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Factors Influencing Cartilage Thickness Measurements with Multi-Detector CT: A Phantom Study

Andrew E. Anderson, BS^{*}, Benjamin J. Ellis, BS^{*}, Christopher L. Peters, MD⁺, and Jeffrey A. Weiss, PhD^{*,+}

^{*}Department of Bioengineering, and Scientific Computing and Imaging Institute, University of Utah, 50 S. Central Campus Drive, Rm 2480, Salt Lake City, UT 84112

⁺Department of Orthopedics, University of Utah, 590 Wakara Way, Salt Lake City, UT 84108

Abstract

PURPOSE—To prospectively assess in a phantom the reconstruction errors and detection limits of cartilage thickness measurements from MDCT arthrography as a function of contrast agent concentration, imaging plane, spatial resolution, joint space and tube current, using known measurements as the reference standard.

MATERIALS AND METHODS—A phantom with nine chambers was manufactured. Each chamber had a nylon cylinder encased by sleeves of aluminum and polycarbonate to simulate trabecular bone, cortical bone, and cartilage. Variations in simulated cartilage thickness and joint space were assessed. The phantom was scanned with and without contrast agent on three separate days, with chamber axes both perpendicular and parallel to the scanner axis. Images were reconstructed at intervals of both 1.0 and 0.5 mm. Contrast agent concentration and tube current were varied. Simulated cartilage thickness was determined from image segmentation. Root mean squared and mean residual errors were used to characterize the measurements. CT scanner and image segmentation reproducibility were determined.

RESULTS—Simulated cartilage was reconstructed with < 10% error for thicknesses >1.0 mm when no contrast agent or a low concentration of contrast agent (25%) was used. Errors grew as concentration of contrast agent increased. Decreasing the simulated joint space to 0.5 mm caused slight increases in error; below 0.5 mm errors grew substantially. Measurements from anisotropic image data had errors greater than those for isotropic data. Altering tube current did not affect reconstruction errors.

CONCLUSION—Our study establishes lower bounds and repeatability of simulated cartilage thickness measurement using MDCT arthrography, and provides data pertinent to choosing contrast agent concentration, joint spacing, scanning plane, and spatial resolution to reduce reconstruction errors.

Keywords

CT arthrography; phantom; reconstruction error; cartilage; thickness

INTRODUCTION

Recent evidence suggests that multi-detector CT (MDCT) arthrography may be more sensitive than MRI for detecting cartilaginous lesions (1–5) and quantifying cartilage thickness (6), although fat-suppressed spoiled gradient-echo in the steady state (FS-SPGR) is still considered the best imaging protocol for imaging articular cartilage (7–16). While a substantial body of research has examined MRI cartilage reconstruction errors (e.g. (17–22)), less attention has been given to CT arthrography (6,18,23). Nevertheless, estimates of cartilage thickness determined via MRI image data are often validated by direct comparison with CT arthrography results (18,19), which may erroneously imply that CT arthrography is the reference standard for such estimations.

Studies that have compared cartilage thickness measurements estimated from CT arthrography data to those obtained from physical measurements of anatomical sections have generally been qualitative assessments (18,23). To our knowledge only one study compared quantitative measurements of cartilage thickness between reconstructed MDCT arthrography images to excised tissue samples (6). However, the use of harvested cartilage plugs in this study limited the range of cartilage thickness that could be analyzed (6).

Recent evidence suggests that cartilage may actually swell in the early stages of OA (24). It would be useful, therefore, to quantify cartilage thickness using CT arthrography in patients who complain of pain that may be related to OA but do not have direct evidence of radiographic thinning or localized defects. From the point of view of experimental investigations of cartilage contact mechanics using cadaveric tissues, quantification of differences in reconstruction errors between standard CT and CT arthrography would clarify whether cadaveric joints should be completely dissected and imaged with air or if the joint capsule should be left intact.

The bounds of cartilage thickness detection and hence the ultimate reconstruction error remains unknown for MDCT arthrography. In addition, the influence of imaging parameters on the ability to detect and reconstruct articular cartilage from MDCT arthrography image data has not been assessed. Thus, the purpose of our study was to prospectively assess in a phantom the reconstruction errors and detection limits of cartilage thickness measurements from MDCT arthrography as a function of contrast agent concentration, imaging plane, spatial resolution, joint space and tube current, using known measurements as the reference standard.

MATERIALS AND METHODS

Phantom Description

An imaging phantom was designed and manufactured to quantify the error in reconstructing cartilage thickness (CNA Precision Machine, Ogden, UT) (Fig. 1). The phantom body was constructed using nylon (Natural Cast Nylon, Professional Plastics Inc., Fullerton, CA). Nine chambers were drilled into the phantom body (Fig. 1). Each chamber was composed of a central nylon cylinder encased by cylindrical sleeves of aluminum and polycarbonate (Standard Polycarbonate, Professional Plastics Inc., Fullerton, CA). The central nylon cylinder simulated trabecular bone, the cylindrical sleeve of aluminum represented cortical bone, and the outer cylindrical sleeve simulated cartilage (Fig. 1, middle). All aluminum cylinders were machined to a wall thickness of 1.00 mm to represent cortical bone with constant thickness. The polycarbonate cylindrical sleeves were machined to wall thickness values of 0.25, 0.50, 0.75, 1.00, 2.00, and 4.00 mm (Phantom Chambers 1-6, Fig. 1, top). An outer polycarbonate four-prong spacer was press-fit into each of the chambers between the outer layer of simulated cartilage and adjacent nylon phantom body (Fig. 1, middle). The spacer held the central cylinders securely in place and provided a “joint space” that could be filled with contrast agent. The joint space in phantom chambers 1-6 (Fig. 1, top) was held constant at 2.0 mm. A varying

joint space (0.25, 0.50, and 1.00 mm) with constant simulated cartilage thickness of 2.00 mm was used in the remaining three compartments (Phantom Chambers 7-9, Fig. 1, top). Finally, nylon threaded caps were used to seal the fluid in the chambers. A micrometer with accuracy of ± 0.01 mm was used by the manufacturer to determine the wall thickness tolerance of the aluminum and polycarbonate cylindrical sleeves, representing cortical bone and cartilage, respectively. The tolerance was reported to be within ± 0.07 mm.

Nylon, polycarbonate and aluminum were chosen because their x-ray attenuation values are similar to trabecular bone, cartilage and cortical bone, respectively (25–28). The size of the phantom (250×250 mm) was representative of a typical field of view (FOV) for imaging human diarthrodial joints. The outer diameter of each compartment (outer boundary of simulated cartilage) was kept constant at 52 mm while the diameter of the aluminum sleeve and central nylon cylinder were adjusted between 38–46 mm to accommodate differences in cartilage thickness and joint spacing. This range of cylinder diameters is similar to that reported in the literature for human femoral and humeral heads (29–31). The range of cartilage thickness (0.25–4.00 mm) was chosen to represent the range reported in the literature for human articular cartilage (32,33).

CT Imaging Protocol

All phantom scans were performed with a Siemens SOMATOM® Sensation 64 CT Scanner (Siemens Medical Solutions USA, Malvern, PA). This scanner makes use of a periodic motion of the focal spot in the longitudinal direction to double the number of simultaneously acquired slices with the goal of attaining improved spatial resolution and elimination of spiral artifacts regardless of spiral pitch. Constant scanning parameters for our study were: 120 kVp, 512×512 matrix, 300 mm FOV, and 1 mm slice thickness. A total of 6 contrast enhanced scans and 4 non-enhanced scans were performed. The imaging protocol detailed below was performed on three separate days to assess the reproducibility of the CT scanner and segmentation procedure.

Contrast Enhanced Scans—Contrast agent (Omnipaque 350 mgI/ML, GE Healthcare, Princeton, NJ) was mixed with 1% lidocaine HCL (Hospira Inc., Lake Forest, IL) in separate concentrations of 25, 50, and 75%. The phantom was scanned using a tube current of 200 mAs for each of the three concentrations ($n = 3$ scans) in the “transverse” or frontal plane (Fig. 1, top). The laser guide was used to align the CT slice axis perpendicular to the phantom chambers longitudinal axes, thereby minimizing volumetric averaging between slices. Additional transverse scans were conducted with tube currents of 150 and 250 mAs using the phantom filled with 50% contrast agent ($n = 2$ scans). A scan with tube current of 200 mAs was performed on the phantom filled with 50% contrast agent parallel to the phantom chambers “longitudinal” axes (Fig. 1, top) to intentionally introduce volumetric averaging.

Non-Enhanced Scans—The phantom was scanned without fluid to estimate the error in cartilage thickness reconstruction for disarticulated, dissected cadaveric joints. Non-enhanced scans were performed in the transverse plane using tube currents of 150, 200, and 250 mAs ($n = 3$ scans). A final non-enhanced scan was performed with a tube current of 200 mAs parallel to the phantom chambers longitudinal axes to intentionally introduce volumetric averaging between successive slices.

Image Segmentation, Surface Reconstruction, and Measurement of Thickness

Phantom image data were transferred to a Linux workstation for post-processing. Image data were re-sampled post-CT using 0.5 mm slice intervals for the contrast enhanced and non-enhanced longitudinal scans to assess changes in reconstruction errors between an anisotropic spatial resolution (0.586×1.0×0.586 mm) and near isotropic resolution (0.586×0.5×0.586 mm). Thinner post-scan reconstructions in the transverse plane would have been ambiguous since

the curvature of the phantom chambers did not change as slices were taken through this direction.

Separate splines for the outer surface of the aluminum cylinder, representing cortical bone, and the boundary between the polycarbonate cylinder and air (non-enhanced scan) or contrast agent (contrast enhanced scan), representing the outer layer of simulated cartilage, were extracted from the image data. Both automatic and semi-automatic thresholding techniques were employed using commercial segmentation software (Amira 4.1, Mercury Computer Systems, Chelmsford, MA).

Each dataset was automatically thresholded using a masking technique available in Amira 4.1, which allows the user to highlight pixels over a range of defined intensities. For datasets with contrast agent included, the mask was adjusted incrementally until all of the pixels representing nylon (the bulk of the phantom body) were *excluded*. Thus, pixels with intensities greater than this value were masked as contrast agent and simulated cortical bone whereas values less were defined as simulated cartilage. The same masking procedure was used for the non-enhanced scan datasets to define the simulated cortical bone boundary; however, the boundary between simulated cartilage and air was defined by reversing the mask such that all pixels representing the nylon body of the phantom were *included*. As mentioned above, the masking procedure was performed for each CT dataset separately to ensure that the appropriate threshold range was chosen independently of alterations in tube current, contrast agent concentration, spatial resolution or scanner direction. Following masking of all of the datasets it was later determined that inter-scan threshold values varied by less than 5%.

Due to CT volumetric averaging it was necessary to utilize a semi-automatic thresholding technique for datasets where contrast agent was included. However, this procedure was only required for phantom chambers with simulated cartilage thickness of 0.5 and 0.25 mm (chambers 5 and 6, Fig. 1, top); simulated cartilage thicker than this was effectively segmented by the automatic method, regardless of contrast agent concentration, tube current, spatial resolution or scanner direction. For the 0.5 and 0.25 mm chambers the baseline automatic threshold value was first used to define a general segmentation spline. Next, regions where pixels blended together were separated using a paintbrush tool available in Amira 4.1 such that the resulting spline followed the general boundary between simulated cartilage and contrast agent. Although volumetric averaging was present, the intensity gradient between contrast agent and simulated cartilage was strong enough to allow for easy visual separation. To ensure uniformity, all of the semi-automatic segmentations were performed by the senior author, A.E.A.

Splines were stacked upon one another and triangulated using the Marching Cubes algorithm (34) to form surfaces that represented the outer surfaces of simulated cortical bone and cartilage. To preserve the native splines of the CT image data, the resulting polygonal surfaces were not altered via decimation or smoothing. A published algorithm was used to assign thickness to each of the nodes defining the simulated cartilage surface (35). The algorithm has been tested for accuracy using concentric cylinders with known thickness. Reported errors were less than 2% (35).

Error Analysis

Thickness values were analyzed to determine the reconstruction errors and detection limit of MDCT and to investigate the influence of tube current, joint spacing, contrast agent concentration, and imaging plane. The overall thickness error for each phantom chamber was assessed using the root mean squared (RMS) error criteria:

$$\text{RMS} = \left[\sum_{i=1}^n (t_i^{\text{CT}} - t^{\text{Phantom}})^2 / n \right]^{1/2}, \quad (1)$$

where the summation is over the number of surface nodes n and t^{Phantom} is a constant thickness that was assessed by direct manufacturer measurement of the phantom. The mean residual error was calculated to determine the directionality of the error:

$$\text{Mean Residual} = \sum_{i=1}^n (t_i^{\text{CT}} - t^{\text{Phantom}}) / n. \quad (2)$$

Statistical Analysis

Descriptive statistics were calculated using statistical software (SPSS 11.5 for Windows 2002, SPSS Inc. Chicago, IL). Specifically, RMS and mean residual errors were averaged for the three days that CT scans were conducted. The resulting means were plotted (SigmaPlot 8.0, Systat Software Inc., San Jose, CA) with standard deviation error bars to indicate the inter-scan variation in reconstruction error.

RESULTS

Contrast Enhanced Scans

There were notable differences in the average RMS and mean residual error due to alterations in contrast agent concentration (Fig. 2). The simulated cartilage of the phantom was reconstructed with less than a 10% RMS and mean residual error for thickness greater than 1.0 mm when the lowest concentration of contrast agent (25%) was used and the direction of the CT scan was transverse to the phantom (Fig. 2). Transverse scan RMS errors grew progressively as the concentration was increased from 25% – 75% for values of thickness greater than 0.75 mm (Fig. 2, top). An increase in contrast agent concentration resulted in a greater tendency for simulated cartilage to be underestimated for values between 1.0 and 4.0 mm thick (Fig. 2, bottom). However, a shift in error from under to overestimation occurred as the thickness approached the spatial resolution of the image data (0.586×0.586 mm) (Fig 2, bottom).

Substantial differences in average reconstruction errors were also noted when the scanner direction and spatial resolution were altered (Fig. 3). The anisotropic longitudinal reconstructions at 50% concentration produced RMS and mean residual errors greater than the corresponding transverse and near-isotropic longitudinal dataset at 50% concentration for simulated cartilage thicker than 1.0 mm (Fig. 3). Finally, altering the tube current resulted in negligible differences over the range of simulated cartilage thickness analyzed (data not shown).

There were differences in RMS errors over the range of joint spaces studied due to changes in contrast agent concentration and scanner direction (Fig. 4) but not due to alterations in tube current (data not shown). Errors increased as the concentration of contrast agent was increased (Fig. 4). RMS errors for each individual transverse scan increased slightly when the joint space was decreased from 2.0 – 0.5 mm; however, below 0.5 mm errors grew substantially (Fig. 4). The anisotropic longitudinal dataset (1.0 mm reconstruction) produced greater RMS errors than the corresponding transverse and near-isotropic longitudinal scan datasets (0.5 mm reconstruction) over the full range of joint spaces analyzed (Fig. 4). Mean residual error analysis

indicated that simulated cartilage thickness was underestimated for all datasets and that these errors were the smallest for the transverse 25% scan (data not shown).

Examination of the standard deviation error bars in Figs. 2 – 4 indicated a high level of reproducibility for simulated cartilage between 0.75 – 4.0 mm thick. The standard deviation error bars also did not overlap adjacent results within this range. Standard deviations were much larger for simulated cartilage 0.25 – 0.5 mm thick and error bars overlapped adjacent data points.

Non-Enhanced Scans

Reconstructions of the non-enhanced transverse scan at 200 mAs resulted in RMS errors less than 10% for thickness values greater than 1.0 mm (Fig. 5, top). RMS errors for the nonenhanced scans were within 2% of those reported for the contrast enhanced transverse scans at 25% contrast agent concentration for simulated cartilage 0.75 – 4.0 mm thick. RMS errors grew substantially for simulated cartilage less than 1.0 mm thick; however, the RMS error leveled out between 0.5 – 0.25 mm (Fig. 5, top). The leveling point in the RMS plot aligned well with corresponding points of inflection on the mean residual error plot (Fig. 5, bottom). Therefore, the lack of increase in RMS error at 0.25 mm was due to a shift from an underestimation to overestimation of cartilage thickness. RMS errors for the longitudinal and near-isotropic longitudinal scan datasets were similar to the transverse scan for 2.0 and 4.0 mm thick simulated cartilage; however errors grew substantially below 2.0 mm (Fig. 5, top). Errors for the longitudinal anisotropic scan were substantially greater than the transverse and near-isotropic longitudinal scans for simulated cartilage less than 2.0 mm thick (Fig. 5, top). Altering the tube 20 current from 150 – 250 mAs did not have an appreciable effect on the RMS or mean residual errors in the transverse plane (data not shown).

As with the contrast enhanced scans, standard deviation error bars of the non-enhanced scans were negligible for thicker simulated cartilage (0.75 – 4.0 mm thick) but increased when thickness was decreased below this range. Standard deviation error bars also did not overlap at adjacent data points within this range but did for thicknesses less than 0.75 mm.

DISCUSSION

To our knowledge our study is the first quantify the detection limits of MDCT using a phantom. The simulated cartilage of the phantom was reconstructed with less than 10% RMS and mean residual error for thicknesses greater than 1.0 mm when either no contrast agent or a low concentration of contrast agent (25%) was used. The results of our study also demonstrated that the CT reconstructions errors were dependent on the concentration of contrast agent, imaging plane direction, spatial resolution and to a lesser extent, joint spacing. Alterations in the scanner tube current did not affect simulated cartilage thickness reconstruction errors in the range tested for both contrast enhanced and non-enhanced scans.

Care was taken to control confounding factors in our study. The physical thickness of the phantom was measured to a tolerance of ± 0.07 mm, thus variations in the true thickness of the phantom would not have a substantial influence on the perceived values of phantom thickness as assessed by CT. In addition, separate scans were done whenever a new intervention (e.g. scan direction, tube voltage, and contrast agent concentration) was performed in an effort to isolate these effects. The entire protocol was repeated on separate days and only minor inter-scan variation was noted for simulated cartilage between 0.75 – 4.0 mm for both contrast enhanced and non-enhanced scans. Therefore, any differences noted in reconstruction error within this range were due to the intervention studied rather than from confounding factors such as CT scanner variability.

The results of the contrast enhanced scan reconstructions showed a direct relationship between contrast agent concentration and reconstruction error. An explanation for this finding is as follows: as the concentration was increased larger pixel intensity gradients were established at the boundary between cartilage and contrast agent. This initiated more intense volumetric averaging at this boundary and resulted in a greater tendency for cartilage thickness to be underestimated. However, a shift in error from under to overestimation occurred as the thickness approached the spatial resolution of the image data (0.586×0.586 mm). This was due to the fact that thickness could not drop much below the width of a single pixel without extensive surface decimation and smoothing. Therefore, although CT has been shown to overestimate the thickness of thin structures (25,26), the results of our study demonstrate that the direction of the error is dependent on the concentration of the joint fluid and spatial resolution of the image data when CT arthrography is used.

El-Khoury et al. (6) compared ankle cartilage measurements obtained from MDCT double-contrast arthrography and three-dimensional FS-SPGR MRI to physical measurements of excised plugs from cadaveric ankles (ranging from 1 – 2 mm thick) and found that CT was more accurate than MRI. Differences in segmentation methodology, joint geometry, and arthrography technique (double contrast) between this study and our work make exact comparisons impossible. Nevertheless, El-Khoury's best-fit line of physical plug measurements plotted against MDCT estimates indicated that CT underestimated cartilage thickness by approximately 5% (6), which has the same direction and similar magnitude of error as the 25% concentration agent results of our study over the same range of thickness.

Study Limitations

The results of our study must be interpreted in light of the inherent differences between measurements obtained from a phantom to those taken from experimental studies that use real cartilage specimens. It is well known that articular cartilage exhibits depth and location dependent inhomogeneities in material structure (36–38), and these factors were not part of our study design. In addition, although similar (25–28), small differences will exist between x-ray attenuation values of real tissue to that of the materials used in our phantom. Finally, diarthrodial joints such as the shoulder and hip have spherical geometry but the phantom chambers were cylindrical. Nevertheless, our approach allowed us to eliminate potential confounding factors such as geometry, tissue homogeneity, and measurement technique. In addition, a total of three scans were performed and descriptive statistics were utilized to assess reproducibility, which was a statistical methodology consistent with a phantom study related to MRI slice thickness (39).

For chambers with simulated thicknesses of 0.25 and 0.5 mm it was necessary to use a semi-automatic method to segment simulated cartilage from the contrast enhanced scan datasets. Reconstruction errors for these chambers could have been influenced by user technique. However, the magnitude of the standard deviation error bars for thicknesses within this range were very similar to the non-enhanced scan deviation bars, and a purely automatic segmentation technique was used for the latter datasets. Therefore, it appears that simulated cartilage reconstruction errors for thickness in this range would remain high regardless of the reconstruction technique employed. Thus, caution should be exercised when attempting to make conclusions regarding cartilage that is thin (below 0.75 mm in our phantom study).

The phantom was designed to simulate the interface between cartilage and cortical bone. There was likely some image thinning of polycarbonate due to volumetric averaging between the polycarbonate and adjacent aluminum cylindrical sleeve; however the wall thickness of aluminum was held constant for each phantom cylinder and the thresholding protocol was not biased to changes between phantom chambers or whole datasets. Thus, any errors introduced

would be consistent over all datasets, which would eliminate simulated cortical bone as a confounding factor to our study.

Practical Applications

The results of our study provide minimum bounds for the errors in cartilage thickness measurement using MDCT and provide guidelines for practical use. It must be emphasized that the reported phantom reconstruction errors are likely a best case result since confounding factors were controlled. A lower concentration of contrast agent is likely to reduce the amount of volumetric averaging between actual cartilage and contrast agent since it was shown that higher concentrations caused the simulated cartilage to appear thinner than its reference thickness. In addition, joint spacing should be maximized prior to scanning, which can be done by completely filling the joint capsule with the diluted contrast solution and/or applying traction to the joint. Failure to do so will result in increased errors when the joint space reaches a critical threshold (occurring at 0.5 mm in our phantom study). Finally, CT image reconstructions should be chosen such that isotropic or near-isotropic spatial resolutions are achieved. Fortunately, MDCT (unlike its predecessors) offers the ability to do this without increased radiation dosage to the patient (40).

From a basic science point of the view the following conclusion can be made: assuming that sufficient joint space is maintained and a low contrast agent concentration is used, one could expect similar cartilage reconstruction errors when cadaveric tissues are CT scanned with or without contrast agent since errors were similar between non-enhanced and contrast enhanced scan (25% concentration) datasets. However, given the additional technical challenges of keeping joint fluid within the capsule of a dissected joint, it seems more appropriate to scan the specimen without contrast agent.

In conclusion, the ability to reconstruct simulated cartilage using a phantom and MDCT with and without arthrography is dependent on several factors including the contrast agent concentration, joint spacing, imaging plane, and spatial resolution. An improved understanding of the detection limits of MDCT cartilage reconstructions will assist in the diagnosis of joint pathologies, interpretation of biomechanical models, and design of epidemiological studies aimed to investigate changes in cartilage thickness.

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Figure 1.

Top- schematic of phantom used to assess the detection limits of MDCT in the transverse plane. The longitudinal (L) imaging plane is also shown. Simulated cartilage thicknesses of 4.0, 2.0, 1.0, 0.75, 0.5, and 0.25 mm with constant joint space of 2.0 mm were used in chambers 1-6, respectively. A constant thickness of 2.0 mm with joint spaces of 1.0, 0.5, and 0.25 mm were used in chambers 7-9, respectively. Middle- exploded view of chamber #1 detailing: A) nylon center cylinder to represent trabecular bone, B) 1 mm thick aluminum sleeve to represent cortical bone, C) polycarbonate sleeve to represent cartilage, D) joint space, E) polycarbonate four-pronged spacer for creating the joint space, and F) bulk of the phantom body made using

nylon. Bottom- CT scan of the phantom with contrast agent and inset showing image details of chamber #1. Letter call-outs correspond to the same details provided above.

Figure 2.

Simulated cartilage RMS (top) and mean residual (bottom) reconstruction errors for the transverse contrast enhanced scan datasets as a function of contrast agent concentration. RMS errors grew progressively as the contrast agent concentration increased (for thickness > 0.75 mm). The directionality of the error was dependent on the contrast agent concentration and simulated cartilage thickness.

Figure 3.

Simulated cartilage RMS (top) and mean residual (bottom) reconstruction errors for the transverse contrast enhanced scan datasets at 50% concentration as a function of imaging plane direction and spatial resolution. Errors were greatest for the anisotropic longitudinal data reconstructions. The longitudinal isotropic reconstructions yielded errors more consistent with the transverse scan results.

Figure 4.

Simulated cartilage RMS errors as a function of joint space thickness, contrast agent concentration, imaging plane direction, and spatial resolution. Errors increased as contrast agent concentration increased. Reconstruction errors from the isotropic longitudinal dataset were less than the anisotropic dataset in the same imaging plane.

Figure 5. Simulated cartilage RMS (top) and mean residual (bottom) reconstruction errors for the non-enhanced scan datasets at 200 mAs as a function of imaging plane direction and spatial resolution. RMS errors for the longitudinal isotropic datasets were consistently less than the anisotropic dataset for simulated cartilage less than 2.0 mm thick.