DESIGN AND TESTING OF A PHANTOM FOR CALIBRATION OF MRI SYSTEMS USED IN DCE-MRI CLINICAL TRIALS

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Introduction: Dynamic contrast enhanced MRI (DCE-MRI) is an important application for multi-center clinical trials for cancer diagnosis and treatment monitoring [1,2]. For these trials MRI systems must provide estimates of DCE-MRI parameters (e.g. T_1, AUC and K^{trans}) that are consistent and comparable across sites, MRI systems, and scanning sessions. This need motivated us to develop a "DCE-MRI phantom" and imaging protocol with which MRI systems could be evaluated and characterized, such that a DCE-MRI scan on any particular MRI system would yield the same image values and derived DCE-MRI parameters as on another system.

Methods: The DCE-MRI phantom (Fig. 1, left) is built from the outer shell and inner scaffolding of the Magphan Quantitative Imaging Phantom manufactured by Phantom Laboratory, Inc (Salem, NY, USA). The DCE-MRI phantom does not model physical perfusion, but is designed with a 90° rotational symmetry to allow generation of graphs of signal intensity versus contrast concentration at multiple physical locations (Fig 1, middle). The phantom contains one central 5cm diameter sphere and eight 3cm stationary spheres filled with fluid and variable levels of contrast agents (T_1 range: 200-800 ms) in two coronal planes, and 116 small 1cm spheres allow for estimation of image geometric distortion. Each 3cm sphere had D20:H20 in a 1:2 parts mixture, with D20 added to match SNR of the spheres with that from in-vivo DCE-MRI liver scans. For setup on the MRI table, wide straps around two wood boards above and below the phantom support the phase array abdominal coil and ensure that the spatial relationship between phantom and coil is the same in all setups (Fig. 1, right). The protocol for DCE-MRI testing uses 3D fast spoiled gradient recalled echo (FSPGR) sequences with the following parameters: TR=5.6ms, TE=1.2ms, matrix=160x256, BW=31.25kHz, FOV=42cm, phase FOV=0.80, slice thickness=3mm, slices 32. FSPGR was run with six different flip angles (FA 3°, 6°, 9°, 15°, 24°, 35°) and six repetitions for each of the four rotational orientations of the phantom. With images at each rotational orientation, image values can then be sorted to provide graphs of signal versus contrast concentration for each physical location of the 3cm spheres. Four clinical imaging sites (labeled A,B,C,D) participated in the multi-site testing. Sites A-C completed two scanning sessions separated by one week. The MRI systems at the four sites were Site A: 1.5T Signa 8-ch Excite HD OS Ver. 12.0, Site B and C: 1.5T Signa 8-channel Excite HD, OS Ver. 11.0, Site D: Signa NV/I, OS Ver 9.1 (GE Healthcare, Waukesha, WI). Image analysis consisted of automated placement of ROIs centered on the 5-cm and 3-cm diameter contrast spheres. Image values were recorded as a function of rotational orientation, flip angles and repetitions. T_1 and M_0 maps were estimated from the images acquired with the different FAs.

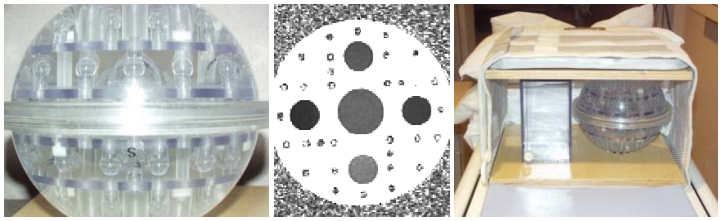

Fig. 1: DCE-MRI phantom, Coronal T_1 map, Setup on MR table

Results: Processing of images from the four sites yielded the following conclusions for all MRI systems: 1) Short term (within-scan) stability was outstanding, while longer term stability (within session and intersession) was significantly less and a concern for statistical comparison. 2) At each sphere location, signal increased with increasing contrast concentration as expected, however, rates of signal increase differed significantly on different MRI systems and sessions (Fig 2, left). 3) T_1 estimates decreased with increasing contrast concentration as expected, however, T_1 estimates differed significantly on different systems and sessions (Fig. 2, right).

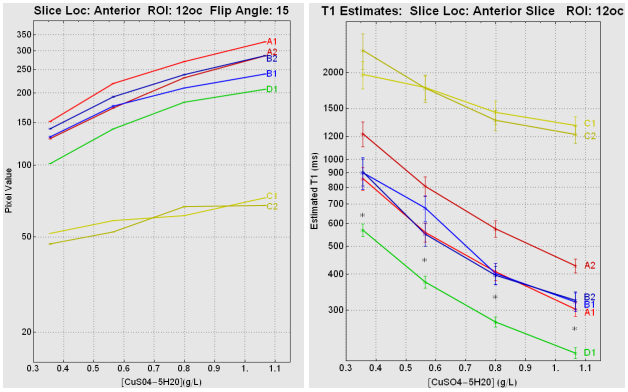

Fig 2: Example signal and T_1 vs. contrast concentration curves from sites A-D.

Discussion: The study revealed data inconsistencies that were unexpected (Fig. 2): In Site A, there was a significant difference in T_1 estimates between scanning sessions 1 and 2 (red lines labeled A1 and A2, respectively). In Site B, there were differences in T_1 estimates between sessions in only a single scan. In Site C, the curve fits were poor and T_1 estimates high (e.g. 2000ms), reflecting inconsistent image values with FA.

Conclusions: This study presents a new DCE-MRI phantom designed for calibration of MRI systems to be used in multi-site clinical trials. Preliminary results at four clinical sites show the ability of the phantom to reveal critical similarities but also expected and unexpected differences in the images and derived DCE-MRI parameters.

References:1. Leach MO et.al. Br J Cancer 2005 May; 92(9): 1599-1610. 2. Evelhoch J et.al. Cancer Res 2005 Aug; 65, 7041-7044.
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