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Multi-oriented windowed harmonic phase reconstruction for robust cardiac strain imaging☆

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ABSTRACT

The purpose of this paper is to develop a method for direct estimation of the cardiac strain tensor by extending the harmonic phase reconstruction on tagged magnetic resonance images to obtain more precise and robust measurements. The extension relies on the reconstruction of the local phase of the image by means of the windowed Fourier transform and the acquisition of an overdetermined set of stripe orientations in order to avoid the phase interferences from structures outside the myocardium and the instabilities arising from the application of a gradient operator. Results have shown that increasing the number of acquired orientations provides a significant improvement in the reproducibility of the strain measurements and that the acquisition of an extended set of orientations also improves the reproducibility when compared with acquiring repeated samples from a smaller set of orientations. Additionally, biases in local phase estimation when using the original harmonic phase formulation are greatly diminished by the one here proposed. The ideas here presented allow the design of new methods for motion sensitive magnetic resonance imaging, which could simultaneously improve the resolution, robustness and accuracy of motion estimates.

1. Introduction

Magnetic Resonance (MR) tagging allows us to track material points through time. This is of special relevance, for instance, in the analysis of myocardial local motion, whose anomalies are directly related with impaired cardiac function. Hence, local functional indicators extracted from this analysis such as the strain tensor may provide a higher predictive value than global cardiac imaging parameters such as the ejection fraction (Axel et al., 2005; Simpson et al., 2013). The basis of MR tagging consists in the generation of a set of saturated magnetization planes which may be subsequently tracked throughout the cardiac cycle so as to estimate the motion of material points (Shehata et al., 2009; Ibrahim, 2011; Jeung et al., 2012). Usually, these planes are arranged in a parallel setting in such a way that the imaged magnetization is modulated by a given wave vector.

Regarding the analysis of MR tagging, an important family of methods are based on extracting the phase of the complex image obtained by filtering the isolated peaks of the spectrum that correspond to the magnetization modulation. Hence, this class of methods performs motion estimation by phase-based optical flow, in which the constant pixel brightness assumption is replaced by the potentially more reliable constant pixel phase assumption. The seminal work by Osman et al. (2000) shows that this HARmonic Phase (HARP) methodology not only permits to reconstruct small displacements[1] but also to reconstruct the deformation gradient tensor without imposing any condition on the deformation field. Furthermore, this technique is generally faster than intensity-based ones and, in addition, dense measurements can be recovered. Despite its potential, without a proper reconstruction scheme, it is prone to be corrupted by intravoxel phase dispersion, noise, and spectral interferences (especially at the endo- and epicardial boundaries), which are further accentuated in the estimation of the deformation gradient tensor, as it involves the application of a gradient operator on the reconstructed phases. Not coincidentally, a recent study by

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[1] Although, for accurately doing so, motion tracking methods should be adapted to the information provided by HARP.

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Swoboda et al. (2014) has reported a somewhat poor interstudy and intrastudy reproducibility of strain measurements using the HARP method.

A comprehensive description of HARP reconstruction that pays attention to its dynamic range, resolution, noise properties, and characteristic artifacts is provided in Parthasarathy (2006). First, following (Osman et al., 2000), a communications theory description of tagged images is developed which establishes that these images can be interpreted as the spatial analogue of an AM-FM signal so that the tissue strain would be equivalent to the instantaneous frequency of the signal. The author claims that HARP reconstruction is a way to obtain the instantaneous (i.e. local) phase of the signal on the basis of the monocomponent assumption, which states that the signal can be described by a single spectral component that modulates a narrow range of Fourier harmonics that vary as a function of time. With this interpretation in mind, HARP can be understood as a spatial phase demodulation technique and therefore part of the vast literature on temporal phase demodulation could be adapted for motion estimation. In Cordero-Grande et al. (2011), we proposed one technique to demodulate the local phase where we showed the relevance of balancing the spatio-spectral concentration of the HARP filters (trying to follow the spatial variation of the orientation and spacing of the tag pattern) by using the Windowed Fourier Transform (WFT). This Windowed HARP (WHARP) technique turned out to be effective in improving the accuracy in the reconstruction of the local phase. Subsequently, the method has been refined in Fu et al. (2013) by estimating the widths of the WFT on the basis of the instantaneous spatial frequencies as given by a Gabor wavelet transform analysis.

On the other hand, also in Parthasarathy (2006), a characterization of the main artifacts observed in HARP reconstructed images is carried out. In the circumferential strain case, the author describes the causes involved in the presence of a so-called zebra artifact as well as the reason why the radial strain estimation using the conventional HARP method is usually highly inaccurate – limitations in estimating radial strain have also been reported for other methods (Tobon-Gomez et al., 2013). However, the proposed strategy for the improvement of strain estimation is based on postprocessing the strain maps in order to diminish the influence of corrupted estimations—see also Abd-Elmoniem et al. (2006). Here we suggest a completely different methodology for improving the strain reconstruction using the HARP method which resembles previous contributions such as using Complementary SPAtial Modulation of Magnetization (Kuijer et al., 2001) (CSPAMM) or TruHARP (Agarwal et al., 2010) in that it tries to resolve the corrupted areas of the estimated tensor using additional data rather than postprocessing. However, instead of varying the phase of the modulation function to remove spectral peaks interferences, we propose a complementary approach that consists in the acquisition of an overdetermined set of stripe orientations in order to get rid of the orientation dependent phase interferences. This idea has been usually applied in diffusion tensor imaging acquisitions (Papadakis et al., 1999; Jones et al., 1999), where an overdetermined set of gradient orientations have proven to be beneficial in order to diminish the noise in the reconstruction of the diffusion tensor (and mandatory in order to obtain non-Gaussian representations of the diffusion). Preliminary versions of this paper have been presented in the past (Cordero-Grande and Alberola-López, 2012; Cordero-Grande et al., 2014) and some other related techniques have been suggested recently (Bruurmijn et al., 2013); here, we give the full details of the equations involved and provide experimental evidence about the improvement in the accuracy and reproducibility of the strain estimates obtained by the proposed Multi-Oriented Windowed HARP (MOWHARP) method with respect to standard HARP.

In Section 2 we present the theory behind the MOWHARP method; the application of this methodology to real data is described in Section 3; main results regarding the reproducibility of measurements and implications of the experiments carried out are included in Section 4; insight into the practical application of the proposed methodology together with its main advantages and drawbacks are provided in Section 5; and conclusions are established in Section 6.

2. Theory

MR tagging is usually performed by SPAMM (Axel and Dougherty, 1989) or a variant of this technique. SPAMM is grounded on the ability of altering the magnetization of the tissue (within the limitations of relaxation times in MR) even in the presence of motion. The tagging procedure is based on the superposition of a spatial modulation over the applied gradients. To achieve this, a ψ_0 radiofrequency (RF) pulse must be applied, followed by P joint applications of a gradient and another pulse ψ_p. This process will generate a spatial modulation by P sinusoidal functions on the gradient direction. Thus, just after the application of a spatial modulation indexed by i and given by a wave vector $\mathbf{k}_i$, with $\mathbf{k}_i = k_i\mathbf{u}_i$, where k_i is the wave number of the modulation (which depends on the amplitude of the applied gradient) and $\mathbf{u}_i$ is its orientation vector (which corresponds to the orientation of the applied gradient), the image equation is written

$$I_i(\mathbf{X}) = I^0(\mathbf{X}) \sum_{p=0}^{P} c_p \cos(2\pi p \mathbf{k}_i^T \mathbf{X}), \tag{1}$$

where $\mathbf{X}$ denotes the material coordinate, $I^0(\mathbf{X})$ corresponds to the anatomic image, P is the SPAMM order, and c_p is a set of amplitude terms that depend on the train of RF pulses applied. This modulation is translated into a convolution with a Dirac δ distribution in the spectral domain:

$$S_i(\mathbf{k}) = \sum_{p=0}^{P} \frac{c_p}{2} (S^0(\mathbf{k} + p\mathbf{k}_i) + S^0(\mathbf{k} - p\mathbf{k}_i)), \tag{2}$$

with S_i and S^0 being, respectively, the FTs of I_i and I^0. In high order SPAMM the sharpness of the magnetization profiles is controlled by choosing different amplitude terms.

The temporal evolution of a SPAMM image can be written as

$$I_i(\mathbf{x}, t) = I^0(\mathbf{X}(\mathbf{x}, t)) \sum_{p=0}^{P} c_p(\mathbf{X}(\mathbf{x}, t), t) \cos(2\pi p \mathbf{k}_i^T \mathbf{X}(\mathbf{x}, t)) \tag{3}$$

where $\mathbf{x}$ denotes the spatial coordinate, $\mathbf{X}(\mathbf{x}, t)$ denotes the material coordinate $\mathbf{X}$ corresponding to the spatial coordinate $\mathbf{x}$ at time t, and $c_p(\mathbf{X}, t)$ varies in t due to the T_1 relaxation of the tissue at $\mathbf{X}$. For the 1-1 SPAMM acquisition one has that $P = 1$ (Crum et al., 1998) and, assuming $\psi = \psi_0 = \psi_1$,

$$\begin{aligned} c_0(\mathbf{X}, t) &= 1 - \sin^2(\psi) e^{-\frac{t}{T_1(\mathbf{X})}} \\ c_1(\mathbf{X}, t) &= \sin^2(\psi) e^{-\frac{t}{T_1(\mathbf{X})}}. \end{aligned} \tag{4}$$

A commonly used value is $\psi = 45^\circ$ in order to preserve the positivity of the image and maximize the tag contrast. Additionally, in 1-1 SPAMM, the wave number is inversely proportional to the wavelength or spacing of the stripes, $k_i = 1/\lambda_i$.

In this paper we focus on 2D MR HARP images, as 3D acquisitions are rarely used in clinical practice. As stated in Osman et al. (2000), 2D HARP motion reconstruction using the SPAMM technique requires a minimum of 2 linearly independent wave vectors. Our proposal extends the aforementioned HARP methodology for the estimation of the deformation gradient tensor by allowing the application of a set of $I \geq 2$ wave vectors (with a minimum of 2 of them being linearly independent) and present the reconstruction equations in this case. The reconstruction is decomposed in the following steps:

- Calculation of the local phase of the image (Section 2.1). This step is based on the method presented in Cordero-Grande et al. (2011).