EXTENDING HARP IMAGING BY ACQUIRING AN OVERDETERMINED SET OF STRIPES

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Introduction

MR tagging allows the tracking of material points over time. This is of special relevance, for instance, in the analysis of myocardial local motion, whose anomalies are directly related with impaired cardiac function. The HARP method for MR tagging allows the dense reconstruction of the deformation gradient tensor [1]. Nevertheless, without a proper reconstruction scheme, it is prone to be corrupted by noise and phase interferences due to the application of a gradient operator on the reconstructed phase. Here, we propose an extension of this method based on the acquisition of an overdetermined set of stripe patterns, as usually applied in DTI [2], which, especially when combined with our previous contribution in [3], is able to avoid the orientation and spacing dependent phase interferences.

Method

The commonly used SPAMM technique for MR tagging is based on the application of a spatial modulation with wave vector $k_i = k_iu_i$, with $k_i$ its wave number and $u_i$ its orientation. For simplicity, we focus on the reconstruction of 2D HARP images and in-1 SPAMM. Nevertheless, extensions to other scenarios are straightforward.

In this paper we extend the reconstruction equations for the deformation gradient tensor in [1] (where 2 wave vectors are used) in those cases in which a set of 1-2 wave vectors are applied. Let $k_i = k_i(u_i, u_j)$, with $1 \leq i \leq 5$ be this set of vectors. They can be arranged in matrix form as $K = (k_1, k_2, \ldots, k_5)$. Now suppose that $\varphi_i(x)$ is the HARP image for the wave vector $k_i$. The (spatial) deformation gradient tensor $f(x) = \partial \varphi_i(x)/\partial x$ is related with the gradient of the image $\varphi_i(x)$ by $f(x) = \partial \varphi_i(x)/\partial x$.

In this work, we extend this approach to the acquisition of an overdetermined set of stripes. Our methodology is built on the Windowed Fourier Transform (WFT). We propose to use the methodology introduced in [3], based on the Windowed Fourier Transform (WFT).

Results

The simulated deformation proposed in [4] has been used for validation, but applied to a real image in order to better reproduce the spectral content of real data. The image corresponds to a medial slice of the end-diastolic phase of a cine echo-like SSFP MR cardiac acquisition. A GE Genesis Signa 1.5T equipment has been used. Image dimensions are $512 \times 512$ and its resolution is 0.86×0.86×8mm³ with a thickness of 6mm. Some modifications have been performed on the deformation in [4] to consistently apply it to the real image. Specifically, in our case $z=20mm$ and $R_i=60mm$, we have nulled the dependence of $\varphi_i$ on $R$ and $\alpha$. The angular parameter with $R$ for $R_i=60mm$, we have nulled the dependence of $\varphi_i$ on $R$ and $\alpha$. The angular parameter with $R$ for $R_i=60mm$, we have nulled the dependence of $\varphi_i$ on $R$ and $\alpha$. The angular parameter with $R$ for $R_i=60mm$, we have nulled the dependence of $\varphi_i$ on $R$ and $\alpha$.

The synthetic tensors for the pixels inside the myocardium have been used for validation, but applied to a real image in order to better reproduce the spectral content of real data. The filter bandwidth for HARP extraction is selected to be $BW=0.35kHz$ and the analysis window has a Gaussian form with a size of $32 \times 32$ pixels (see [3]). The reconstruction is applied to a set of wave vectors which span the plane uniformly and have a common orientation. Each wave vector is selected to give an image for the wave vector $k_i$, $i=1,2,\ldots,5$.

Discussion and conclusions

A method for the reconstruction of the deformation gradient tensor is presented which builds upon the acquisition of an overdetermined set of stripes in order to limit the influence of the outliers derived from the combination of phase interferences and the gradient order. The method has considerably improved the precision in the estimated tensor for an analytic model of myocardial deformation. We believe that this methodology brings new opportunities in the design of SPAMM acquisition sequences for HARP imaging, especially when combined with modifications of acquisition protocols such as [5]. The overlap introduced by gathering an overdetermined set of stripes can be compensated by the acquisition of a reduced subset of the $k$-space in order to reconstruct the local phase [6]. Finally, we believe that these ideas can also be applied with slight modifications to DENSE or even PC acquisitions [7]. With these considerations in mind, new families of acquisition protocols can potentially be designed for motion sensitive MR imaging, which could simultaneously improve the resolution, robustness and precision in the analysis of motion.

References


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