Hybrid Intensity- and Phase- Based Optical Flow Tracking of Tagged MRI

Safaa M. Eldeeb, Ayman M. Khalifa, and Ahmed S Fahmy

Abstract— **Accurate tracking of the myocardium tissues in tagged Magnetic Resonance Images (MRI) is essential for evaluating the cardiac function. Current tracking methods utilize either the image intensity or the image phase as landmarks that can be tracked. In either case, the performance is vulnerable to the image quality and the fading of the tag lines. In this work, we propose a hybrid optical flow tracking method that combines both the intensity and the phase features of the image. The method is validated using numerical cardiac phantom as well as real MRI data experiments. Both experiments showed that the proposed method outperforms current intensity-based optical flow tracking and the phase-based HARP method with maximum error of 1 pixel at extreme conditions of tag fading.**

I. Introduction and Background

Tagged Magnetic Resonance Imaging (MRI) is a powerful imaging technique that allows evaluation of the cardiac regional function [1]. In tagged MRI, the magnetization of the heart is spatially modulated such that a pattern of alternating dark and bright lines are temporary created [2]. Accurate tracking of this tagging pattern is essential to quantitatively estimate the myocardium deformation. Two major classes of tracking methods can be found in literature: intensity-based and phase-based. The first class is based on tracking myocardial points using the intensity pattern, or appearance, of the tagged MR images. This class includes a number of techniques such as optical flow tracking [3, 4], manual tracking [5], registration [6, 7], tag detection and tracking [8]. Among these techniques, the intensity based optical flow (IBOF) tracking has gained a lot of interest from researchers due to its simplicity and effectiveness in estimating the motion field. In IBOF, the local distribution of the image intensity is assumed fixed in time [9]. While this assumption is generally valid in many cases, fading of the tag pattern violates this assumption and renders the method inaccurate. Therefore, a number of attempts [10, 11] have been made to model the fading effect. For example, Gupta et al [11] modeled the tag fading using an exponential function whose parameters were estimated using special MRI scan. The latter increases the time and cost of the scan and, in addition, might not be effective in case of inter-breath-hold patient motion. On the other hand Carranza-Herrezuelo et al [3] added a new regularization term to the optical flow equation that includes points with high phase stability in attempt to improve the tracking accuracy. However, Ihor Smal [12] showed that this method is less consistent than phase based methods and also depends on image quality. The second class, namely phase-based techniques [13 -19], assumes that each image point has a unique phase value that is fixed with time and thus can be tracked to estimate thedeformation. The difference among the techniques is the way of estimating the phase at each point. Fleet et al [13] proposed to use steerable complex Gabor filters in the Fourier space to create phase images that can be tracked using optical flow. One of the most commonly used phase-based methods is Harmonic phase (HARP) tracking [14 -17]. The technique makes use of the frequency representation of the tagged image which is composed of a central (DC) peak and a number of harmonic peaks. Theoretically, the latter carries information of tissue displacement and thus is utilized for tracking purposes [16]. In HARP, the phase image is created by applying a bandpass frequency-domain filter to reconstruct a complex image representing the harmonic peak. The phase of this complex image is then unwrapped to create the desired phase image. One of the advantages of HARP is its ability to directly estimate the motion field and the tissue strain tensor [14-17]. Nevertheless, similar to phase-based techniques, HARP is vulnerable to tracking errors at late cardiac phases due to the fading of tag lines. In addition, mistracking of the myocardium borders can be severe due to the discontinuity of the tag pattern at the blood-myocardium border [12]. It has been shown that the phase based techniques outperform the intensity based techniques at the beginning of the cardiac cycle, then due to tag fading effect the accuracy of the phase based methods began to deteriorate and the intensity based techniques become more accurate [20]. In this work we propose a hybrid myocardium tracking technique that combines both intensity-based and phase-based image features. The technique is shown to have superior results to both intensity and phase based tracking techniques. The following section introduces the theory and methods, followed by the results and discussion in Section III.

Safaa M. Eldeeb is with Diagnosoft Inc., USA. Ayman M. Khalifa is with Biomedical Eng. Dept., Helwan University, Egypt and Diagnosoft Inc., USA. Ahmed S Fahmy is with Biomedical Eng. Dept., Cairo University and Center for Informatics Science, Nile University, Egypt.

II. Theory and Methods

II.A. Hybrid OF – Intensity-based component

Let $I(x, y; t)$ represent the intensity of a tagged cardiac MR image pixel (x, y) at time t. Let u and v be the local displacement of a myocardium point (x, y) in the x and y-directions respectively. Then, assuming no temporal variation of the image intensity, the classical IBOF equation is given by [9]:

$$\frac{\delta}{\delta x}I(x,y;\ t)*u+\frac{\delta}{\delta y}I(x,y;\ t)*v=-\frac{\delta}{\delta t}I(x,y;\ t)\quad(1)$$

Where $\frac{\delta}{\delta \mathrm{x}}, \frac{\delta}{\delta \mathrm{y}}$ and $\frac{\delta}{\delta \mathrm{t}}$ are the spatiotemporal derivatives of the image. As mentioned above, fading of the tags violates the intensity constancy assumption, and thus it should be compensated for Eq. (1) to be valid. In this work, simple contrast stretching is applied to each image in the sequence. This is done by applying the following equation:

$$I'(x,y;t)=\frac{I(x,y;t)-min_{x,y}I(x,y;t)}{max_{x,y}I(x,y;t)-min_{x,y}I(x,y;t)}\quad(2)$$

Using Lucas-Kanade solution of the optical flow equation which assumes uniform deformation within a small window of neighborhood N having the same displacement u and v. Let us define an Nx1 vector at time t containing the image intensity values of each tagged MR image at these N points. For the vertical tagged intensity image this vector can be written as

$$\mathbf{I}_{\mathrm{t}}^{\mathrm{V}}=[I^{V\prime}(x_1,y_1;t),\dots,I^{V\prime}(x_N,y_N;t)]^T\quad(3)$$

Where $I^{V\prime}(x,y;t)$is the intensity value of the vertical tagged image at pixel $(x,y;t)$after applying contrast stretching.

Therefore the optical flow constraint equations can be written as

$$\frac{\delta}{\delta x}\mathbf{I}_{\mathrm{t}}^{\mathrm{V}}*u+\frac{\delta}{\delta y}\mathbf{I}_{\mathrm{t}}^{\mathrm{V}}*v=-(\mathbf{I}_{\mathrm{t}}^{\mathrm{V}}-\mathbf{I}_{\mathrm{t-1}}^{\mathrm{V}})\quad(4.a)$$

$$\frac{\delta}{\delta x}\mathbf{I}_{\mathrm{t}}^{\mathrm{H}}*u+\frac{\delta}{\delta y}\mathbf{I}_{\mathrm{t}}^{\mathrm{H}}*v=-(\mathbf{I}_{\mathrm{t}}^{\mathrm{H}}-\mathbf{I}_{\mathrm{t-1}}^{\mathrm{H}})\quad(4.b)$$

Where the last term in the above equation approximates the temporal derivative of the image intensity.

II.B. Hybrid OF – Phase-based component

To create a phase image from a given tagged image, we follow the basic operation of HARP tracking [16]. First, a complex image,$\check{I}(x,y;t)$, representing the first harmonic peak is constructed by applying a band-pass frequency-domain filter. The center frequency and bandwidth of the filter are set equal to $\omega_o$and $\omega_o/2$ respectively, where ω_o is the initial frequency of the tag pattern. Let the constructed complex image is given by,

$$\check{I}(x,y;t)=M(x,y;t)e^{j\emptyset(x,y;t)}\quad(5)$$

where M is the magnitude image and $\emptyset$ is the raw (unwrapped) phase image. For tracking purpose, the latter is wrapped to obtain a phase image, $a(x,\ y;\ t)$ that can be tracked from one timeframe to another [16]. It is worth noting that, in tagged MRI, two image sequences are acquired for the same cross-section with: horizontal tag pattern; and vertical tag pattern. That is, one gets two sequences of phase images to track, namely: $a_h(x,\ y;\ t)$ and $a_v(x,\ y;\ t)$. Because the phase is constant for each tissue point across the different timeframes, then the optical flow constraint equation applies and thus:

$$\frac{\delta}{\delta x}a*u+\frac{\delta}{\delta y}a*v=-\frac{\delta}{\delta t}a\quad(6)$$

Applying the Lucas-Kanade solution of the optical flow equation within a small window of size N, we get,

$$\frac{\delta}{\delta x}\mathbf{a}_{\mathrm{t}}^{\mathrm{V}}*u+\frac{\delta}{\delta y}\mathbf{a}_{\mathrm{t}}^{\mathrm{V}}*v=-(\mathbf{a}_{\mathrm{t}}^{\mathrm{V}}-\mathbf{a}_{\mathrm{t-1}}^{\mathrm{V}})\quad(7)$$

where $\mathbf{a}_{\mathrm{t}}^{\mathrm{V}}$ is an Nx1 vector at time t containing the values of the vertical phase image at the N points. This vector is defined as,

$$\mathbf{a}_{\mathrm{t}}^{\mathrm{V}}=[a_V(x_1,y_1;t),\dots,a_V(x_N,y_N;t)\ ]^T\quad(8)$$

Where a_V is the harmonic phase angle image of the vertical tagging MR image. Similarly for the horizontal phase images, the optical flow constraint equation can be written as

$$\frac{\delta}{\delta x}\mathbf{a}_{\mathrm{t}}^{\mathrm{H}}*u+\frac{\delta}{\delta y}\mathbf{a}_{\mathrm{t}}^{\mathrm{H}}*v=-(\mathbf{a}_{\mathrm{t}}^{\mathrm{H}}-\mathbf{a}_{\mathrm{t-1}}^{\mathrm{H}})\quad(9)$$

II.C. Hybrid OF – Combined components

Both the intensity and phase based optical flow equations (Eq. 4, 7 and 9) can be combined in one matrix-vector form as follows:

$$\begin{bmatrix}\frac{\delta}{\delta x}\mathbf{I}_{\mathrm{t}}^{\mathrm{V}} & \frac{\delta}{\delta y}\mathbf{I}_{\mathrm{t}}^{\mathrm{V}}\\ \frac{\delta}{\delta x}\mathbf{I}_{\mathrm{t}}^{\mathrm{H}} & \frac{\delta}{\delta y}\mathbf{I}_{\mathrm{t}}^{\mathrm{H}}\\ \frac{\delta}{\delta x}\mathbf{a}_{\mathrm{t}}^{\mathrm{V}} & \frac{\delta}{\delta y}\mathbf{a}_{\mathrm{t}}^{\mathrm{V}}\\ \frac{\delta}{\delta x}\mathbf{a}_{\mathrm{t}}^{\mathrm{H}} & \frac{\delta}{\delta y}\mathbf{a}_{\mathrm{t}}^{\mathrm{H}}\end{bmatrix}.\begin{bmatrix}u\\ v\end{bmatrix}=-\begin{bmatrix}\mathbf{I}_{\mathrm{t}}^{\mathrm{V}}-\mathbf{I}_{\mathrm{t-1}}^{\mathrm{V}}\\ \mathbf{I}_{\mathrm{t}}^{\mathrm{H}}-\mathbf{I}_{\mathrm{t-1}}^{\mathrm{H}}\\ \mathbf{a}_{\mathrm{t}}^{\mathrm{V}}-\mathbf{a}_{\mathrm{t-1}}^{\mathrm{V}}\\ \mathbf{a}_{\mathrm{t}}^{\mathrm{H}}-\mathbf{a}_{\mathrm{t-1}}^{\mathrm{H}}\end{bmatrix}\quad(10)$$

Solving this equation for the vector $[u\ v]^T$at each pixel results in the required motion field.

II.D. Validation

The performance of the proposed technique is compared to the intensity based optical flow tracking [9] and the phase-based HARP technique [16].The validation is based on numerical phantom and real data. The details are discussed in the following subsections.

Numerical Phantom

A set of synthetic datasets have been generated using a numerical phantom to evaluate the accuracy of the proposed method at different noise levels. The phantom simulates the contractility function of the myocardium and the tag fading effect (simulation details are discussed in [20]). Fifteen pairs of tagged images (horizontal and vertical tags) with resolution 100×100 pixels are generated to simulate one cardiac cycle. In this simulation, the initial radii of both the epicardium and endocardium equal 50 and 25 respectively, where the endocardium exhibits a radial contraction (minimum radius of 15) and the epicardium remains static (as shown in Fig. 1). To test the effect of the noise, additive white Gaussian noise is added to the images to simulate two SNR-levels (5 and 15dB).

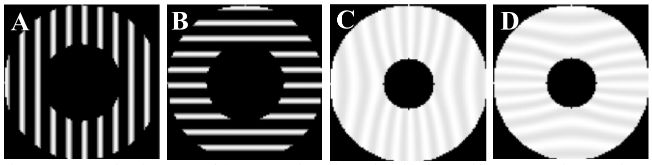

Figure 1 Synthetic tagged images showing the fading effect over time, where (A, B) are the initial, (C, D) are the eighth vertical and horizontal tagged images in the generated sequence (with SNR level = 15 dB).

Three sets of points are selected on the endocardium, mid-wall and epicardium. Each set contains a number of 100 equally spaced points. The tracking error is calculated as the distance between the estimated and the true displacement of each point. The mean and standard deviation (SD) of the error for each of the three sets of points are calculated.

Real Data

A data set of three patients with (*basal*, *mid* and *apex*) short- axis view of the left ventricle (LV) is used to evaluate the introduced technique. Each dataset contains a number of (20 - 25) images per each slice covering an entire cardiac cycle, with image resolution of 256×256 pixels. Datasets were acquired with a pixel size of 1.4 mm and slice thickness 8mm. Image acquisition was performed with a 1.5T MRI scanner. Twenty four points uniformly distributed on the endo-, epi-, and mid-wall contours have been selected and tracked using HARP [16], IBOF [9] and the proposed hybrid OF method. A grader was asked to manually track each point and subjectively classify the tracking accuracy at the end-systole, and late-diastole timeframes. Three classes of errors are selected by the grader as follows. The first class, *Cl_A*, includes points with error less than one pixel (correctly tracked points). The second class, *Cl_B*, includes points with error greater than one pixel but less than two pixels (moderate error). The third class, *Cl_C*, includes points with error more than two pixels. Assuming that points in *Cl_C* are bounded by three pixels error, the average error is calculated as follows:

$$Error = \frac{1.5 * \#Points\ in\ Cl\_B + 2.5 * \#Points\ in\ Cl\_C}{Total\ number\ of\ points}$$

The results are recorded, tabulated and used to compare the three methods.

III. Results and Discussion

The simulation results are summarized in Fig. 2. The figure shows the mean±SD of the error calculated for each technique between the consecutive timeframes. As can be shown, the calculated error of the hybrid OF method is much lower than that of the other methods. For example, at the endo-contour (SNR = 5 dB), the mean error of the hybrid OF between the first and second timeframes equals 0.1 pixels (SD = 0.028) compared to 0.3 pixels (SD = 0.04) for the HARP method and to 1.5 pixels (SD = 0.4) for the IBOF method. This error increases to 1.1 pixels (SD = 0.2) for the hybrid OF method and 3.5 pixels (SD = 0.77) for HARP method and to 7 pixels (SD = 2.5) for the IBOF method at the last two time frames. It can also be observed that the rate of increasing the error with time is much higher in the IBOF and HARP method than the hybrid optical flow method.

The results of the real data experiments were generally found similar to those in the phantom experiment. Table I shows the mean±SD of the error results from tracking the endocardium, mid-wall and epicardium contours using the different techniques. The table shows separate results for the error of the end-systole and late-diastole phases. The HARP technique has shown good performance in both epicardium and mid-wall contours, however its performance have been

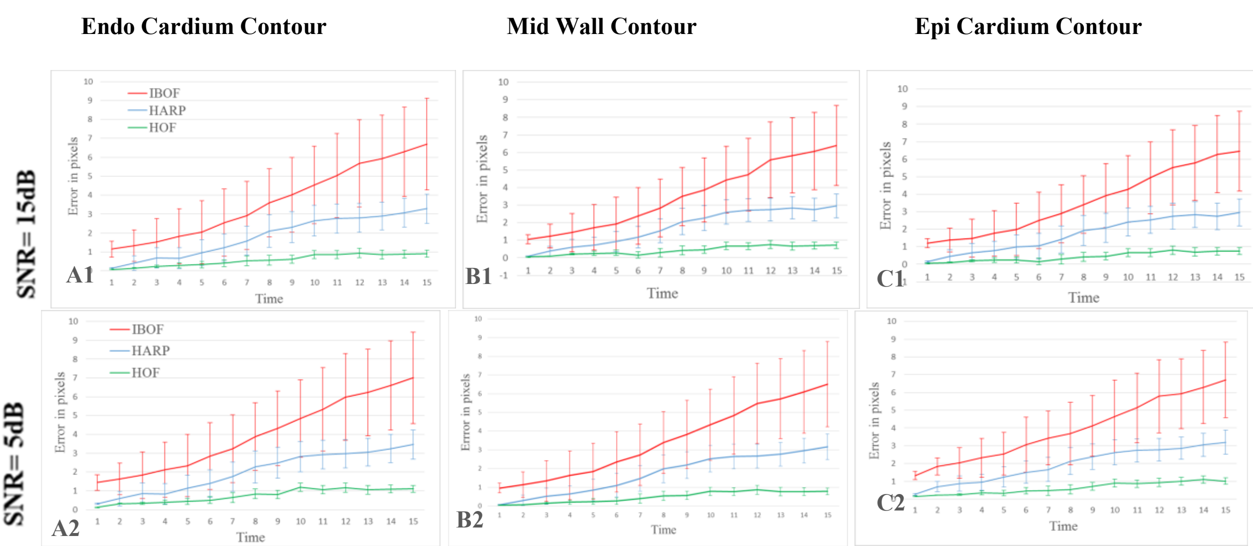

Figure 2 The mean and standard deviation of the error produced by the synthetic data using intensity based optical flow, IBOF, (red color), Harp (blue color) and the hybrid optical flow method, HOF, (green color). This error is calculated at the endocardium (A1, A2), mid wall (B1, B2) and the epicardium (C1, C2) at SNR levels equals 5 and 15 dB.

deteriorated at the endocardium contour. It can also be observed that the hybrid optical flow method results in the lowest error for all contours at the different cardiac phases.

Table I The mean and standard deviation of error of the real datasets

	Optical Flow	**HARP**	**Hybrid OF**
End Systole			
Epi	0.28 ± 0.03	0	0
Mid	0.25	0	0
Endo	0.3 ± 0.062	0.094 ± 0.03	0.0625
Late Diastole			
Epi	0.34 ± 0.09	0	0
Mid	0.53 ± 0.03	0.03 ± 0.03	0
Endo	0.75 ± 0.25	0.34 ± 0.094	0.187 ± 0.031

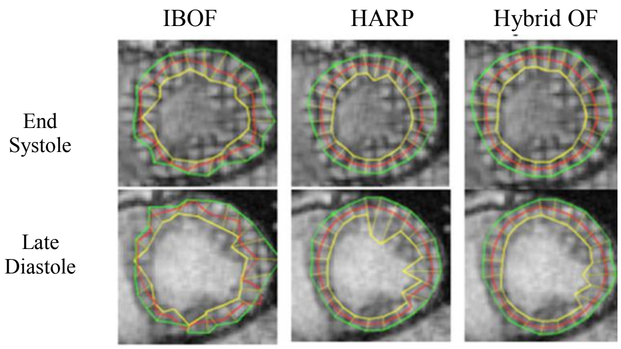

Figure 3 the results of tracking endocardium (yellow color), mid-wall (red color) and epicardium (green) contours at the end systole and late diastole phase using the three methods.

Figure 3 shows a plot of the tracked myocardium contours at both end-systole and late-diastole phases. It can be observed that the IBOF method shows the lowest accuracy among the three methods. The HARP and the hybrid optical flow methods show almost similar results at the mid-wall and epicardium contours. As can be seen, however, the number of mistracked points and the severity of mistracking in the endocardium contours are significantly lower in the hybrid optical flow method compared to the other methods.

IV. Conclusion and Future work

In this work, a method for robust tracking of the myocardium was presented. The method combines both intensity and phase information in an optical flow constraint formulation. It was shown that the proposed method outperforms current intensity-based and phase-based methods.

Acknowledgement

The authors would like to thank Ahmed Dallal for providing the numerical phantom simulation. This work is supported by a grant (PDP2012.R12.4) from ITIDA agency, Ministry of Communication and Information Technology, Egypt.

References

[1] Zerhouni EA, Parish DM, and Rogers WJ, Yang A and Shapiro E,"Human heart: Tagging with MR imaging-a method for noninvasive assessment of myocardial motion," *Radiology*, 1988.

[2] L. Axel and L. Dougherty, "MR imaging of motion with spatial modulation of magnetization," *Radiology*, vol. 171, pp. 841–845, 1989.

[3] N. Carranza-Herrezuelo, A. Bajo, F. Sroubek, C. Santamarta, G. Cristo- bal, A. Santos, and M. J. Ledesma-Carbayo, "Motion estimation of tagged cardiac magnetic resonance images using variational techniques," *Comput. Med. Imaging Graph*, vol. 34, pp. 514–522, 2010.

[4] J. L. Prince and E. R. McVeigh, "Motion estimation from tagged MR image sequences," *IEEE Trans. Med. Imaging*, vol. 11, no. 2, pp. 238– 249, 1992.

[5] AA Young and L. Axel"Three dimensional motion and deformation of the heart wall: estimation with spatial modulation of magnetization – A model based approach" *Radiology*, 1992.

[6] Chandrashekara, R.; Mohiaddin, R.H.; Rueckert, D., "Analysis of 3-D myocardial motion in tagged MR images using nonrigid image registration," *IEEE Transactions on Medical Imaging*, 2004.

[7] D. Rueckert, L. I. Sonoda, C. Hayes, D. L. G. Hill, M. O. Leach, and D. J. Hawkes, "Nonrigid registration using free-form deformations: Application to breast MR images*," IEEE Trans. Med. Imaging*, vol. 18, no. 8, pp. 712–721, 1999.

[8] Guttman, M., Prince, J., McVeigh, " Tag and contour detection in tagged MR images of the left ventricle." *IEEE Trans. Med. Imag.* 1994.

[9] Lucas and T. Kanade, An Iterative Image Registration Technique with an Application in Stereo Vision, Proc. DARPA IU Workshop, 121-130,1981.

[10] J. L. Prince and E. R. McVeigh, "Motion estimation from tagged MR image sequences," *IEEE Trans. Med. Imaging*, vol. 11, no. 2, pp. 238– 249, 1992.

[11] S. N. Gupta and J. L. Prince. "On variable brightness optical flow for tagged MRI", *Proceedings of the 29th Annual Conference of Information Sciences and Systems*, 1981.

[12] Smal, I.; Carranza-Herrezuelo, N.; Klein, S.; Niessen, W.; Meijering, E., "Quantitative comparison of tracking methods for motion analysis in tagged MRI," *IEEE International Symposium on Biomedical Imaging*, 2011.

[13] D. J. Fleet and A. D. Jepson, "Computation of component image velocity from local phase information*," Int. J. Comput. Vision*, vol. 5, no. 1, pp. 77–104, 1990.

[14] N.F. Osman and J.L. Prince, 'Direct calculation of 2D components of myocardial strain using sinusoidal MR tagging', *Proc. of the SPIE Medical Imaging: Image Processing Conference*, 1998, San Diego, California.

[15] N.F. Osman and J.L. Prince, 'Angle Images for Measuring Heart Motion from Tagged MRI', *Proceedings of ICIP98*, Chicago IL, Oct 1998.

[16] N.F. Osman, William S. Kerwin, Elliot R. McVeigh and Jerry L Prince, 'Cardiac motion tracking using CINE harmonic phase (HARP) magnetic resonance imaging', *Magnetic Resonance in Medicine*, 1999.

[17] N.F. Osman and J.L. Prince, "On the Design of the Bandpass Filters in Harmonic Phase MRI," *Proc. IEEE Int'l Conf. Image Proc., Vancouver*, 2000.

[18] Arts, T.; Prinzen, F.W.; Delhaas, T.; Milles, J.R.; Rossi, A.C.; Clarysse, P., "Mapping Displacement and Deformation of the Heart with Local Sine-Wave Modeling," *IEEE Transactions on Medical Imaging*, May 2010.

[19] Xiaomei Yang; Murase, K., "A Multi-Scale Phase-Based Optical Flow Method for Motion Tracking of Left Ventricle," *4th International Conference on Bioinformatics and Biomedical Engineering (iCBBE)*, June 2010.

[20] Dallal, A.H.; Khalifa, A.M.; Fahmy, A.S., "Accurate analysis of cardiac tagged MRI using combined HARP and optical flow tracking," *Cairo International Biomedical Engineering Conference (CIBEC),* 2012.