Improvement of perfusion imaging of the human lung using retrospective gating

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Introduction

In various diseases the perfusion of concerned tissue shows pathological characteristics. Therefore perfusion imaging and quantification can provide an additional tool for more specific diagnosis of certain disease patterns. The lung is an organ where certain obstacles like low proton density, high susceptibility fluctuation, and motion due to respiration and heartbeat impose challenges for imaging. In functional imaging with Arterial Spin-Labelling (ASL) the standard measurement is performed within a single breath-hold in order to obtain data that can validly be used in the subsequent image calculation. These methods are restrained to short scan times of up to 25 seconds due to the limited capability of patients with lung disease to hold their breath.

Improvement of SNR in perfusion images can be achieved by increasing the number of averages. This necessitates longer acquisition times which can only be realized when the patient may breath freely. A method of grouping the obtained images after acquisition into sets of equal position within the respiratory cycle is the main focus of this work.

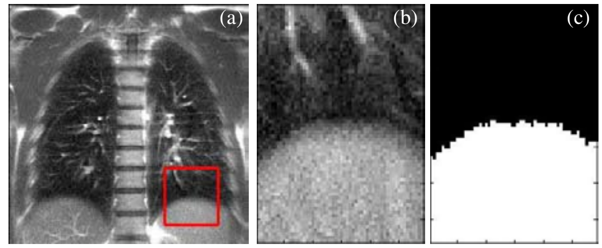

Figure 1: Processing of a ROI in the diaphragm region. (a) Selection of the ROI, (b) ROI before filtering and delineation, (c) ROI with delineated diaphragm edge.

Materials and Methods

The data sets to be used in the evaluation of the presented algorithm contain between 50 and 200 images (scan time of approximately 4 min to 15 min respectively) of which one half was acquired after applying a global (tag) and the other half after a selective (control) inversion pulse of a FAIR [1] tagging scheme. ECG triggering was used in order to ensure acquisition at the same instant of the blood inflow cycle for all images. The tagging and the acquisition (using fast HASTE readout) were performed on a 1.5 Tesla whole body scanner (MAGNETOM Avanto, Siemens Medical Solutions, Erlangen, Germany) in two subsequent diastolic cycles by choosing the inflow time (TI = 900 – 1600 ms) suitably in order to minimize incomplete tagging or flow artefacts. Patients were asked to breath freely during the measurement.

A software routine for post-processing of the acquired image data was developed in MatLab (The MathWorks Inc., Natick MA, US). It contains the standard averaging and subtraction scheme necessary for the analysis of ASL data together with a tool allowing for the identification and grouping of images acquired at the same moment in the respiratory cycle using the position of the diaphragm as an indicator. This retrospective respiratory gating uses the fact that the diaphragm is a high signal region in T_1-weighted images. It works as a three-step process.

First a region of interest (ROI) is marked in a way as to include the upper part of the diaphragm and a certain area of the lung above as shown in figure 1(a). The image content of this region is extracted from each image of the data set and filtered in a three-step cycle: Suppression of low signal pixels ($S_{pix} \leq 0{,}5 \cdot \overline{S}_{image}$) followed by zeroing of the sections of columns above (cranial) a significant amount of zero signal pixels, as well as the suppression of low signal lines ($\overline{S}_{line} \leq 0{,}4 \cdot \overline{S}_{image}$). This filtering is necessary to avoid false location of the diaphragm edge due to signal from the lung parenchyma in the following step.

The pre-filtered ROIs are fitted column-wise with a Heaviside step function (Eq. 1) utilizing the correlation coefficient as defined in equation 2 (where index i ranges from one to the maximum number of pixels per column). The position of the step (x_0) with the highest correlation coefficient marks the edge of the diaphragm in each image column (see Fig. 2). Figure 1(c) shows the delineation of the diaphragm obtained from figure 1(b) by applying the method described above.

$$\Theta(x - x_0) = \begin{cases} 0 : x \leq x_0 \\ 1 : x > x_0 \end{cases} \quad (1) \qquad r(x_0) = \frac{\sum_i S(x_i) \cdot \Theta(x_i - x_0)}{\sqrt{\sum_i S^2(x_i) \cdot \sum_i \Theta^2(x_i - x_0)}} \quad (2) \qquad r_{ab} = \frac{\sum_{ij} S_a(x_{ij}) \cdot S_b(x_{ij})}{\sqrt{\sum_{ij} S_a^{\,2}(x_{ij}) \cdot \sum_{ij} S_b^{\,2}(x_{ij})}} \quad (3)$$

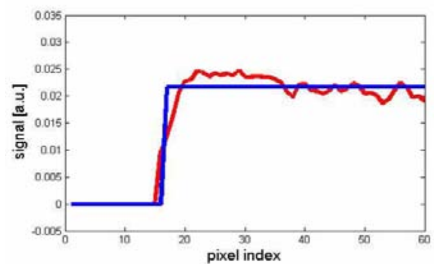

Figure 2: Fit of the filtered signal distribution in a significant column of the chosen ROI (red) with a Heaviside step function (blue). The edge of the diaphragm can clearly be recognized by the sharp transition of signal between lung region (left) and diaphragm (right).

In a third step all the ROIs with delineated diaphragm are correlated among each other using the correlation coefficient defined in equation 3 (where indices a,b represent different images and index i,j range from one to the maximum number of pixels per line or column respectively in the ROI) and grouped into sets of images that correlate better than a certain threshold $r_{ab,thresh}$ among themselves. The acquired images corresponding to the ROIs of such a set are grouped, averaged and subtracted according to the standard procedure for breath-hold ASL images.

Results

Processed ASL measurements in coronal as well as in sagittal slices of the lung show excellent suppression of static tissue and high SNR as compared to conventional breath-hold imaging. Figure 3 presents a comparison with respect to image quality of a single subtraction perfusion image (SNR ~ 5,5) to a perfusion image (SNR ~ 14) calculated from 9 control and 11 tagged images out of a set of 150 acquisitions grouped by the algorithm ($r_{ab,thresh}$ = 0,99). Blood vessels in the lung and liver region show very detailed structures and no signs of distortion induced by inaccuracies in averaging. The separation of the three lung lobes is also visible.

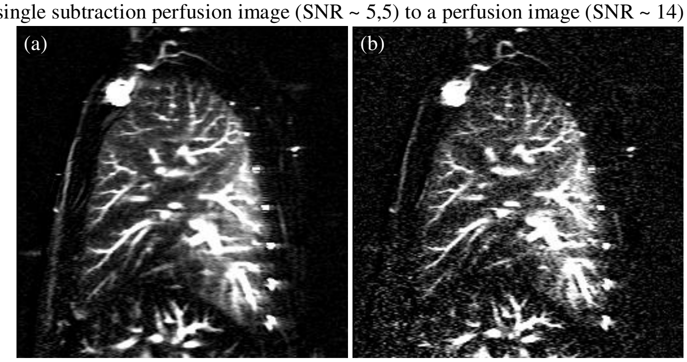

Figure 3: Comparison of (a) an ASL image calculated from 9 control and 11 tagged images out of a set of 150 acquisitions during free breathing to (b) an ASL image calculated from only one tag/control pair (b) of the same sample.

Discussion and Conclusion

The presented method shows a high level of accuracy in the evaluation of the current lung position within the respiratory cycle as can be verified by the exact delineation of blood vessels and the lung boundaries after averaging. The use of gating allows for measurements over an extensive period of time during free breathing of the patient.

Future work will focus on including the presented algorithm directly into the image-reconstruction program to allow for an online feedback on the current image quality.

References

[1] Kim S.G., *Magn. Reson. Med.* 34(3):293-301, 1995