Contrast-Agent Concentration Quantification during First-Pass MRA using Susceptibility-Induced Magnetic Field Shifts

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SYNOPSIS

Paramagnetic contrast agents (CA) modify tissue susceptibility and distort magnetic fields in proportion to their concentration. The frequency shift is described by $\Delta f=[CA]\chi_m f_0 \beta$ where χ_m is the CA molar susceptibility and β a spatially varying shape factor[1,2] which can be calculated provided that the shape as well as its orientation with respect to B_0 are known. Thus [CA] can be determined using an inversion method. Here, [CA] was measured continuously in the aortic arch during first-pass MRA. Cardiac output was measured and compared to phase-contrast (PC) evaluating the technique ability to derive physiological parameters.

METHODS

Imaging protocol: Six volunteers and 1 patient were imaged at 1.5T with a multiphase 2D spoiled gradient-echo sequence ECG gated to acquire one diastole-image per heartbeat in the plane of the aortic arch during First-pass of a Gd-DTPA bolus injection (Magnevist). Single doses were injected at a flow rate of 2.5 to 3 ml/s, followed by a 20-ml saline flush. Imaging parameters included: FOV=30-40 cm, phase FOV 0.5-0.7, 128 readout size, slice-thickness=8 mm, BW=25kHz with full-echo readouts, TR/TE/flip=2.2/5/30, 32 phases and an 8-channel cardiac coil. Injection and acquisition were started simultaneously. For comparison, cine phase-contrast was also acquired in the ascending aorta with venc=150-250 cm/s.

Field map extraction: To optimally combine signals from each coil, relative complex sensitivity maps were estimated from the sum over all acquired cardiac phases. For each cardiac phase, phase corrected signals from the multiple coils were combined according to weighted least squares. Using TE, phase maps were converted into field maps. 4) To assess the effect of the CA only, a pre-contrast initial field map was estimated by the mean value of the first 3 acquired cardiac phases, which was subtracted from subsequent cardiac phase field maps.

Aortic arch 3D model construction: From the 2D signal intensity map, a 3D surface mesh for the aortic arch was generated. 1) The aorta was manually outlined on the 2D image and a contour was interpolated using spline (Fig. 1-a). 2) 2D Delaunay triangulation was performed on the spline points. 3) 2D triangulation was converted into a closed 3D surface mesh by creating circles over each 2D triangle edge (Fig. 1-b).

CA quantification: Knowing the shape and its orientation with respect to B_0, the shape factor β was calculated for each pixel within the imaged slice and inside the aortic model with an algorithm using Maxwell boundary element method on the 3D surface mesh[3] (Fig. 1-c). For each cardiac phase, least-square fitting was performed to extract concentration: $[CA]=(\beta'\beta)^{-1}\beta\Delta f/(\chi_m f_0)$. χ_m=308 ppm mol/L at 310 K.

Cardiac output quantification: Similar to the dilution methods for flow rate estimation, the Stewart-Hamilton principle was applied by fitting the concentration evolution to a dispersion model[4] using non-linear least-square to estimate the area under the curve. Reproducibility of the processing was estimated by repeating the model construction and the fitting procedures. Mean cardiac flow rate was also obtained from standard processing of the PC data.

RESULTS

The calculated shape factor correlated well with the measured field map (Fig. 1-c,d). The field was homogeneous along straight segments of artery which had constant orientation with respect to B0. Within the curved part of the aortid arch, the field varied as a function of angular orientation reaching negative values for the aorta portion perpendicular to B0, as predicted from the infinite cylinder model for shape factor[5].

A good correlation (r between 0.5 and 0.8) between the fitted field and the measured one and a precision ~0.15 mmol/L were obtained.

A reproducibility of 2% on flow rate was obtained for the shape dependence. A summary of the measured flow rates compared to PC (Table 1) shows a single outlier (case 5). If considered, Bland-Altman comparison gave a significant bias of 17% and a standard deviation of difference of 26% with a poor correlation (0.1). If excluded, a non-significant bias of 3% and an agreement of 17% with a correlation of 0.6 were obtained. Physiological variations, motion during breath-hold or injection induced biases may partially explain the discrepancy between techniques. In particular, for case 5, the overestimation may be due to an incomplete flush of the bolus leading to a higher apparent dilution factor through the aorta.

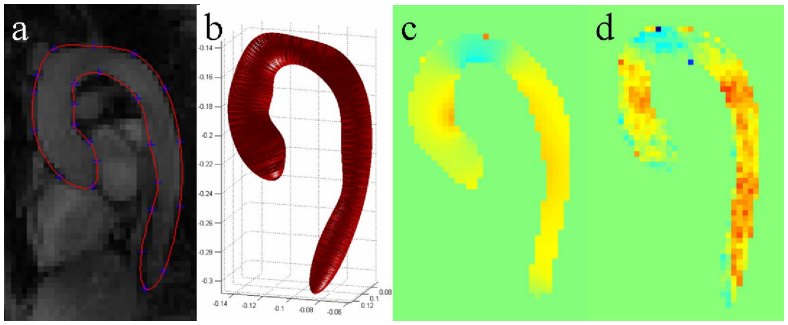

Fig. 1: Selected points on aorta contour (blue) and 2D spline (red) superimposed on signal intensity map (a). 3D rendering of the aortic model (b). Simulated shape factor for the points inside the aorta within the imaged slice (c) and measured field map for the cardiac phase with highest contrast (d).

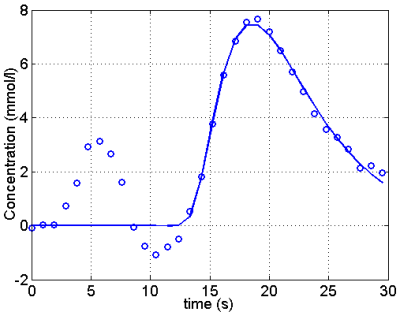

Fig. 2: [Gd] evolution as a function of time from injection start and fit to the dispersion model for CO quantification.

Table 1. Flow rates (in ml/beat).

Case	PC	Gd
1	94	89
2	119	99
3	100	126
4	138	151
5	70	142
6	116	138
7	123	109

DISCUSSION AND CONCLUSION

The constructed aortic model predicted the spatial shape factor observed in vivo in the aortic arch during Gd injection. [Gd] precision of 0.15 mmol/L was obtained. These results suggest that MR can measure [Gd] and physiological parameters (CO) by taking advantage of geometrically dependent susceptibility induced field distortions in curved vascular structures like the aortic arch. Although requiring a more complex protocol than cine PC, this approach could be easily inserted in Gd clinical studies that are not using first-pass for imaging and provide an independent method for assessing cardiac function that could corroborate other techniques, or evaluate injection quality. Although, it is not clear why such an overestimation compared to cine PC occurred in one case, reproducibility studies may allow distinguishing between protocol issues or intrinsic technique-induced biases. Additionally, supplemental parameters can be derived such as mean transit and dispersion times. Similar approaches may be adapted to other organs (kidney, liver) for functional studies (e.g. perfusion) as the main advantage among relaxation techniques rely on the strict linearity with CA concentration and the simple MR acquisition scheme.

REFERENCES 1. Hoffman et al., JMR, 178p237 **2**. Salomir et al., CMR, 19bp26 **3.** de Munck et al., IEEE TMI 15p620 **4.** Millard et al., Am J Physiol 272, H2004 **5.** Haacke et al., Magnetic Resonance Imaging p755