Ventilation Dependent Blood Volume in Fourier Decomposition 1H Lung Imaging

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Introduction

Pulmonary proton MRI has long been plagued with low SNR because a majority of the lung is gas with low proton density. A promising new approach acquires serial images during free breathing. Each image is registered to a reference image [1-5] and the time series of each voxel is Fourier transformed [3,5]. Peaks in the spectrum at the breathing and heart rate frequencies are used to identify the amplitude of regional proton density changes related to "ventilation" and "perfusion", respectively. This is known as the Fourier Decomposition (FD) method. The registration algorithm is a non-rigid transformation that tracks the displacement of each position within the lung. The position and size of each voxel is registered to a corresponding voxel in the reference image, but the intensities are not corrected for voxel size changes. Previously, the ventilation related change in proton density has been solely attributed to changes in proton density from any additional inhaled gas volume [1]. There is, however, a second component to the ventilation related proton density change that has not been considered. As lung volume increases, this second effect causes an additional reduction in the proton density due to a decrease in blood volume caused by the stretching of alveolar septa and compressive force on the capillary blood [6]. Here we consider both causes of ventilation dependent proton density changes and describe how to independently measure them.

Method

The septal tissue is modeled as having two components, non-blood parenchymal tissue and blood. The existing method described above measures the total "ventilation" change in proton density that is due to both the dilution of the proton density from inhalation of additional gas and also due to a reduction in blood volume. We note that the non-rigid registration algorithm can also be used to determine the geometric size change of each voxel, ignored in the existing FD implementation. The ratio of the final to initial voxel volumes determined from the registration algorithm is, we claim, a measure of the proton density change due solely to the first component, i.e. the dilution of the proton density due to inhalation of gas. Therefore, subtracting the fractional change in proton density determined from the registration algorithm from the total fractional change will provide a measure of the second component of the change in proton density that is due to blood volume change.

Relative Magnitude of Ventilation Dependent Proton Density Changes

The regional proton density in a voxel is given by: $\rho = (M_T + M_B)/(V_G + V_T + V_B)$, where M_T and M_B are the masses of the non-blood tissue (*T) and blood (B)* compartments, and V_G , V_T, and V_B are the volumes of gas, non-blood tissue, and blood, respectively. Here we estimate the relative changes in proton density from the two contributions described above. We start at FRC with an estimated lung density ρ = 0.2 and, in some region of interest, $M_T = M_B = M_0$, where M_0 is a reference mass. In this example, 50% of septal tissue is composed of blood and 50% of non-blood tissue. Since the proton density within the septal tissue is ~1, $V_T = V_B = M_0$ and in order for ρ_{FRC} = 0.2, V_G = 8 M_0. Now consider a tidal volume V_{tidal} raising V_G by 25% to 10 M_0, with no change in blood volume. This results in a proton density at the new lung volume from mechanism 1 (gas volume change): ρ(FRC+ V_{tidal}) = 0.17. We also consider the effect on ρ due just to a blood volume change. We interpolate this from estimates at TLC, where we take the blood/tissue fractionation to be 20:80, which represents a change in blood volume from FRC to TLC of ΔV~0.75 M_0. If we estimate TLC – FRC ~ 3.5L and FRC = 2.5L, then a 25% change in V_G = 0.625L and the interpolation estimate of the change in blood volume (and mass) is 0.1786*0.75 M_0 = 0.134 M_0. The new blood volume and mass at FRC + V_{tidal} is then (1-0.134) M_0 = 0.866 M_0. Substituting these values into the density expression, we have ρ(FRC+ V_{tidal}, blood volume contribution) = 0.155 showing the two effects are of similar magnitude.

Discussion

As presently practiced, the FD method acquires data from a single slice rather a full 3D volume. Therefore, the geometric scaling factor determined from the non-rigid registration algorithm only accounts for volume changes in 2 of the 3 spatial dimensions. We note, however, that for coronal images where the slice selection direction is anterior/posterior, A/P lung motion is likely to be small.

In the limit where septa are heavily fibrotic (i.e. advanced fibrosis or interstitial lung disease), the composite septa are essential incompressible because the blood volume component is small, and this, together with the fact that stiff incompressible vessels tend to preserve luminal volumes, implies that the proton density changes associated with blood volume changes goes to zero. Therefore, we hypothesize that a measure of interstitial fibrosis will be a decreasing magnitude of the ventilation dependent blood volume change.

Note also that the sensitivity to this effect can be increased with larger tidal volumes. At increasingly larger lung volumes in normal subjects, the reduction in blood volume also increases. We conclude that a determination of the proton density change associated with blood volume changes during ventilation could be used as an independent, noninvasive, and regional measure of interstitial fibrosis.

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