Assessing Measurement Accuracy Near Orthopaedic Materials in Magnetic Resonance Imaging

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Target Audience. Clinicians and scientists with an interest in the distortion effects of commonly used orthopaedic materials in magnetic resonance imaging (MRI).

Purpose. Adverse local tissue reactions from wear debris generated by total joint replacements are a common cause of premature implant failure. The superior soft tissue contrast of MRI allows for monitoring the local response of the synovial tissues; however, MRI is limited by in-plane and through-plane artifact generated by the implant components. The multi-acquisition variable-resonance image combination (MAVRIC) technique minimizes image distortion by combining image datasets acquired at numerous frequency bands offset from the dominant proton frequency (1, 2). We have previously evaluated the in-plane and through-plane artifact reduction capabilities of MAVRIC (3). This study will determine the *accuracy* of artifact reduction between standard of care 2D fast-spin-echo (FSE) images and corresponding MAVRIC images as compared to a gold standard, by using a novel MRI phantom of known dimensions to scan samples of materials frequently used in arthroplasty.

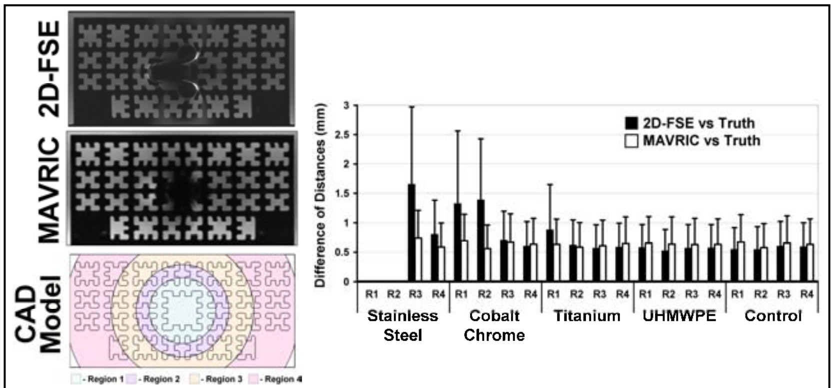

Figure 1. Left - Representative 2D-FSE, and MAVRIC images for a cobalt chrome material sample. The subregions utilized the analysis are shown in the CAD model. Right - Comparison of the difference of distances (mean ± SD) of corner points between 2D-FSE and MAVRIC images with the known phantom dimensions for different material samples.

Methods. A custom phantom composed of a series of polycarbonate grids was designed to hold small prismatic bars of known dimensions of common orthopaedic materials: stainless steel 316, cobalt chrome, titanium, and ultra-high molecular weight polyethylene (UHMWPE). <u>Imaging Protocol</u>: Images were acquired in a clinical 1.5T scanner with an 8 channel cardiac phased array coil (General Electric, Waukesha, WI). The phantom was aligned with the bore of the scanner, placing the material samples orthogonal to B_o. 2D-FSE images were acquired: TE:23ms, TR:3500ms, ETL:21, NEX:4, matrix:512x256, RBW: ±125kHz, voxel volume: 0.41x0.41x3.0 mm^3. MAVRIC images, with the same voxel size and at identical slice locations of the 2D-FSE images were also acquired: TE:40ms, TR:4000ms, ETL:24, NEX:0.5, RBW: ±125kHz (Fig. 1 – Left). Image acquisition was repeated with no material samples, as a Control scan. <u>Image Analysis</u>: A semi-automated program (Matlab, Natick, MA) detected all grid corners in the FSE and MAVRIC images, and the three-dimensional (3D) coordinates of the grid corners were calculated from the DICOM header information. The location of the grid corner points in the MAVRIC and FSE series were compared to the known dimensions, the gold standard, of the phantom in the sagittal plane. The dimensions of the phantom were known from the computer aided design (CAD) model used to design the phantom. A direct calculation of the distance between individual matching points in the CAD vs. MAVRIC, and CAD vs. FSE series was not possible since each image dataset was acquired (FSE and MAVRIC) or existed (CAD) in its own coordinate system. The known relative distances between the corner points in the phantom were available from the phantom CAD model. Therefore, the Euclidian distances matrices were generated for the corner points from the CAD model, and for the corner points from the FSE and MAVRIC scans for each material sample. The data points were further subdivided into 4 subregions defined by their radial distance relative to the center of the phantom: (R1) 0-20mm, (R2) 20-28mm, (R3) 28-43mm, (R4) 43-65mm (Fig.1 – Left). Corner points which were not visible in the individual FSE or MAVRIC series were excluded from the corresponding CAD model dataset. The difference of the distance matrices between the CAD data points and the identified points in the FSE and MAVRIC scans was calculated and summarized as means and standard deviations.

Results. The average difference between the known phantom data points and MAVRIC data points was less than 2 pixels (0.82 mm) for scans of all materials (Fig. 1 - Right). A similar result for the 2D-FSE scan was only found for the Control and UHMWPE sample. The Titanium scan in Region 1 had a difference of 0.88±0.77 mm, and the Cobalt Chrome in Regions 1 and 2 had differences of 1.33±.24 mm and 1.39±1.04 mm, respectively, for the 2D-FSE scans.

Discussion. This study utilized a novel measurement methodology to assess distortion due to materials used in total joint arthroplasty, specifically the accuracy of geometric representation as a function of distance from the implant material. The MAVRIC scans reduced the amount of distortion when in the presence of metal samples, when compared to the known phantom dimensions. The maximum pixel distortion of a MAVRIC image is given by the ratio of one-half of the utilized excitation/refocusing bandwidth to the applied frequency-encoding bandwidth. For the acquisitions in this study, the maximum pixel distortion is approximately 2 pixels. As our results indicate, the maximum distortion of a MAVRIC image is well inside of our measurement precision for the distortion of 2D-FSE images, and the MAVRIC images are effectively ground truth, relative to the systematic deviations of our measurements. It would be beneficial to confirm these findings in the through-plane direction; however, the approximate locations of the reformatted coronal images and the non-uniform cross-sectional geometry of the phantom in the top-down direction prevented the exact dimensions on the CAD model from being determined. Future studies will use the novel phantom to develop additional distortion quantitation techniques and assessment of wear debris from material samples.

Conclusion. MAVRIC produced less distortion than the standard-of-care imaging sequence, and the grid inter-point distances were similar in magnitude to the gold standard of the known dimensions of the CAD model. Understanding the effect of materials on the visualization of surrounding soft tissues is of importance since distortion may over or underestimate the amount of pathology present, obscuring clinically relevant information.

References. 1. Koch KM et al. Magn Reson Med 2011;65:71. 2. Koch KM et al. Magn Reson Med 2009;61:381. 3. Koff MF et al., ISMRM 2012, 2437.

Acknowledgements. Institutional research support was provided by General Electric Healthcare.