Non-contrast enhanced 3D pulmonary MRA with vein-artery differentiation

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Introduction: Pulmonary embolism is a life-threatening condition that requires prompt diagnosis and treatment. Contrast enhanced CT angiography (CTA) is the first-line imaging modality when pulmonary embolism (PE) is suspected. For those who are allergic to iodinated contrast, Contrast Enhanced MR angiography (CE-MRA) is an alternative imaging method that allows direct visualization of the pulmonary vessels. However, there is a group of patient with impaired renal function that cannot receive intravenous contrast agents. Pregnant patients, who are at elevated risks for PE, would also benefit from a radiation-free (and gadolinium-free) alternative. Compared to the currently-available Fresh Blood Imaging (FBI) method, the proposed approach resolves the entire cardiac cycle (rather than only a diastolic and a systolic phase), thus bypassing the need for preparation scans to accurately adjust delay times for each individual[1].The present work is based on a modified ECG-gated 3D spoiled gradient-echo sequence, and uses flow-related enhancement(FRE) as an intrinsic source of contrast to detect vessels and evaluate flow hemodynamics. A hybrid parallel-imaging acquisition scheme was implemented, along with slab-interleaved coverage [2, 3] (more details below).

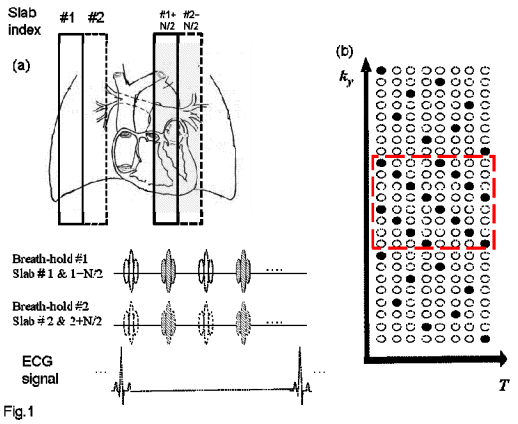

Fig.1

Methods: The entire chest was covered using a number of sagittal 3D slabs (Fig. 1). These slabs were kept fairly thin (about 20 mm) to allow for strong FRE. One slab in the right lung and one in the left were acquired during each breath-hold (Fig. 1). Interleaving the acquisition of two slabs in this manner allowed for longer TR values, and thus stronger FRE. Accelerated imaging proved crucial toward keeping scan time within a reasonable breath-holding duration. The subsampling scheme in Fig. 1 was employed, where *T* represents cardiac phase. The hybrid scheme employed here combines parallel imaging with a retrospectively-gated version of the UNFOLD method[2], along with adaptive regularization [3]. All acceleration was performed along the *y* direction (A/P), as slabs tended to be too thin to allow for significant acceleration along *z* (R/L). A net acceleration of 6.5-fold was obtained using a four-fold acceleration with shear grid sampling [4] near k-space center, and an acceleration of 8 for the rest of k-space. While this high acceleration factor may lead to artifacts in the cardiac-phase resolved images, these intermediate results are used only to detect pulsatility and differentiate veins from arteries. Final angiograms were generated from all data at all phases, and did not require any special reconstruction method. Once all slabs were reconstructed, dynamic properties of the vessels were analyzed to differentiate veins from arteries, slabs were coregistered to generate a single 3D volume over the whole chest, and this volume could be reformatted to coronal planes if so desired.

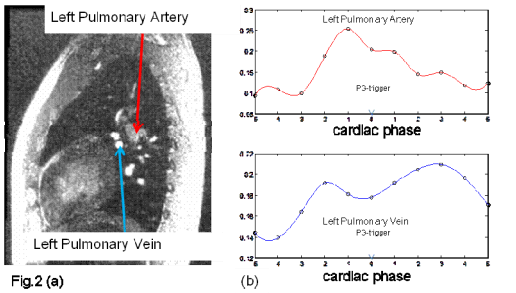

Fig.2 (a) (b)

Volunteer scans were performed on a 1.5T and a 3.0 T scanner (in both cases: flip angle = 20°, TE ≈ 1.5 ms, sagittal slabs with in-plane FOV = 30×24 cm^2, 200×160 matrix size, 1.5×1.5×2.0 mm^3 resolution, at least 8 cardiac phases, 12 to 20 slabs that required 6 to 10 separate breath-holds).TR and breath-hold duration were 6.4 ms and 15 s, respectively, on the 3.0 T scanner, while they were equal to 7.5 ms and 20 s on the 1.5 T system.

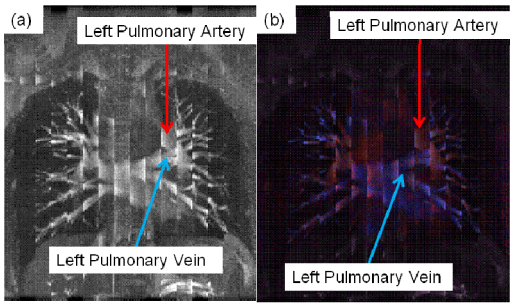

Fig.3

Results: Fig. 2 shows one reconstructed slice for a healthy volunteer imaged on the 1.5T system. The arrows point at one branch of the left pulmonary vein(blue) and of the left pulmonary artery(red). Temporal signal changes at these two locations are depicted in Fig. 2(b) using spline interpolation to fill the space in-between the 11 known cardiac phases. The zero point on the horizontal axis (blue arrow in Fig. 2b) indicates the reconstructed frame when the peripheral gating (PG) trigger was received. Due to the delay inherent to PG, the ventricular contraction is expected to occur ahead of it (red curve in Fig. 2b).Peak enhancement in veins is expected to occur later than in arteries (blue curve in Fig. 2b).As can be seen from Fig. 2b, dynamic information from the reconstructed cardiac-phase resolved data allowed for arteries and veins to be differentiated. Fig. 3(a) shows reformatted coronal maximum intensity projection of the posterior slabs after coregistration. Inconsistent breath-hold positions led to non-rigid deformations that were not entirely accounted for in the present implementation, and some misregistration may still be observed. Despite the slight misregistration, the lobar and segmental pulmonary arteries are well demonstrated. Fig. 3(b) further includes the use of color to label the dynamic properties of the vessels in Fig. 3 (a), i.e., to differentiate veins (blue hue) from arteries (red hue). While veins appear mostly blue and arteries mostly red, as they should, vessels tend to be 'peppered' with wrongly-identified pixels with the wrong color. Further improvements in registration and in vein-artery discrimination can be seen as work in progress.

In summary, we propose a non-contrast enhanced chest imaging MRA method capable of discriminating between veins and arteries based on the pulsatility of these vessels through the cardiac cycle. The approach is meant as an alternative imaging modality for CTA and CE-MRA in patients with impaired renal function (who cannot receive intravenous contrast agents) and/or in pregnant patients (who cannot receive gadolinium and preferably should not be exposed to ionizing radiation).

References:[1] Miyazaki M et al. Radiology,2008.[2] Madore B et al. MRI 2011. [3] Chao TC et al. MRM 2010.[4]Kellman, P. et al, MRM, 2001

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