Improved Vessel Visualization in MIP/MinIP of MRA Images

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Introduction

Maximum intensity projection (MIP) and minimum intensity projection (MinIP) are the standard techniques used for the display of three dimensional (3D) MRA data sets. These display algorithms work very well when the contrast-to-noise ratio (CNR) between the vessels and surrounding tissues is high. But the MIP/MinIP angiograms of high-resolution MRA datasets can lose some vessel details due to lower CNR. In this study, a technique to create MIP/MinIP angiograms with pre-defined vessel voxel projection probability has been developed.

Theory

If the probability distribution function (pdf), *f(x)*, is known, then the probability that a random sample is less than *x* is given by: $F(x) = \int_{-\infty}^{x} f(u)du$. The pdf of MIP images, *g(x,n)*, and MinIP images, *h(x,n)*, can be obtained by an analysis similar to [1] as

$$g(x,n) = \frac{d}{dx}(F(x))^n = n(F(x))^{n-1} f(x) \qquad h(x,n) = \frac{d}{dx}(1-F(x))^n = n(1-F(x))^{n-1} f(x) \tag{1}$$

where n is the number of voxels along the projection path. Assuming that a single vessel voxel is present in the projection path through *n* voxels, the probability of the vessel voxel being projected into the MIP image and MinIP image are given by

$$P_{v,MIP} = \int_{-\infty}^{\infty} g_v(x,1) F_b^{n-1}(x) dx \qquad P_{v,MinIP} = \int_{-\infty}^{\infty} h_v(x,1)(1-F_b(x))^{n-1} dx \tag{2}$$

where subscripts *v* and *b* specify vessel and background distributions. For a given acquisition scheme, the only free parameter for increasing the probability of projection of a vessel voxel into MIP/MinIP image for the given path length is σ, the noise variance. The relationship between σ and the fraction of *k*-space, *f*, used for image reconstruction is given by $\sigma_f = \sigma\sqrt{f}$, where σ and σ_f are the noise variances of images reconstructed using the entire and *f* fraction of the *k*-space data, respectively. Hence, to improve the vessel visualization in MIP/MinIP images, $P_{v,MIP/MinIP}$ can be increased by adjusting the fraction of *k*-space used for image reconstruction.

In black-blood 3D fast spin echo (FSE) images, the vessel intensity distribution can be modeled by a Rician distribution with zero true value (assuming complete suppression of the blood signal), whereas the background tissue intensity distribution can be described by Gaussian distribution due to high SNR [2]. In this case, $P_{v,MinIP}$ is given by

$$P_{v,MinIP}(\sigma_f) = \int_{-\infty}^{\infty} \frac{x}{\sigma_f^2} \exp\left(-\frac{x^2}{2\sigma_f^2}\right) (0.5)^{n-1} \left(erfc\left(\frac{x-\mu}{\sqrt{2}\sigma_f}\right)\right)^{n-1} dx \tag{3}$$

where μ is the mean of the background tissue distribution. Similarly, $P_{v,MIP}$ can be calculated for TOF-MRA images, where the pdf of the background is also Gaussian and the pdf of the vessel intensity can be obtained empirically as suggested in [3].

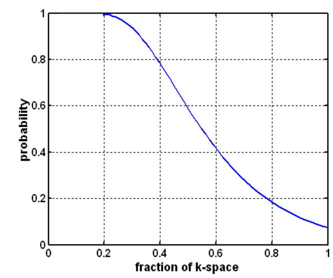

Figure 1. Dependence of the probability of projection of a single vessel voxel in the path in MinIP on the fraction of *k*-space used for image reconstruction. T2w data with μ=2.8 and σ=1.

<u>Steps of the Algorithm</u>

1. Reconstruct the image volume using the complete *k*-space data.
2. Estimate the parametric description of the background and the vessel distribution.
3. Numerically solve Eq. 3 to find *f* for the desired value of $P_{v,MinIP}$
4. Reconstruct the image volume using the central part of *k*-space (*f* fraction) by zero-padding the outer part.
5. Construct the MIP/MinIP image.

Results

The proposed technique was validated using 3D FSE data from a volunteer scanned on a 1.5 Tesla GE SIGNA Lx 8.4 MRI scanner (GE Medical Systems, Waukesha, WI) with NV/CVi gradients. The acquisition parameters for dual contrast 3D FSE were: TR=1600 msec, TE_{PD}=13 msec, TE_{T2}=93 msec, ETL=12, Rbw=±31.5 kHz, FOV=230×153.3 mm, acquisition matrix= 384×256, 16 slabs, 6 slices/slab, 0.6 mm slice thickness. The variation of $P_{v,MinIP}$ with respect to *f* for the T2 weighted (T2w) image volume is presented in Fig 1. Theoretically, the smaller the fraction of *k*-space used for reconstruction, the higher is the probability of projecting vessel points in the MIP/MinIP images. However, decreasing *f* causes not only reduction of noise variance but also a loss of image resolution. Therefore, trade-off should be made between the desired $P_{vMIP/MinIP}$ and the MIP/MinIP angiogram resolution. To improve the probability of projection of a single vessel voxel in the path into the MinIP image from the original $P_{v,MinIP}$ =0.0745 to the desired $P_{v,MinIP}$ =0.5 for the studied T2w image volume, *f* was estimated using Eq. 3 to be 0.55. The MinIP images obtained using *f*=1.0 and *f*=0.55 are shown in Fig 2.

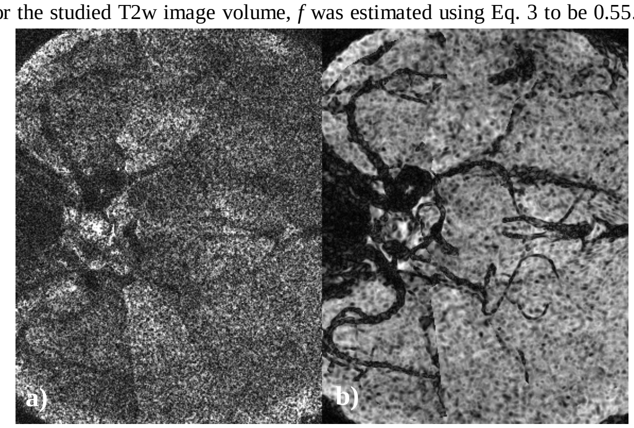

Figure 2. MinIP of T2w images reconstructed with **(a)** *k*=1.0 **(b)** *k*=0.55

Discussion and Conclusions

In this work, we have shown that the knowledge of vessel and background tissues intensity distributions in MRA datasets and the statistical properties of MIP/MinIP algorithms can be exploited to improve the quality of the resulting MIP/MinIP angiograms. This analysis demonstrates that the amount of *k*-space required for optimal visualization of vessels in MIP and MinIP images depends upon the background tissue noise properties and the intensity distribution functions of both vessel and background tissue. Thus, for a specific signal and noise conditions, high resolution acquisition can be made time efficient by acquiring lesser amount of *k*-space and zero-padding to the same resolution.

Acknowledgments

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References

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