3D Non-Rigid Motion Modeling of the Liver from Undersampled Golden-Radial Phase Encoding (G-RPE) Acquisitions

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INTRODUCTION: Motion models derived from MRI have become a valuable tool for respiratory motion compensation in several applications, such as radiotherapy treatment planning [1] or X-ray guided cardiac catheterizations [2]. However, MR-based motion modeling of abdominal organs remains a challenge due to the large non-rigid deformations caused by respiratory motion [3] and the difficulty of acquiring multiple respiratory phases in 3D with adequate contrast and spatial resolution for accurate motion estimation. Recently, the retrospective reconstruction of multiple 3D near-motion-free respiratory phases from free-breathing acquisitions has been shown using a Golden-Radial Phase Encoding (G-RPE) trajectory [4,5]. In this work we combine this acquisition scheme with an SSFP contrast to reconstruct multiple images with both adequate contrast and high spatial-temporal resolution. This acquisition provides one high quality image for the most frequent respiratory position as well as several images at different respiratory positions with any arbitrary respiratory resolution (i.e. number and size of respiratory phases, also called bins). These images allow us to apply accurate non-rigid motion modeling of the abdominal tissue deformation.

METHODS: 1) Acquisition and Reconstruction: G-RPE combines Cartesian sampling in the readout direction \( k \), with an undersampled radial-like scheme in the phase encoding plane \((k_i,k_z)\), having an angular step between consecutive profiles of 111.25° given by the Golden ratio (Fig 1a). A self-gated respiratory signal is obtained from the central \( k \)-space spokes, which are acquired along with each radial profile (Fig 1a-b). G-RPE allows the flexibility to reconstruct several images \( I_i \) at any arbitrary respiratory position or bin \( B_i \), by selecting different radial profiles according to their position in the respiratory cycle (Fig 1c). To ensure that each \( I_i \) has enough quality to be registered, a maximum angle \( \alpha_i \) in the trajectory of \( I_i \) is applied as registration quality measure, similar as in [6]. Simulations showed that images reconstructed using non-Cartesian iterative SENSE [6] require an angle of \( \alpha_i < 6.25^\circ \) to ensure enough quality for registration, not necessarily satisfying the Nyquist criterion.

2) Motion modeling: We apply a non-rigid intensity-based registration algorithm [7] to compute the registrations \( R_i \) which align the features of the highest quality image \( I_1 \) at the most visited respiratory position (most commonly at end-exhale) with the features of all other phases \( I_i \) (Fig 1d). These registrations are used to model the complete respiratory cycle with multilevel B-Splines [8] allowing us to predict an image \( M_i \) as a deformation of \( I_1 \) at any arbitrary respiratory position \( t \) between end-exhale and end-inhale.

3) In-vivo experiments: Five healthy volunteers were scanned on a 1.5T Philips scanner using a 32-channel coil and the proposed method. Relevant sequence parameters include: balanced SSFP sequence, FOV = 287mm², isotropic resolution = 1.75mm³, TR/TE 3/1.43ms, flip angle 30°, \( P = 1640 \) profiles with radial undersampling of 2, scan duration \( t = 6.8 \)min. For each volunteer, \( N = 9-11 \) respiratory phases were reconstructed from exhale \( (I_1) \) to inhale \( (I_N) \). To validate the accuracy of the deformations, for each image pair \((I_i,I_j)\) we manually defined 10 landmark correspondences at anatomical structures within and around the liver. As a measure of deformation accuracy, we computed the target registration error \( (TRE) \), i.e. the mean distance between the landmarks in \( M_i \) and those in \( I_j \) before and after deformation.

RESULTS: Fig 2 (top) shows two exhale \((I_1,I_2)\) and two inhale phases \((I_{10},I_{11})\) in coronal view from a sequence of \( N = 11 \) reconstructions of a single volunteer. Although image quality decreases from exhale to inhale due to increased undersampling artifacts, our image quality measure \( \alpha_i \) ensures sufficient quality of \( I_i \) for registration. The corresponding model predictions \( M_i \) at the same respiratory positions overlaid by example motion fields are shown in Fig 2 (bottom) for coronal and transversal views. Image quality now is highly improved since each \( M_i \) is a deformation of \( I_1 \).

CONCLUSIONS: We applied a non-rigid motion modeling technique to retrospective and self-gated reconstructions of multiple respiratory phases of the abdominal region from free-breathing MRI acquisitions. Accurate model predictions have been shown with an overall error of 1.83 ± 1.22mm. As future work we will investigate the use of different motion surrogates (e.g. motion of the diaphragm, chest, belly or body surface) that can be extracted from the 4D images to steer the motion model.