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Validation of radiocarpal joint contact models based on images from a clinical MRI scanner

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Abstract

This study was undertaken to assess MRI-based radiocarpal surface contact models of functional loading in a clinical MRI scanner for future *in vivo* studies, by comparison with experimental measures from three cadaver forearm specimens. Experimental data was acquired using a Tekscan sensor during simulated light grasp. MR images were used to obtain model geometry and kinematics (image registration). Peak and average contact pressures, contact forces and contact areas were determined in the radiolunate and radioscapoid joints. Contact area was also measured directly from MR images acquired with load and compared to model data. Based on the validation criteria (within 25% of experimental data), out of the six articulations (three specimens with two articulations each), two met the criterion for average contact pressure (0%, 14%); one for peak pressure (20%); one for contact force (5%); four for contact area with respect to experiment (8%, 13%, 19%, 23%), and three contact areas met the criterion with respect to direct measurements (14%, 21%, 21%). Absolute differences between model and experimental peak contact pressures were reasonably low (within 2.5 MPa). Overall, the results indicate that MRI-based models generated from 3T clinical MR scanner appear sufficient to obtain clinically relevant data.

Keywords

magnetic resonance imaging; contact modeling; validation; wrist; radiocarpal joint

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1. Introduction

More than 20 million Americans are estimated to suffer from osteoarthritis (OA) (Buckwalter and Martin 2006). Developing tools for prevention and early detection of OA is an important research area. The underlying cause of OA is not well defined and is likely multifaceted. Excessive articular surface contact stress is believed to be a predominant factor (Buckwalter and Martin 2006). Similarly, abnormal kinematics will lead to changes in normal cartilage stresses and strains (Andriacchi et al. 2004). Patient-specific tools for OA risk assessment using a combination of *in vivo* imaging and computational modeling could help clinicians mitigate problems before OA develops.

Several techniques exist for joint contact analysis using computational models. Model geometry is often acquired using either computed tomography (CT) (Beek et al. 2001; Donahue et al. 2002; Carrigan et al. 2003; Anderson A et al. 2005; Anderson D et al. 2005; Koolstra and van Eijden 2005; Anderson et al. 2007) or magnetic resonance imaging (MRI) scans (Roarty and Grosland 2004; Cheng et al. 2007; Ezquerro et al. 2007; Troy and Grabiner 2007). Modeling approaches include rigid body spring modeling (RBSM) (Genda et al. 1995; Tsumura et al. 1998; Genda and Horii 2000; Manal et al. 2002; Fregly et al. 2003; Fregly et al. 2005) and finite element modeling (FEM) (Anderson and Daniel 1995; Li et al. 1999; Buchler et al. 2002; Penrose et al. 2002; Gardiner and Weiss 2003; Taddei et al. 2004; Mesfar and Shirazi-Adl 2005; Li et al. 2006; Ahn and DiAngelo 2007). A third technique is multi body contact modeling, which is similar to RBSM, but considers full interaction between three-dimensional (3D) surface models. Solutions are determined from interpenetration of contact surfaces as an estimate of cartilage deformation and solved to satisfy specific force and/or kinematic boundary conditions (Kwak et al. 2000). Surface contact modeling is computationally efficient compared to FEM and has more flexible contact rules than RBSM (Li et al. 1997; Manal et al. 2002).

Contact mechanics from surface models based on *in vivo* MRI could help predict OA risk. The feasibility of this approach for joint contact analyses of active *in vivo* functional loading has been previously evaluated (Pillai et al. 2007; Thoomukuntla et al. 2008; Fischer et al. 2011). Contact mechanics data such as articulation forces, areas and pressure distributions can be determined using this technique.

Prior to use of computational models for evaluation of patient-specific *in vivo* joint contact mechanics, the approach must be validated. Different techniques have been used for validation, including comparison with previously published experimental or analytical investigations (Brown 2004). While this method may establish general confidence, a more definitive validation can be provided by comparison with specimen-specific experimental measures (Song et al. 2004; Pfaeffle et al. 2005; Anderson et al. 2007; Anderson et al. 2008; Papaioannou et al. 2008; Shim et al. 2008). Generally, validation is based on how closely the experimental and model data agree. Validation criteria vary, and too often validation criteria are never quantitatively defined or there is no attempt whatsoever at experimental confirmation (Brand 2005). One criterion for validation is a model error of less than two times the accuracy (standard deviation) of the experimental measure for comparison

(Henninger et al. 2010), which assures that model data are within the 95% confidence level of the experimental measure.

The objective of this study was to validate MRI-based modeling of functional wrist loading in a clinical MRI scanner by quantitative comparison of model data with specimen-specific experimental data. Protocols matched those planned for future evaluation of *in vivo* joint mechanics, as the purpose was to establish the reliability of such data.

2. Methodology

To validate data from the MRI-based joint contact models, radiocarpal contact model results were compared with specimen-specific experimental measurements from an electronic pressure sensor. Results of experimental accuracy studies conducted in our lab have shown a wide range of errors in experimental measures, typically at about 12% (Fischer et al. 2011). Based on this and the aforementioned criterion [model within twice the experimental error (Henninger et al. 2010)], modeling error within 25% of experimental error was selected for contact force, contact area, and peak and average contact pressure validation.

2.1. Specimen preparation and experimental setup

Three cadaveric forearm specimens with no indication of prior forearm injury or pathology (as determined by screening radiographs) were used for this study. Age and gender were generally unknown, but were not considered critical to the validation study, for which the results were not meant to be generalized. For each specimen, dissection involved isolation of the radius and ulna for mounting the forearm and preservation of tendons in the mid-forearm for specimen loading. The elbow was left intact. The wrist was also dissected volarly to open the radiocarpal joint capsule for sensor insertion. Care was taken to preserve the flexor tendons and minimize disruption of the volar wrist capsule.

To simulate muscle loads, we used the approach of Fischer et al. (2011) whereby the flexor digitorum superficialis (FDS), the flexor digitorum profundus (FDP) and flexor pollicis longus (FPL) were loaded to simulate grasp. Two extensors, extensor carpi ulnaris (ECU) and extensor carpi radialis (ECR), were also loaded, providing resistance to wrist flexion during the simulation of light grasp (Figure 1). Due to similar proximity and function, the ECR longus and ECR brevis were sutured together to form a single ECR tendon unit. Suture loops and nylon lines were also added to the tendons, in order to attach the loading weights. Tendons were wrapped in saline soaked gauze to prevent drying and loss of function and care was taken to leave the interosseous membrane intact as studies have shown its importance for wrist joint stability (Kapandji 2001).

The load distribution between tendons was roughly proportional to approximate relative muscle cross-sectional area, and extensor loads were set to consistently resist wrist flexion (Fischer et al. 2011). During both experimental measurements and MRI acquisition, loads of 30 N were first applied to both the ECR and ECU, then 20 N was applied to both the FDP and FDS, followed by 10 N to the FPL, for a total of 110 N. Before load application, a cylindrical pipe was placed in the hand as an object for simulated grasp. Fingers were loosely taped around the pipe to maintain finger position even when no tendon loads were

applied. The forearms were mounted in neutral rotation for testing and imaging. Due to release of the volar capsule and stabilizing ligaments, experimental conditions did not necessarily replicate physiological grasp. Instead, it was most important that the magnitude and manner of loading were the same for each measurement and also for image acquisition, to make the specimen-specific comparison of results more meaningful.

2.2. Experimental pressure measurements

This study made use of the electronic K-Scan™ Wrist Sensor #4201 (Tekscan Inc., South Boston, MA), due to its high sensitivity and especially the dynamic real-time measurements (Wilson et al. 2003). The dynamic measurement of the Tekscan system eliminates insertion/extraction artifacts, which can occur when using dye-based pressure sensitive films (Kazuki et al. 1991).

To better fit within the radiocarpal joint space and minimally disrupt the volar ligaments and capsule, the sensor was trimmed approximately 2 mm on the lateral and medial sides. The calibration and testing protocol were similar to prior work (Thoomukuntla et al. 2008; Fischer et al. 2011). Briefly, the Tekscan sensor was calibrated for force under a series (10 levels) of known loads using a materials testing machine (Instron, Model #5811, Instron Corp., Canton, MA) immediately prior to specimen testing. For testing, the sensor was inserted into the radiocarpal joint and the joint was subsequently loaded in simulated grasp, as described earlier. No sensor readings were observed prior to loading and pressure measurements were taken immediately after the loads were applied and the forearm was stable. Three loading trials were conducted for each specimen to assure that the loading was consistent and to minimize the possibility of experimental error. Contact pressures were interpolated based on measured raw values and the local sensel pressures and area (3.63 mm^2 per sensel) were used to determine contact forces and contact areas. While some measures were fairly consistent, measurement variability was also commonly observed within the individual specimen trials. Hence, the average of the three trials was taken for each specimen, which was also helpful to compare with the corresponding single specimen-specific model parameters.

2.3. MR data acquisition and model construction

MR images were acquired using a 3T clinical MRI scanner (Allegra, Siemens Medical Solutions USA, Malvern, PA). For each specimen, frontal plane images were acquired without tendon loading (matrix = 640×416 , field of view [FOV] = $95 \times 61.8 \text{ mm}^2$, slice thickness = 0.5 mm, number of slices = 104, repetition time [TR] = 14.8 ms, echo time [TE] = 7.4 ms, flip angle = 60° , number of average = 1) and during simulated light grasp (loaded tendons, matrix = 320×208 , slice thickness = 1.0 mm, number of slices = 52, TR = 12 ms, TE = 6 ms) using a constructive interference steady state (CISS) pulse sequence and a custom-made double loop surface coil that closely conformed to the wrist (Figure 2). The high resolutions scans without tendon loading were used to obtain accurate undeformed model geometry. For the loaded configuration, the image resolution was reduced (to half that of the unloaded scans) to shorten scan time (to 196 seconds), which is planned for *in vivo* studies with active grasp.

Unloaded and loaded radius and carpal bones (without cartilage surfaces) were manually segmented and isolated on a black background (Figure 3) in preparation for image registration. Manual segmentation was used in all cases, because the image contrast and signal-to-noise ratio was not high enough to allow for automatic or semi-automatic segmentation. The basis of kinematic analysis was the assumption that bone tissue deforms negligibly during light grasp, while cartilage deforms even under the low loading. Analyze 5.0 (Analyze Direct, Overland Park, KS) was used to perform registration and a combination of manual and auto 3D voxel registrations was used to obtain the best fit. For this study, the radius was selected as the reference (fixed) bone. The loaded radius was registered to the unloaded radius, to determine the kinematic transformation from the loaded to the unloaded image coordinate system (to align image sets). This transformation was then used to transform the loaded carpal bones into the unloaded image coordinate system. Lastly, the unloaded carpal bones were registered to match the (transformed) loaded carpal bones. The final transformations were used to place the bone models constructed from the unloaded state in the proper loaded configuration for contact analysis.

Manual segmentation of radius and carpal bones (including their respective cartilage surfaces) was performed on the unloaded image sets, and Geomagic Studio 9 (Raindrop Geomagic, Research Triangle Park, NC) was used to construct 3D surface models of the unloaded bones (Figure 4).

The kinematics from image registration were applied to the surface models in the Joint_Model software (Kwak et al. 2000) to determine the contact mechanics. The Joint_Model program determined contact pressures from interpenetration of body (cartilage) surfaces with a defined contact relationship. This allowed contact area and location, and first order estimate of cartilage strain to be determined directly from the geometry and loaded configuration of the bones. A linear contact rule was used for analysis and therefore an effective relaxation modulus of 4 MPa was assigned to the cartilage of each bone (Kwak et al. 2000). Cartilage was also assigned a uniform thickness of 1 mm for each surface (total thickness of 2 mm for each articulation) based on average measured values from the images. The program determined the solution for the specified boundary conditions and provided contact mechanics data for each defined bone contact pair. Local cartilage strain was estimated from the local interpenetration and the total thickness of the contacting pairs. Local contact pressure was linearly proportional to the strain at any point on the cartilage surface and the pressure distribution was computed over the contact area. Contact pressures were integrated over each contact area to obtain contact force. Average contact pressure was determined from the ratio of contact force to contact area. These contact parameters were compared to the experimental measures and the contact area was also compared to direct measurements from the loaded-state MR images.

2.4. Direct contact area measurement

Contact areas were determined directly from MR images as an additional form of validation for measurements obtained from models. Radiocarpal contact was reasonably visible in the loaded-state MR images, as regions where gaps between the articular surfaces were not easily distinguishable. These regions of contact were segmented (Figure 5). The arc of

contact was defined for each contact pair in each image throughout the entire grasp image set. Contact arc lengths were multiplied by slice thickness to get an effective area, and the areas from all images were summed to determine total contact area for each contact pair.

3. Results

Qualitatively, relative size, intensity and location of scaphoid and lunate contact on the radius corresponded between models and experimental data. Consistent mid to dorsal contact was observed on the radius fossa from both the experiment and model (Figure 6). Quantitatively, model results predicted contact parameters that were similar to experimental measures. Tables 1 and 2 show a summary of all experimental, model and image-measured data across specimens.

Comparing peak pressures between Tekscan and model data, results obtained were not very consistent (Table 1). Tekscan and model peak contact pressures closely matched for Specimen 3 radiolunate (RL) and radioscapoid (RS) contacts (within 0.5 MPa). On the other hand, Tekscan measured high peak contact pressures for Specimen 1 RL contact and Specimen 2 RS contact—both over 3.5 MPa. Otherwise, peak Tekscan contact pressures fell within a range of about 1–3 MPa. The trend was for Tekscan to measure higher peak contact pressure values than were found using the model. Peak model contact pressures were all under 2 MPa. Model average contact pressures matched within 0.3 MPa of Tekscan measures for five out of the six articulations (Table 1).

Tekscan and model results matched reasonably well (within 12 N) for contact forces of four out of six articulations (Table 2). The general trend was for the model to predict lower RL contact forces compared to RS contact forces. Model contact force was relatively low for Specimen 2 RL contact. Since measurements were only made in the radiocarpal joints, and the applied tendon loads are also transferred through ulnocarpal joints, the sum of the Tekscan and Model contact forces is expected to be less than the 110 N applied during simulated grasp. Except for Specimen 2 Tekscan measures, this is evident from both the experimental and model results (Figure 7).

Comparing model contact area results to Tekscan and direct measurements (Figure 8), it was observed that the three matched closely for Specimen 1 RL and Specimen 2 RS contacts (within 9 mm² and 12 mm² respectively) and were reasonably close for Specimen 2 RL and Specimen 3 RS contacts. Tekscan experimental results closely matched direct contact area results for four out of the six articulations (within 8%, 13%, 19%, and 23%).

To evaluate model validation, Table 3 shows specimen-specific percent differences between experimental and model measurements. Peak pressure met the validation criterion for Specimen 3 RS contact. Average contact pressure met the validation criterion for both articulations of Specimen 2. Contact force difference was very small for Specimen 2 RS contact and met the 25% difference criterion. Contact area met the validation criterion for both articulations of Specimen 2 and for RL contact in Specimen 1, with respect to direct image measurements. Contact area met the validation criterion for Specimen 1 RL and Specimen 2 RS contacts and both articulations of Specimen 3, compared to Tekscan data. No other measurements met the specimen-specific validation criteria.

4. Discussion

Overall, MRI-based models yielded results that were quite reasonable qualitatively and quantitatively. Prior *in vivo* and experimental studies indicated most of the contact to be in the dorsal region of the radius (Pillai et al. 2007; Thoomukuntla et al. 2008; Fischer et al. 2011) and we consistently observed mid to dorsal contact.

Contact area was the most consistent and reliable model measure, compared to Tekscan and direct image measurements, since contact area resulted from direct surface interaction. Model contact area results that were within 25% of the direct measurements; the same direct measurements were also within 25% of the Tekscan measurements. The reliability of the direct contact area measurement has also been shown in previous validation studies (Thoomukuntla et al. 2008; Fischer et al. 2011). The outcome of direct measurements can depend on image quality, and identifying the contact regions required care, especially with the lower resolution loaded images. Still, the direct contact area measurements from loaded MRI provide a means to compare model results when other parameters are not easy to verify. For instance, direct measurements of contact area would also be available from images in future *in vivo* studies. Since direct MR contact area closely matched Tekscan contact area for four out of the six articulations, it appears to be robust for use in human subjects studies and can therefore be used to verify *in vivo* modeling accuracy. The specimens that had good contact area matches similarly had good contact force matches. As it would be possible to verify *in vivo* contact area accuracy with direct contact area measurements, accuracy in contact area would suggest accuracy in contact force measurements as well. Unfortunately, accurate contact area does not appear to guarantee accurate peak pressure data. Peak contact pressure is highly dependent on model geometries and kinematic details, and thus shows higher variability.

Model peak contact pressure results fell in the generally expected range (1–3 MPa), per prior *ex vivo* studies (Hara et al. 1992). Peak contact pressures over 3.5 MPa for Tekscan do not appear to be accurate compared to prior studies of light grasp (Thoomukuntla et al. 2008; Fischer et al. 2011). High experimental variation contributes to the difficulty in achieving validation of the models. Specimen 3 had the best peak pressure match with the largest difference of about 0.5 MPa. When absolute values are small and measurement error is relatively large, the percent error may appear quite high. For peak pressure especially, a reasonable absolute difference error may be a better alternative validation criterion. To some extent these observations indicate the arbitrariness of our validation criterion based on percent error versus absolute error (for example, the percent difference between experimental and model average contact pressure values for Specimen 3 RS contact was 204%, while the absolute difference was only 0.3 MPa). With injury, changes in contact pressure distribution on the articular surface are to be expected. Changes in peak contact pressures as a result of injury, can perhaps easily vary from a 50% increase to possibly as high as 100% (Short et al. 1995; Ledoux et al. 2001; Anderson et al. 2005). Thus, the current validation criterion may also be more stringent than necessary to distinguish differences between normal and injured contact mechanics.

The Tekscan sensor appeared to provide acceptable results for the majority of measurements, especially for contact area. Only the sum of contact forces for Specimen 2 (117 N for total contact force) shows clear measurement error, as this sum is higher than the applied load. Due to the presence of a sensor in the joint space, articulations will have been somewhat altered due to sensor thickness and stiffness (Wu et al. 1998). Thickness and stiffness of the sensor affect conformity between articulating surfaces and this could have impacted the model-experimental comparisons. The effect of sensor presence is most likely to affect peak pressure measurements.

Currently, there is no gold standard to validate computational models (Henninger et al. 2010). Tekscan measurements are also prone to variability within and across specimen trials. This variability may be present no matter how many specimens are tested. Even so, experiments can provide a reasonable idea of model performance. The reliability of direct measurements may indicate its potential as a gold standard especially for *in vivo* studies under functional loading. While statistics can be helpful to evaluate overall mean behavior, a better level of model performance may be achieved by comparing specimen-specific parameters individually. This study was not intended to draw conclusions of anatomical/physiological behavior from model results, but rather to demonstrate the feasibility of the model to evaluate *in vivo* joint mechanics. In such studies, statistics will be used to draw conclusions on *in vivo* behavior and function (for instance, normal versus injured) and minimize the effect of variability in the results, knowing that these results are reliable.

Results obtained from MRI-based models are comparable to results from similar radiocarpal contact analysis studies conducted in the past (Thoomukuntla et al. 2008; Fischer et al. 2011). The significant difference between this study and prior studies is the disparity in image resolution between unloaded and loaded MR images. Lower resolution loaded images in this study may have reduced the accuracy of kinematic registrations. Prior validation studies have acquired both unloaded *and* loaded images in high resolution (Thoomukuntla et al. 2008; Fischer et al. 2011). However, longer scan times were required to obtain these high resolution images. This would not be feasible for human subjects studies with active grasp, as the long scan time would increase the possibility of subject fatigue and the potential for motion artifacts in the loaded images. Hence, the scan times were appropriately reduced (lowering the resolution) while trying to maintain sufficient image quality. So the lower resolution used for loaded images could reduce kinematic accuracy providing a possible (or partial) explanation for some of the un-validated model results. Size and shape of the bone may also be a factor in model accuracy, since smaller bones (like the lunate) are more difficult to segment precisely, and also have fewer distinct features for image registration. This could further contribute to geometric and kinematic errors as illustrated by the overall lower percent differences for RS compared to RL contact.

The CISS imaging sequence is a T2 weighted gradient echo sequence. It was primarily selected for its high spatial resolution and signal to noise ratio. For this study the radiocarpal joint capsule was opened prior to imaging for experimental measurements. This could have lead to air entering the joint space causing MR susceptibility artifacts, which would be further emphasized by high-field gradient echo acquisition (Yang et al. 2011). The CISS sequence also displays signal from fat. These may have consequently lead to a poorer

delineation of articular surface boundaries, decreasing segmentation accuracy. Using a fat suppressed imaging sequence for future *in vivo* studies may help improve geometric model accuracy.

Also, this study used a uniform thickness of 1 mm based on typical cartilage thickness seen on our MRI scans. Overestimation of cartilage thickness will result in lower strain values and therefore, lower model contact pressure and force values. One way to improve future accuracy would be to use patient specific cartilage measurements. It is possible to measure cartilage thickness from the individual high resolution MR images. The thickness could either be averaged to obtain a single specimen-specific value, or variable cartilage thickness could be incorporated in the model. Either approach could improve future analyses. Further, changes errors in elastic modulus will affect the contact pressure distribution and contact force. In addition, a linear model may cause an underestimation of the peak contact pressure. However, the same kinematics and geometry could also be implemented in a deformable contact FEM analysis, including the non-linear material properties of articular cartilage.

While the current MRI-based models of active functional loading are clearly not perfect, they demonstrate substantial potential for the approach. Therefore, further development work is warranted, as well as preliminary work for the next phase with *in vivo* measurements in injured and normal joints.

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Figure 1.
Attachment of the radius and ulna to mounting plate with isolation of tendons and sectioning of the radiocarpal joint capsule. Sutured loops and nylon line connections also shown.

Figure 2.
Custom double loop surface coil designed specifically for wrist imaging with close proximity to dorsal and volar surfaces.

Figure 3. Illustration of the isolation procedure (Specimen 3 Radius) for image registration. The radius (similarly the carpal bones) was isolated from the original MR image (left).

Figure 4.
Sample radiocarpal contact model shown in typical position and orientation in the loaded configuration; (A) lateral, (B) oblique and (C) medial views in Joint_Model.

Figure 5.

Example of segmentation of arc of contact (red line on the right) for the radioscapoid region indicating contact for Specimen 3. The unloaded image (left) showed a gap between the radius and scaphoid, which were then in contact in the loaded image (center). The regions of contact were segmented from all loaded images indicating contact (right), to obtain the effective contact area.

Figure 6.

Intensity profiles from (A) Tekscan pressure sensor and (B) MRI-based model for Specimen 1 (S1), Specimen 2 (S2) and Specimen 3 (S3). Shown with dorsal at the top, lunate fossa to the left (L) and scaphoid fossa to the right (R). For model, the color is linearly scaled from light (minimum) to dark red (maximum = 2.0 MPa) and for Tekscan, the scale varies from dark blue (minimum) to light (maximum = 4.5 MPa).

Figure 7.

Comparison of total radiocarpal contact forces for all specimens (S1–S3). Except for S2 Tekscan measurements, total force for both Tekscan and model measures fell within the 110 N applied force (indicated by the horizontal red line) during simulated grasp.

Figure 8.

Comparison of radiolunate (RL) and radioscapoid (RS) contact areas for all specimens (S1–S3). Model data matched well with Tekscan and direct MR measures in most cases, while Tekscan and direct MR measures matched well for Specimen 1 and Specimen 2.

Table 1

Comparison of experimental Tekscan data (T) and MRI-based contact model data (M) for radiolunate (RL) and radioscapoid (RS) peak contact pressures (PP) and average contact pressures (AP) for all specimens.

Specimen	PP (MPa)				AP (MPa)			
	RL		RS		RL		RS	
	T	M	T	M	T	M	T	M
1	3.7	1.1	2.6	1.2	0.8	0.4	0.4	0.6
2	2.7	1.6	4.5	2.0	0.6	0.6	0.7	0.6
3	1.9	1.4	1.0	1.2	0.5	0.3	0.1	0.4

Table 2

Comparison of experimental Tekscan data (T), MRI-based contact model data (M) and direct MRI measurement data (D) for radiolunate (RL) and radioscapoid (RS) contact force and contact areas for all specimens.

Specimen	Contact Force (N)				Contact Area (mm ²)					
	RL		RS		RL		RS			
	T	M	T	M	T	M	T	M	D	
1	32	21	39	27	40	49	43	105	46	90
2	75	45	42	44	131	79	100	60	68	56
3	24	15	15	37	52	48	107	109	88	63

Table 3

Percentage comparison of experimental, MRI-based contact models and direct MRI measurements for radiolunate (RL) and radioscapoid (RS) peak contact pressures (PP), average contact pressures (AP), contact areas and forces (T-M: Tekscan vs. Model; D-M: Direct vs. Model).

Specimen	Contact	PP (MPa)		AP (MPa)		Force (N)		Contact Area (mm ²)	
		T-M	RL	T-M	RS	T-M	RL	T-M	D-M
1	RL	70.1	46.3	33.5	23.8	13.9			
	RS	53.0	59.2	29.8	55.9	48.4			
2	RL	41.6	0.4	39.7	39.5	21.1			
	RS	54.6	6.9	4.4	12.1	21.4			
3	RL	29.6	33.4	39.1	8.6	55.4			
	RS	20.1	204.0	146.1	19.0	39.0			