Fat-Based Registration of Breast DCE Water Images

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Synopsis

Three-dimensional breast dynamic contrast-enhanced imaging is susceptible to deformable motion and affects both semi-quantitative and pharmacokinetic parameters. B-Spline motion registration with a mutual information metric is often used to register DCE images but is sometimes susceptible to introduction of new motion. Here we have introduced a fat-based motion registration, using a mean-squared-difference signal metric, to register the water images without introducing new motion. The acquired images and both registration methods were qualitatively assessed in 16 breasts. Voxel-by-voxel pharmacokinetic mapping was also performed in 21 tumors. Our results show that fat-based registration can be used to register the water images with improved image quality and reduced errors in quantification.

Introduction

Breast dynamic contrast-enhanced (DCE) imaging often acquires many three-dimensional datasets for over 7.5 minutes and is susceptible to deformable motion. This motion affects both semi-quantitative measures such as washout slope as well as voxel-by-voxel pharmacokinetic parameters, and therefore needs to be corrected before quantification. B-Spline based motion registration [1] is commonly used to register deformed 3D images along with a normalized mutual information metric to account for temporally varying water signal due to contrast dynamics. Although this method reduces motion in several cases, additional jittery motion is also sometimes introduced with the mutual information metric. Here we introduce a fat-based registration method, using mean-squared-difference signal metric, to register the water images without introducing new jitteriness.

Methods

The water signal in the breast changes with the contrast concentration; however, the fat signal does not exhibit temporal dynamics (Fig.1) and hence the smoothed temporal fat signal changes can be used to estimate the motion when Dixon-based methods are used. 3D fat-based motion registration was performed with mean-squared-difference signal as the metric and b-spline transformation with a grid dimension of five along each direction. Water images were then warped using the corresponding fat-based b-spline transform. The steps of this registration method are detailed in Fig.2.

The original image series and images from both fat- and water-based registration methods were evaluated in 14 patients (2 bilateral) with known masses. 3D RF-spoiled gradient echo fat-water separated DCE images were acquired using DISCO [2], a pseudorandom k_y-k_z sampling scheme enabling a favorable tradeoff between temporal and spatial resolution, on a 3T scanner (GE Healthcare, Waukesha, WI). The imaging parameters were: FOV= 270×324 mm, TR/TE_1/TE_2= 6.3/2.2/3.3 ms. One pre-contrast and four post-contrast images were acquired with high spatial resolution of 0.5×0.6×1.0 mm and low temporal resolution of 2 min. Fourteen images were acquired during the wash-in period with high temporal resolution of 13s and lower spatial resolution of 0.5×1.2×2.0 mm. All the images were reconstructed to lower spatial resolution before performing fat and water-based registration.

The image quality of the original and the registration methods were presented in random order to 2 experienced readers and scored in a scale of zero to two. A score of 0 was given to images with minimal motion of < 2 voxels in each direction in both tumor and surrounding tissue. Similarly a score of 1 was given to residual motion of 2 to 4 voxels and score of 2 for > 4 voxels. The readers also ranked the images from best (rank 1) to worst (rank 3).

Voxel-by-voxel pharmacokinetic mapping was also performed using a standard Tofts model [3] with modified Fritz-Hansen AIF as well as modified local density random walk (mLDRW) Dispersion model [4] in 21 tumors. The average error between the data and the fitted model was evaluated over the tumor ROI for both the models using the original images as well as fat-based registered images.

Results

Fig.3 shows an example dense breast movie (viewable in browser) with minimal motion (white arrows). The water-based registration method introduces jittery motion (cyan arrows), and the fat-based registration reduces motion without introducing new motion, even in this case of minimal fat. The average image quality score of the original images, water-based and fat-based registration methods are tabulated in Fig.4a. Friedman test showed significant (P<0.01) differences between the three datasets. All the fat-based registered images were scored better than the original images. All the four images of fat-based registration that were ranked 2 had identical image quality score compared to water-based registration. Three to four water-based registered datasets received a score of 2 that was equal to or higher than the original images due to insufficient motion compensation or new motion.

Fig.5 compares example wash-out slope maps (top row), kep map of the Tofts model (middle row) and kappa map of the mLDRW model between the original and fat-based registration methods. These maps show the erroneous enhancement in the edges (arrows) that are corrected after motion correction. The average error between the data and the fitted model is also reduced after motion correction (Fig. 4b).

Discussion

Our results indicate that the fat-based registration of breast DCE images reduced the motion compared to both original and water-based registration method and improved image quality. Both registration methods, however, also introduced minimal blurring in patients with motion >4 voxels. The image quality including blurriness will be further evaluated in more patients with significant motion.

Conclusion

Fat-based breast DCE registration can be used to register the water images with improved image quality and reduced errors in quantification.

Acknowledgements

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References

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[3] Tofts P, et al., JMRI 1999;10:223-232

[4] Mischi M, et al., IEEE EMBS, 2013

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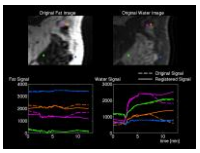
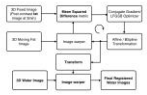
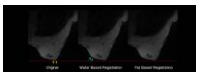
Figures



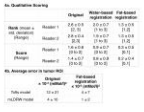
Fat signal at the fat/water interface (magenta and orange curves) show signal increase in the later time points corresponding to the motion. After registration (solid line), the corresponding fat signal is temporally constant and the water signal is also corrected to a persistently increasing signal. Blue and green curves far away from the fat-water interface also have a non-enhancing fat signal. Hence fat images can be used to estimate motion using mean-squared-difference signal as metric.



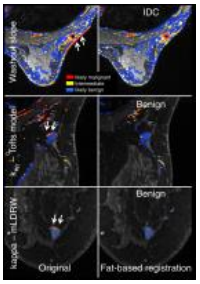
3D fat images were registered individually to the post-contrast fat image acquired at 3min with mean-squared-difference metric. Bulk motion in the fat images was corrected by affine transformation and the images were subsequently b-spline transformed. The final registered water images were obtained by transforming the individual 3D water images with the transform from the corresponding fat images.



Example dense breast movie showing motion in anterior to posterior direction (white arrows - left image). The corresponding water-based registration (middle image) introduced additional motion (cyan arrows) whereas the fat-based registration (right image) reduced the original motion without introducing any additional jitter. (Movie can be viewed in browser).



(a) Both the mean score and rank of the registration methods are better compared to the original images. The fat-based registration method had better qualitative score than the water-based registration method. (b) The average error between the data and the fitted pharmacokinetic model within the tumor ROI is reduced after image registration.



Washout slope maps (top) of IDC showing washout (red) in the breast surface due to motion (arrows). The kep map (middle) of Tofts model and the kappa (bottom) of the mLDRW model show higher values along the edges of

benign tumors (arrows) due to motion which were corrected in the corresponding motion registered images (right).