Optimal MRI sampling and binning for online 4D retrospective respiratory motion analysis of the abdomen

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PURPOSE Motion caused by respiration is a well-known problem for radiotherapy in upper abdominal organs. The recent introduction of MR-guided radiotherapy opens up a way to adapt the radiation to daily anatomical changes and variations in motion[1]. Online patient specific motion characterization of the tumor and surrounding organs allows for adaptive radiotherapy treatment plans in order to increase tumor dose coverage while minimizing the delivered dose to radiosensitive healthy tissue. However, such an exam has to fit within a clinically acceptable timeframe (e.g. <15 min) and with sufficient SNR, contrast and coverage to capture all structures. Due to the relatively slow acquisition speed of MR imaging, retrospective binning of multi-slice 2D images to multiple respiratory phases was proposed for 3D motion characterization[2]. Buerger *et al.* showed good image quality using retrospective binning of *k*-space data acquired with a custom 3D readout[3], but their reconstruction time was ~7.5 hours, which is undesirable for our application. In this study we propose to use 3D imaging to calculate 3D motion fields for multiple respiratory phases within a clinically acceptable timeframe, using clinically available protocols and retrospective binning of the acquired *k*-space data. For this purpose, two binning methods (absolute and relative amplitude binning) were explored and two sampling strategies were investigated. 3D motion fields were calculated using a 3D optical flow algorithm[4,5].

METHODS

MR-imaging: Six healthy subjects and one patient with a pancreas carcinoma were scanned on a Philips Achieva 1.5 T scanner with either a Cartesian or a radial acquisition. Cartesian: 3D transient bSSFP (Philips balanced Turbo Field Echo, bTFE) data was acquired during free-breathing for 10:00 minutes (TE/TR=1.45 ms/3.0 ms, $\alpha=30^0$, FOV=330x250x64 mm^3, voxel size=1.88x1.85x4.00 mm^3, 50 dynamics). Radial: 3D bTFE with radial in-plane sampling (TE/TR=1.45/3.0 ms, $\alpha=30^0$, FOV=350x350x96 mm^3, voxel size=2.0x2.0x4.00 mm^3, 25 dynamics) data was acquired during free-breathing for 8:30 minutes. The turbo direction was placed along *kz* for both acquisitions, acquiring all *kz* lines in a single TFE shot (shot length 60 ms (Cartesian), 93 ms (radial)). A respiratory navigator was acquired prior to each TFE shot to obtain the respiratory signal for each Cartesian segment/radial stack-of-spokes (i.e. 'blade'). *Binning methods*: To get insight into imaging time and corresponding *k*-space filling, simulations were performed using the respiratory navigator signal. The navigator information was used to rebin the 3D data into ten respiratory bins. Two methods were implemented; 1) Absolute amplitude binning, and 2) Relative amplitude binning. For the first method, the navigator data was divided into ten equal amplitude bins, with a distinction between in- and exhalation, to capture the differences between in- and exhale motion paths (i.e. hysteresis). The second binning method first normalizes the navigator envelope to the minimum and maximum amplitude and then defines ten phases. This will result in a more uniform filling for the different phases, but may decrease image quality. For both binning methods the amount of k-space filling was determined for various scan durations (i.e., cut-off points). Based on these simulations, the most efficient binning method was chosen. Image reconstruction was performed in Matlab (Mathworks, Natick, MA) on an 8-core CPU personal computer using ReconFrame software (Gyrotools, Zurich, CH). The interleaved navigator signal was used to assign each TFE shot to a respiratory phase according to the relative amplitude binning method. Complex averaging was used when multiple TFE shots were assigned to the same respiratory phase. No phase correction was applied. To assess the tradeoff between scan duration and image quality, multiple reconstructions were performed using an increasingly larger subset (10, 20, and 25 dynamics) of the acquired data. To estimate 4D motion for ten respiratory phases, a 3D optical flow algorithm was applied using the first phase as reference.

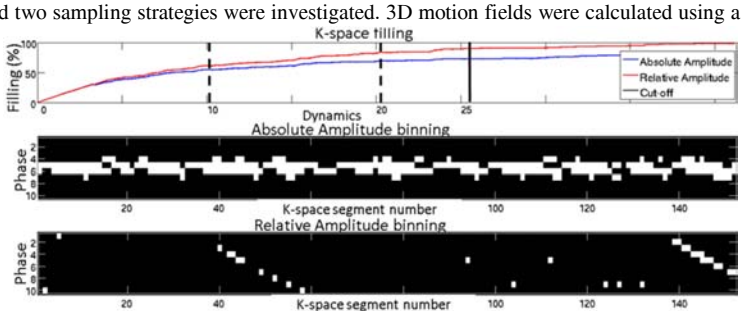

Figure 1 Filling of *k*-space (a) and corresponding filling of different respiratory phases for absolute (b) and relative (c) amplitude binning

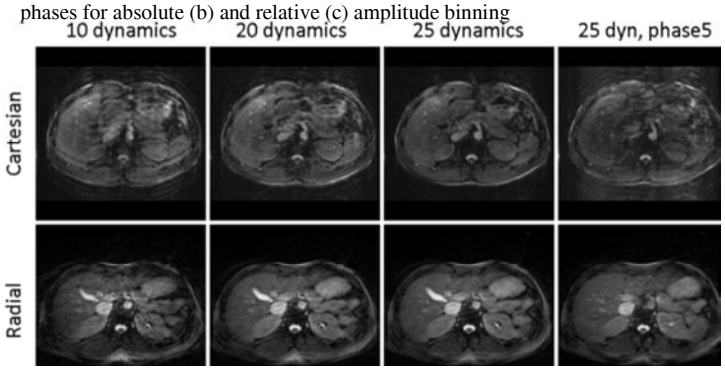

Figure 2 Reconstructed images of two sampling strategies using a different amount of dynamics for respiratory phase 1 or phase 5 (right column)

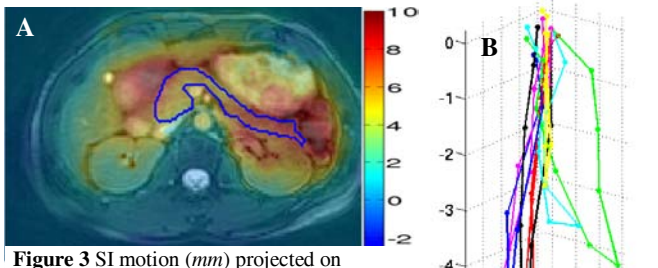

Figure 3 SI motion (*mm*) projected on anatomical image the pancreas delineated in blue. Right image shows the mean 4D motion of the whole pancreas of all the subjects and the mean tumor patient (cyan) of the patient

RESULTS AND DISCUSSION Simulations showed a rapid increase in *k*-space filling at first, but filling approaches 100% asymptotically due to the stochastic character of respiratory filling (**Fig. 1a**). **Fig. 1b** and **1c** show for each phase which of the TFE-shots are collected after the data is pooled using 25 dynamics (white denotes missing data). Relative amplitude binning shows a more uniform *k*-space filling, especially for phase numbers 5 and 6 (which resemble end-inhalation) and was therefore used for further analysis. For Cartesian imaging, an imaging time of 10:00 minutes was chosen, which resulted in 99% ± 0.5% *k*-space filling, whereas for radial sampling an imaging time of 8:30 minutes was chosen, which resulted in 93% ± 2.5% filling (**Fig. 1a**). **Fig. 2** shows the reconstructed Cartesian and radial data for phase 1 (using 10, 20 or 25 dynamics for reconstruction) and phase 5 (using 25 dynamics). The dashed lines in **Fig. 1a** resemble 10 and 20 dynamics. Cartesian sampled images show ghosting artifacts for all reconstructions, whereas radial sampled images show only minor streaking artifacts when 10 dynamics are used. This shows that radial sampling is more benign for undersampling. The right column shows the reconstructed images for respiratory phase number 5 (i.e. end-inhalation), since this is the most problematic phase in a respiratory cycle to reconstruct due to intra-phase variation (i.e. *k*-space inconsistencies due to positional differences, similar to intra-view motion corruption in conventional imaging). Despite sufficient *k*-space filling, the Cartesian sampled image shows severe ghosting artifacts, whereas the radial image shows equal image quality as the first respiratory phase, which can be explained by the fact that radial acquisitions are less prone to motion artifacts as the center of *k*-space is sampled for each spoke. The total time (acquisition + reconstruction/rebinning + 3D optical flow analysis) was 8:30 + 2:44 + 1:39 = 12m53s for radial sampling. Longer reconstruction time was required due to the gridding step in radial sampling. **Fig. 3a** shows an example of the SI-motion between phase 1 (inhale) and phase 5 (exhale) projected on top of the anatomical image with the pancreas delineated in blue. Large motion can be observed for the abdominal organs, whereas the motion in the spine and skin is considerably less. **Fig. 3b** shows the mean 4D motion in *mm* for the whole pancreas and the tumor all of subjects. Hysteresis and inter-subject variation within the motion trajectories can be observed (**Fig. 3b**), next to intra-organ motion for the pancreas (**Fig. 3a**).

CONCLUSION We presented a way to estimate 3D motion for ten respiratory phases based on retrospectively rebinning of *k*-space data. Radial in-plane sampling combined with relative amplitude binning showed the best results in terms of image quality, SNR, and imaging time. In combination with a fast 3D optical flow algorithm, we were able to calculate motion estimations within 13 minutes for all structures within the FOV using a clinically available pulse sequence.

REFERENCES **1.** Lagendijk JJW et al. *Radiotherapy Oncol*, 2008;86:25-29 **2.** Tryggestad E, et al. *Med Phys*, 2013;4(5):051909-21 **3.** Buerger C et al. *IEEE Trans on Med Imag*, 2012;31(3):805-815 **4.** Horn B, et al. *Artif Intel* 1981;17:185 – 203 **5.** Roujol S, et al. *IEEE Trans on Med Imag* 2011;30:1737 – 1745