure, mean contact pressure, and contact area, respectively, were based on magnitudes of contact outcomes observed in the wrist. In our prior validation of the SCM method, we used a criterion of model results within two standard deviations of the experimental error, that is, a difference within 25% [23]. The values chosen in this study were well within 25% of the corresponding maximum experimental measurement during simulated grasp and were, therefore, considered reasonable. While at least 80% of contact outcomes were within the criteria, the goal is to develop a tool to evaluate joint injuries and surgical outcomes in an efficient manner for clinical applications. This is possible if the method is able to differentiate between changes in contact mechanics and contact patterns that occur due to injury or surgical repair. SCM prediction of contact outcomes were found to be consistent across normal or abnormal states in this study, demonstrating that SCM can differentiate between changes as a result of injury or surgical repair. With injury, changes in joint mechanics can vary up to a 100% [23] and we have recently shown differences in contact forces and contact areas as a result of scapholunate ligament injury to be almost twice that of the criteria chosen for this study [29]. In this context, our results indicate that the SCM method should be able to identify the majority of moderate to severe injuries. When the differences are closer to the threshold, as would be the case with minimal instability, then there is the possibility of failing to identify one out of five injuries. However, since diagnosing soft tissue injuries like ligament injuries in their early stages is already a challenge, with many missed diagnoses, the ability to identify four out of five mild to moderate injury cases would still provide benefit.

MRI-based surface contact modeling can provide useful in vivo joint contact mechanics data in a computationally efficient manner compared to FEM, making the technique more clinically feasible. Some differences between SCM and FEM results could also be attributed to the specific implementation of the FEM analysis. For instance, studies have shown a rigid subchondral interface assumption, used in the FEM analysis, to predict higher contact pressures than a deformable bone-cartilage interface [38]. While more complex and time consuming, FEM can be used to perform detailed analyses to investigate the influence of depth-dependent anisotropy on stresses or the effects of fluid layer between articulating surfaces on contact mechanics. Thus, FEM is likely the tool of choice for most research studies. In this study, we demonstrated the feasibility and applicability of the SCM method and illustrated the qualitative and quantitative similarity between SCM and

FEM. With further development of the SCM method and inclusion of more subjects for statistical comparisons, we believe more definitive conclusions can be made about the accuracy of the method. Results from this study indicate MRI-based SCM to be a practical tool that can efficiently obtain clinically relevant data, which can differentiate between normal and abnormal conditions and potentially assess OA risk based on contact mechanics data [29,30].

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