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Cardiac Motion Estimation based on Transverse Oscillation and Ultrafast Diverging Wave Imaging

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***Abstract*ð Ultrafast ultrasound imaging using plane waves (PW) has demonstrated its potential in assessing complicated motion patterns in the blood or in the tissue. On the other hand, the introduction of transverse oscillations (TO) combined with Phase Based vector Motion estimation algorithms (PBM) has shown to be a very promising technique to improve transverse motion estimation. Cardiac imaging could greatly benefit of a combination of ultrafast TO and PBM. Unfortunately, due to the presence of the ribs, cardiac imaging has to be done with a phased array. Consequently ultrafast imaging of the heart is usually performed with diverging waves (DW) instead of PW. In this paper, the objective is to extend our previously developed ultrafast PW TO technique to ultrafast imaging of the heart using DW. A validation of the method is proposed using CREANUIS simulations with a realistic cardiac sequence.**

Keywords*ð *Ultrafast imaging, Phase motion estimation, Transverse oscillations, Diverging Waves, Cardiac imaging

I. INTRODUCTION

Cardiovascular diseases are the first cause of death in the world. According to the World Health Organisation, 17.5 million people died from cardiovascular diseases in 2012, representing 31% of all global deaths [1]. People with cardiovascular disease or who present a high cardiovascular risk need early detection. Consequently it is fundamental to develop imaging techniques to detect heart diseases. Echocardiography is one of the most useful modality for imaging the heart due to its high temporal resolution, its low cost and because it is a safe diagnostic-imaging modality. The quantification of the heart deformation, the strain, is relevant information to qualify the good functioning of the heart [2].

We have previously developed motion estimation techniques based on plane waves imaging, transverse oscillations and phase-based motion estimators to assess the displacement of the carotid artery wall [3]. The objective of transverse oscillations is to produce a pattern in a direction perpendicular to the wave propagation direction. Thanks to these interference fringes patterns, the phase-based motion (PBM) estimator [4] can quantify the displacement in both axial and lateral directions with a high precision. These motions have to be lower than the half-period of the oscillations generated; this condition is guaranteed thanks to ultrafast imaging.

In this context, we propose to combine ultrafast transverse oscillations imaging and phase-based vector motion estimation algorithms to determine the 2D cardiac motion. Unfortunately, for cardiac ultrasound imaging, we need to use phased array probes because of the ribs. Consequently ultrafast imaging is performed using diverging waves instead of plane waves [5]. Then we developed a polar beamforming technique and applied a filter in the Fourier domain on polar images to produce the TO images. This particular sectorial beamforming allows us to filter the images in regard to the Fraunhofer conditions [6] and to create transverse oscillations. The PBM can finally be applied on these polar filtered images to estimate the motion [7]. We validated this technique in simulation thanks to a realistic cardiac sequence [8] and CREANUIS [9], a software developed at CREATIS.

II. DIVERGING WAVE TRANSVERSE OSCILLATION IMAGING

A. *Diverging wave imaging for Fourier domain filtering*

Diverging waves are used in order to perform ultrafast imaging with a large enough field of view to observe the whole cardiac muscle. From one diverging wave transmission, an image of the heart can be built. We generate this diverging wave assuming a virtual source behind the probe. Applying the appropriate emission delays on each element on the probe, a diverging wave focused on the virtual source is obtained.

Then, receiving the radio-frequency signals coming back from the insonified media, there are several approaches to perform beamforming: we opted for sectorial beamforming considering the virtual source as the origin. The Fraunhofer approximation is respected because the beamforming is made in the focus plane of a converging wave [6]. The whole idea is that in the focal plane, the spatial distribution of the amplitude of the wave is proportional to the Fourier transform of the distribution amplitude of the wave in the media.

Consequently the image is built with a delay and sum sectorial beamforming considering the virtual source as the origin of the polar coordinate system (Fig. 1.). Moreover, this beamforming aims at avoiding the ñsheared appearanceò of the point spread function in the images. Indeed with this beamforming, we preserve the characteristic point spread function pattern which is the same in the whole image.

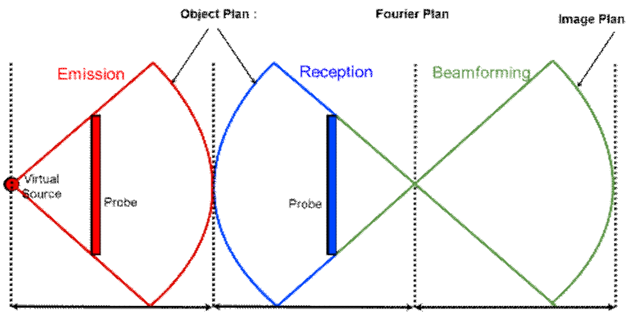

Fig. 1. *Beamforming in a specific polar space represented by analogy with the 4F optical system*

B. TO imaging

Thanks to our specific [R-ɗ] beamforming, the Fraunhofer approximation is respected. So we can apply a filter in the Fourier domain and then take the inverse Fourier transform to get back to the filtered image in the spatial domain. The filter we choose works like Youngôs slits: it is like two diffracting sources which interfere in the image plane to produce a diffraction pattern with transverse oscillations. So the filter is weighting the amplitude distribution of the Fourier spectrum with two peaks. This weighting function w(x) can be expressed in Cartesian coordinates for one depth z and the transverse oscillation wavelength obtained:

$$w(x) = \frac{1}{2}\left(e^{-\pi\left(\frac{x-x_0}{\sigma_0}\right)^2} + e^{-\pi\left(\frac{x+x_0}{\sigma_0}\right)^2}\right)$$

$\lambda_x = \frac{z\lambda}{x_0}$, where λ is the wavelength of the transmitted pulse and +/- x_0 is the position of the peaks.

In the polar space we defined,

$\lambda_\theta = \frac{\lambda}{x_\theta}$, where +/- x_θ is the position of the peaks in the apodization function.

C. Transverse oscillations in polar space

The 2D Fourier transform of a polar transverse oscillation image shows four regions which define the axial and transverse oscillations frequencies. As the images are beamformed in the [R-ɗ] coordinate system described above, the filter operates in the [nu_R nu_ɗ] Fourier domain to generate transverse oscillations. So the axial oscillation frequency is a radial frequency and the transverse oscillation frequency is an angular frequency. Consequently the cardiac motion will be estimated in these two radial and angular directions. It can be noticed that the polar transverse oscillations images here can be achieved solely by filtering in the Fourier domain.

III. SIMULATIONS VALIDATIONS

A. CREANUIS

CREANUIS is a software developed by our laboratory CREATIS. It allows generating ultrasound radio frequency (RF) images considering both linear and non-linear propagation in the media. The non-linear propagation field simulator and our sectorial TO image formation algorithm were coupled to perform the images.

B. Simulations

In order to validate our approach we performed simulations with the CREANUIS software described above and different phantoms motion sequences: a simple numerical square phantom to validate the method and a realistic numerical cardiac phantom. The virtual source was placed 17.6 mm behind the probe in order to insonify the media with a ´ /2 sector. The images were then beamformed in the polar space with 256 angles.

We first used simple square medium in which all the scatterers describe the same constant motion in the angular direction. This phantom was composed of 10^5 scatterers distributed uniformly over a square of 60*20 mm^2 whose top was positioned at 40 mm depth. The amplitude distribution of the scatterers is a unit normal distribution. In order to be as close as possible to the realistic sequences, we applied low motions in the ɗ direction: from 0.01 pixels to 2 pixels, that is to say from 0.0036° to 0.70°. Different approaches were used to generate TO images. We first used a single lateral frequency corresponding to a 5 pixel wavelength (1.78°) with a lateral bandwidth characterized by the sigma value of the filtering. The radial wavelength was equal to 0.308 mm. We then tried another kind of filtering: mono-frequency filtering with no angular or radial frequency bandwidth for each filter. In this case the final motion estimation was obtained by averaging the motion estimations based on different mono-frequency filters placed over the whole initial 2D spectrum.

Finally we made simulation with a realistic cardiac sequence. A simulated sequence was used, generated thanks to a real sequence acquired from a healthy subject. The frame rate of the initial simulated sequence was 33 frames per second. Only the scatterers composing the myocardium were kept and their positions were interpolated to increase the frame rate up to 3301 frames per second. TO images were generated with radial and angular wavelengths equal to 0.308 mm and 1.78°, respectively.

The motion was then estimated with the phase based motion estimator on the entire images to obtain maps.

C. Results

The next figures, show the TO image and its spectrum for the square phantom.

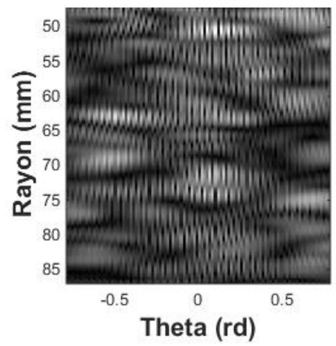

Fig. 3. *TO image B-mode, with λ=5px (1.78°), σ=9px (3.20°)*

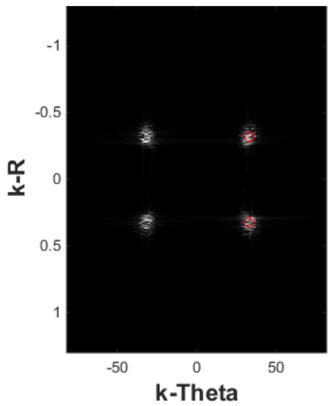

Fig. 4. *TO image spectrum, with λ=5px (1.78°), σ=9px (3.20°)*

Then, the angular motion can be estimated using the PBM described in [4] on the entire image with our sequence from 0.01 pixels to 2 pixels motion. The errors are shown in pixels (Fig. 5.). Big standard deviations of motions estimated are observed here (Fig. 6.). We think this is due to the width of the spectrum which generates random interference patterns with oscillation wavelengths different from the one used in the estimator.

So we had the idea to make motion estimation with an average the estimations obtained with one frequency filter. The comparison of the map obtained is convincing enough to recommend this second approach (Fig. 7.). Very little standard deviations and good estimations can also be observed. Indeed the angular motion error was first 0.057 +/- 0.1[pixels], and with the second method the angular motion error is 0.029 +/- 0.0001 [pixels].

The figure 8 shows our initial result on the realistic cardiac simulation. Here only the approach without filtering could be used. The effect of the width of the spectrum is critical here and we cannot consider that the generated motion map is satisfactory (Fig. 8.). The mono-frequency approach could not be applied yet because a segmentation of the phantom has to be performed first to apply this specific filter on particular region of the phantom.

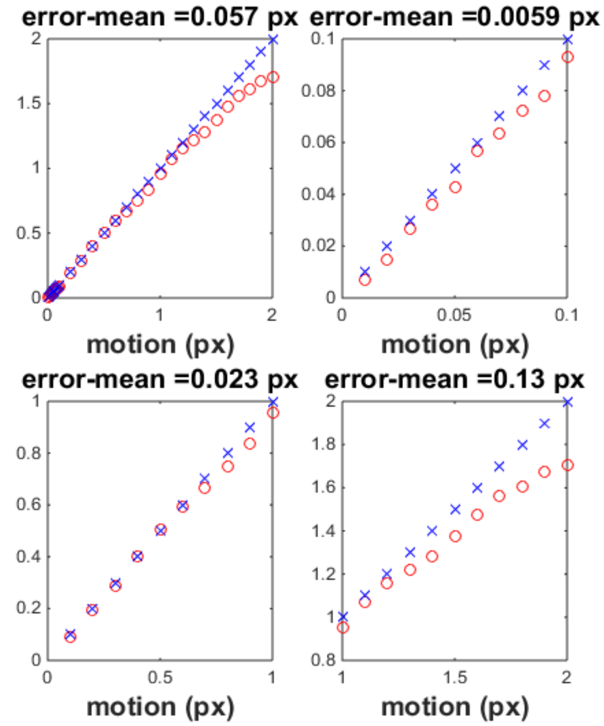

Fig. 5. *Angular motion estimation errors: real motion* **O** *and estimated motion* **X***. The plot at the top left represents all motions from 0.01 to 2 pixels, the next plots shows zoom from 0.01 to 0.1 pixels, from 0.1 to 1 pixel and from 1 to 2 pixels*

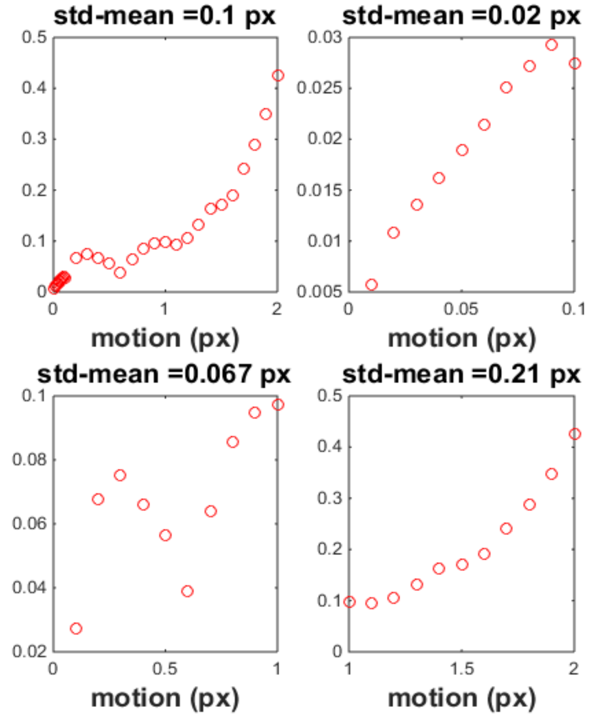

Fig. 6. *Angular* motion *standard deviations. The plot at the top left represents all motions from 0.01 to 2 pixels, the next plots shows zoom from 0.01 to 0.1 pixels, from 0.1 to 1 pixel and from 1 to 2 pixels*

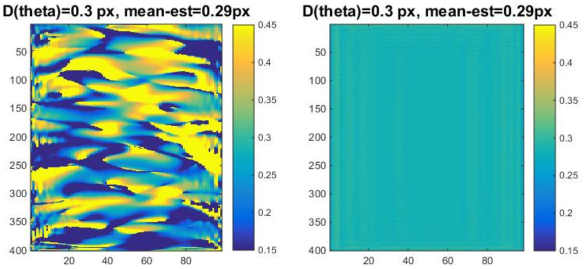

Fig. 7. *Motion estimation map comparison (frequency bandwidth on the left, monofrequency + average on the right)*

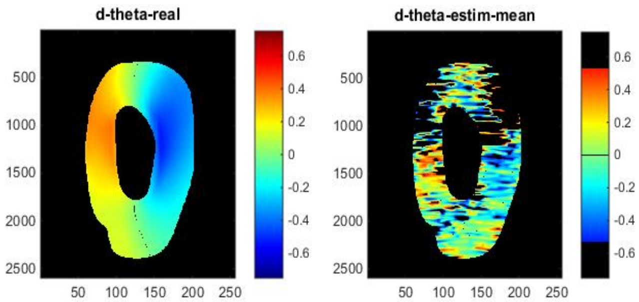

Fig. 8. *Angular Motion: Reference (left) and Estimation (right)*

IV. DISCUSSION

The specific sectorial beamforming technique we propose allows generating images in a polar space without ñsheared appearanceò of the PSF. The filtering approach in the Fourier domain, based on the Fraunhofer diffraction theory, provides transverse oscillation patterns with wavelengths which can be controlled. The wavelength issue is essential because it is the key of the phase based motion estimator. If a bandwidth of the spectrum, centered around a central frequency, is kept the motion estimated can be wrong. Indeed we observe oscillations corresponding to this central frequency but also other random oscillation patterns due to the other frequencies of the spectrum. The consequence of this bandwidth is a big standard deviation in the motion estimated. This effect can be corrected for homogeneous motion by applying filters with only one frequency and averaging estimations with different frequencies spread over the full spectrum. The advantage of this approach is the use of the whole spectrum instead of choosing one central frequency. Moreover with this method the motion estimated standard deviation is much reduced. We have to notice that this approach is more time consuming than the ñbandwidth approachò but it should also be easily parallelized.

For the cardiac application, the motion of the muscle is not homogeneous and the method has to be improved. The model of the square doing only translation in the radial or transversal direction is too simple to be reproducible for the heart motion. So the next stage to validate the model will be an intermediate situation and should consist in applying our method to phantom with complex circular motions.

Finally, for the cardiac application the method will have to be implemented in 3 dimensions because the heart motion is not constant in a 2D plan.

V. CONCLUSION

In this paper we presented a new approach to combine transverse oscillation and ultrafast imaging using diverging waves. We performed a specific sectorial beamforming which allows filtering in the Fourier domain. The transverse oscillations are generated in an angular direction and the motion estimation is performed in the polar space. The simulations showed good results for simple translations in the angular transverse direction. Realistic heart motion models are for now too complex to take advantage of our method. But there are some ideas of better use of the spectrum of the image to improve the method. It is now important to pursue simulations with more complex motions and to apply the technique on real phantom with control motions.

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