3D ultrasound imaging of tissue anisotropy using spatial coherence: comparison between plane and diverging waves

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***Abstract*—The heart, like skeletal muscles, is made up of muscular cells arranged regularly and packed into sort of bundles commonly called cardiac fibers. After a myocardial infarction (MI) both during the acute phase and the reperfusion of the myocardium, several cells die and some local fibers orientation modifications appear. Consequently, the heart does not contract properly anymore. The evaluation of the fibers orientation and more particularly the modifications of their orientation can be useful to evaluate the extent of the lesion and to adapt the treatment after a MI. The purpose of our work is to develop ultrasound based imaging techniques to determine the cardiac fibers orientation in order to be able *in fine* to determine fibers orientation changes that appear due to MI. Recently, a fiber orientation ultrasound imaging technique has been proposed based on the spatial coherence of backscattered signals. In this paper, we study the spatial coherence technique in ultrafast imaging. We evaluate the feasibility to extend the previous published technique to diverging waves instead. The purpose is to extend the field of view which is typically reduced in cardiac imaging due to the limited ultrasound window possible between the ribs. Experimental data were acquired with a 1024 channels system and a 32x32 matrix array on a phantom constituted by seven wires at different orientations as a function of depth. These first results obtained with two different transmission strategies are compared and are in favor of diverging waves.**

Keywords—ultrasound; heart; fibers; spatial coherence

I. INTRODUCTION

Skeletal muscles are made up of fibers allowing them to contract. Similarly, the heart is also constituted of fibers made of cardiomyocytes. These cardiac cells are elongated and oriented along the fibers direction. Contrary to skeletal muscles, cardiac fibers orientation changes as a function of location and particularly the depth inside the heart tissue [1] which allows the heart contraction. During a myocardial infarction (MI), one of the coronary arteries supplying blood to the myocardium gets blocked leading to cellular death in a part of the heart. This cellular death induces a modification of local fibers orientation and leads to anomalies in the heart contraction [2]. With a proper evaluation of the fibers orientation changes, the extent of the lesion caused by the MI could be quantified and the treatment better adapted. The purpose of this work is to develop an ultrasound based imaging method able to render the local fibers orientation *in vivo*. This preliminary step should lead, *in fine*, to an imaging mode providing the modification of fibers orientation in patients during and after MI.

In this field, diffusion MRI is the reference [1] but, due to its long acquisition time (approximately 25 minutes in a clinical setup), it is difficult to make an *in vivo* moving heart image. On the other hand, the fast and even ultrafast acquisition modes produced with the most recent ultrasound imaging sequences represents a very attractive alternative given that it could render somehow the local tissue orientation. Three methods based on ultrasound imaging have been developed with this objective: (i) the backscattered coefficient has been used [3], (ii) the shear wave velocity [4] and finally (iii) the spatial coherence [5]. In this work, we have chosen to work with the spatial coherence given its simplicity and the good results shown in the literature.

The spatial coherence has been first developed with focalized transmissions [6]. However, in order to image a moving heart in 3D, such a transmission (TX) scheme is not realistic given the number of TX that would be required to cover the whole 3D heart volume. For these reasons, plane wave (PW) TX have been proposed and it has been shown that with such approach a minimal TX time is necessary to provide the fibers orientation map [5]. However, based on PW, the field of view is limited by the small probe footprint. On 2D probes, such dimension is small in order to be able to image the heart between the ribs. We propose to extend the field of view by replacing PW with diverging waves (DW). We show the feasibility of such DW insonifications for cardiac fiber imaging. Results obtained with both PW and DW are studied and compared in this work.

The spatial coherence method is detailed in the next section and followed by the results description.

II. METHOD

Spatial coherence can be mathematically expressed in several forms. In this work, the coherence function is used [5].

A. *Coherence function*

Initially, spatial coherence was studied by A. Derode and M. Fink in [6] with focused waves. They showed that to calculate spatial coherence in each point in the medium, it is necessary to focus the ultrasound signals at each point, both in transmission and in reception. Moreover, by using spatial coherence calculation, it is possible to determine the local orientation in each point. This is linked with the presence or not of some coherence in the backscattered field [6]. The drawback for *in vivo* applications is the need to transmit a different focused wave for each point in the medium where the orientation should be evaluated. In a moving heart such approach is not relevant. Recently, with the emergence of high frame rate acquisition modes able to provide a synthetic transmit focusing in the whole medium with much less transmissions [7], spatial coherence techniques have regained interest. Ultrafast methods are based on broad plane or diverging wave transmissions. The use of plane waves in spatial coherence calculation requires creating a synthetic focusing in emission in order to calculate the spatial coherence. The advantage is that with around forty transmissions only, it is possible to create a focusing in transmission and reception in the whole medium and to obtain the same results as with focused transmissions but in much less time. This makes plane or diverging waves much more suitable for clinical applications.

In order to create this synthetic focusing, it is necessary to use coherent plane waves compounding [7]. The first step is to calculate the delays for each plane wave on each probe element for a point in the medium [7].

These delays are then applied on raw signals received by each probe elements to be rephased and focused in reception. After delays application, these raw signals are already focused in reception. This calculation is realized for each plane wave transmitted with an angle α (Fig. 1). Next, in order to create a focusing in emission, these receive-focused raw signals from all different plane waves are added up as represented in Fig. 2.

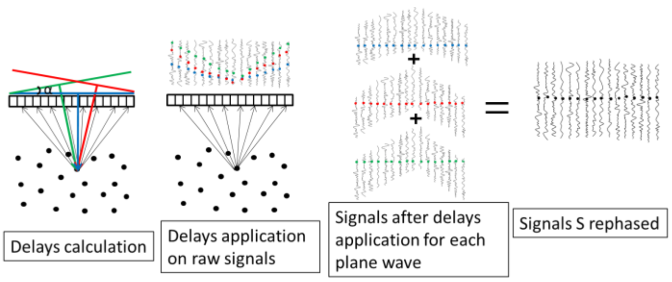

Figure 1 – Plane waves coherent compounding in order to create a synthetically focusing in emission and in reception.

The signals obtained are focused in transmission and reception for any point in the medium. They are used in the spatial coherence calculation.

The coherence map is obtained from the calculation of intercorrelation coefficients. It involves comparing signals received by the probe elements (1) after their focalization in both transmission and reception. For 2D coherence function evaluation, signals are correlated two by two as a function of their distance in the two lateral directions, x and y, respectively. For each inter-elements distance, all possible correlation coefficients between pairs of signals from elements separated by the same [Δx;Δy] values are averaged. A coherence map is obtained and can be expressed as a function of the inter-elements distance:

$$R(\Delta x,\Delta y)=\frac{1}{N-|\Delta x|}\frac{1}{N-|\Delta y|}\sum_{i=1}^{N-|\Delta x|}\sum_{j=1}^{N-|\Delta y|}\frac{\sum_{t=T_1}^{T_2}S(i,j,t)*S(i+\Delta x,j+\Delta y,t)}{\sqrt{\sum_{t=T_1}^{T_2}S(i,j,t)^2*\sum_{t=T_1}^{T_2}S(i+\Delta x,j+\Delta y,t)^2}} \quad (1)$$

where $S(i,j)$ is the focused signal received by the element number i(j) in x(y) direction, N is the number of signals in x and y directions, considered in the coherence function calculation, [T_1 T_2] is the time window and Δx(Δy) are the distance in x(y) direction between two correlated signals.

In this way, if the medium is anisotropic, that is to say, it is more coherent in one specific orientation, then, the signals correlation is high in the privileged direction and decreases in the other directions. On the contrary, if the medium is isotropic, there is no particular coherence between the signals and their correlation is low. The coherence function is low for all direction in the medium [5, 6].

In a "coherent" medium, the coherence map exhibits an ellipsoidal shape in a favor orientation. In fact, the coherence function is high in this main direction and decreases quickly in the other directions. In contrary, in an "incoherent" medium, the coherence function is more circular or can present a diamond shape because there is no privileged orientation. The coherence function decreases quickly in all directions. In a "coherent" case, the ellipse orientation indicates the fibers orientation. By determining the ellipse direction, the fibers angle can be deduced. Fig. 2 represents an example of coherence function in a "coherent" or anisotropic medium and in an "incoherent" or isotropic medium.

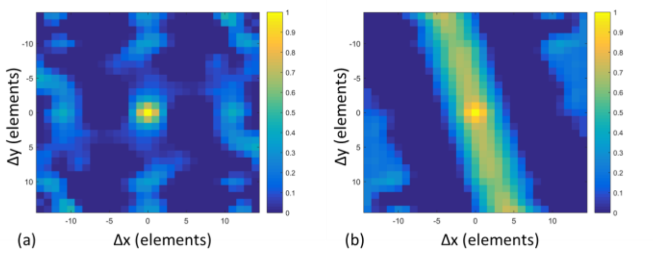

Figure 2 – Coherence function in 2D in a point of an (a) isotropic medium and (b) anisotropic medium.

This procedure is repeated for each point in the medium in order to extract fibers angle at each point. So, by calculating the coherence function and according to its appearance (circle or ellipse) it is possible to determine the medium local nature (isotropic or anisotropic) and the privileged direction of the medium.

The principle of this method is the same for diverging waves because the only modification is the virtual source position and the computed delays in the beamforming to create the focalization in both emission and reception. The determination of cardiac fibers orientation using ultrasound diverging waves has never been reported in the literature, only plane waves have been used in [5].

B. *Experimental set-up*

Data were acquired on a phantom (Fig. 3) constituted by seven wires oriented in different directions from [-45°;45°] as a function of the depth. The diameter of each wire is 0.3 mm. This phantom was immersed in agar at 4% in order to have speckle around wires.

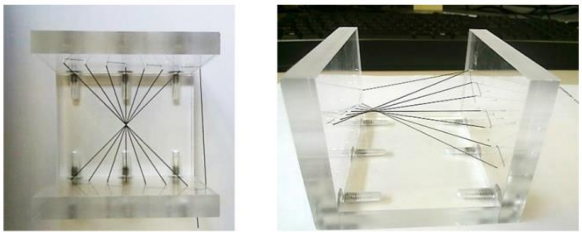

Figure 3 – Top and side view of the wires phantom consisting of 7 wires used to have data with different orientations.

Four research systems Verasonics Vantage 256 (Verasonics Inc., USA) that allow controlling the 1024 elements of a matrix array, both in transmission and in reception were used [9]. These systems were associated with a 3 MHz 32x32 matrix probe (Vermon, Tours, France) with a pitch of 0.3 mm in both directions.

Data were acquired with transmissions of plane and diverging waves in the same location in order to compare results obtained with these two emissions. The same parameters were used for the two kinds of waves. The ultrasound frequency was 3 MHz and the sampling frequency was 12 MHz. Two angles have been defined to send a titled wave, an angle in x direction and another in y direction. In these two cases, 5 waves from [-10°;10°] in x direction and y direction were combined for a total of 25 transmit events. The virtual source in diverging waves was computed to have a field of view of 90°, leading to a position of 4.8 mm behind the probe.

III. RESULTS

Data were acquired with plane waves and diverging waves for the same location to evaluate the accuracy of the coherence calculation as a function of the emission type. By using the method described in the previous section, the 2D coherence function was calculated on each wire in plane and diverging waves. A number of N=15 signals and a time window of five periods were used for the coherence function calculation. Results obtained with diverging waves are presented in Fig. 4. On these different maps, the wire orientation is clearly visible through the ellipsoidal shape. Then, the angle of each wire was determined and was plotted in Fig. 5 for the two kinds of waves.

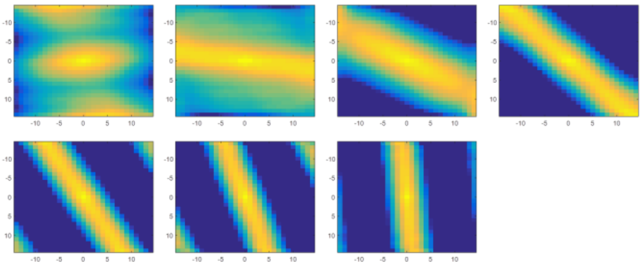

Figure 4 – Coherence functions for a point in the center of the probe corresponding to the depth of each wire of the wire phantom. Results are obtained in diverging waves.

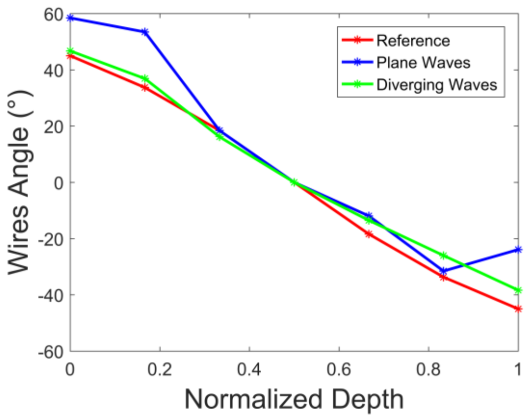

Figure 5 – Curves of the estimated wire angles from the wire phantom in diverging waves (green), plane waves (blue) and the reference (red).

Results obtained with plane and diverging waves are close to the reference even if results in diverging waves are closer to the reference than results in plane waves. The root means square error (RMSE) is 4.8° for diverging waves and 11.2° for plane waves.

IV. DISCUSSION AND CONCLUSION

Results obtained with plane and diverging waves are close to the reference as justified by the obtained RMSE. However, results obtained with diverging waves are slightly better than results with plane waves. Moreover, the advantage of diverging waves is that it should enable to acquire a larger field of view that is more suitable for clinical applications. That is why for future work, diverging waves will be used to study muscle and heart fiber orientation.

In transmission, an angle of 10° in plane waves is a maximal limit. If higher angles are used there is not enough energy for deep penetration depth. For diverging waves, the number of waves, the maximal angle and the virtual source position should be studied more deeply to evaluate the variation on results with these different parameters. For example, these parameters in diverging waves should be studied to create the same synthetic focalization as in plane waves.

The next step of this work is to study more complex medium with known reference. Then, biological tissue such as muscle will be studied *ex vivo* and then *in vivo*. Finally, the last step is to study heart first *ex vivo* and then *in vivo*. The difficulty of *in vivo* heart acquisitions will be its movement. It will probably be necessary to study how to take this aspect into consideration. For example, it could be necessary to study the number of plane waves to make acquisition in less time to not be disturbed by heart movement [8].

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