Full 3-D Transverse Oscillations: A Method for Tissue Motion Estimation

Sebastien Salles, Hervé Liebgott, Damien Garcia, and Didier Vray

Abstract—We present a new method to estimate 4-D (3-D + time) tissue motion. The method used combines 3-D phase-based motion estimation with an unconventional beamforming strategy. The beamforming technique allows us to obtain full 3-D RF volumes with axial, lateral, and elevation modulations. Based on these images, we propose a method to estimate 3-D motion that uses phase images instead of amplitude images. First volumes featuring 3-D oscillations are created using only a single apodization function, and the 3-D displacement between two consecutive volumes is estimated simultaneously by applying this 3-D estimation. The validity of the method is investigated by conducting simulations and phantom experiments. The results are compared with those obtained with two other conventional estimation methods: block matching and optical flow. The results show that the proposed method outperforms the conventional methods, especially in the transverse directions.

I. INTRODUCTION

Motion estimation is an essential part of ultrasound imaging. It can be used in various applications such as tissue motion tracking [1], speckle suppression [2], motion-based image segmentation [3], elastography [4], blood flow velocity estimation [5], [6], or strain rate imaging [7].

Several methods for 2-D ultrasound tissue motion tracking and blood velocity estimation have been proposed and clinically applied, including speckle tracking [8], optical flow [9], and the border tracking method [10]–[12]. Doppler imaging is widely used for blood flow measurements, and tissue Doppler imaging has been described in several studies [13], [14].

Conventional Doppler techniques have inherent sources of error, including aliasing, frequency-dependent attenuation, and most importantly, the inability to detect nonaxial motion, i.e., motion perpendicular to the beam axis. On the other hand, speckle tracking and optical flow methods do not have these limitations. However, these 2-D approaches are limited to measuring in-plane motion and are prone to errors due to out-of-plane motion, which are very often present in vivo. This limitation can be overcome using 3-D ultrasound imaging.

With the emergence of ultrasound volumetric imaging, a method that is capable of obtaining the complete 3-D motion vector would potentially be of high clinical interest. Several ultrasound methods have been proposed for estimating 3-D motion over the past few decades. These include nonrigid image registration methods [15] and different speckle tracking methods [16]–[22]. These studies confirm the ability of 3-D speckle tracking to obtain more complete 3-D motion information. However, despite significant algorithmic advances, 3-D motion estimation remains challenging because of the limited field of view, low image quality, large computation load, and difficulty in estimating a motion in the plane perpendicular to the beam axis.

In 2-D, to increase the definition of the motion estimation in the lateral directions, several methods based on transverse oscillations (TO), which were initially proposed for blood flow imaging [23], have been proposed to address multiple applications in medical imaging. Since the seminal papers, in the field of blood flow estimation, several studies have presented improved techniques to assess the velocity vectors [24]–[26]. In elastography, the use of TO has been investigated to enhance the estimation of local tissue deformation [27]–[30]. In echocardiography, recent studies have presented techniques to assess the motion and deformation of the heart wall [31]–[33]. As for tissue motion estimation, the US tagging technique has been investigated to assess the displacement of the carotid wall, especially in the lateral direction [34].

Finally, the TO method has been recently extended in 3-D by J. A. Jensen’s group [35], [36]. This method enables one to obtain two separate ultrasound fields, each one featuring oscillations along the axial dimension plus one perpendicular (lateral or elevation) direction. This pseudo-3-D technique consists of a twice-performed 2-D. Here two different apodization functions must be used for generating both elevation and lateral oscillations.

This paper presents a method able to create a volume featuring full 3-D oscillations in real time, using a single apodization function and the corresponding 3-D motion estimation method. The 3-D motion vector estimation is
the 3-D application of the N-D method proposed by Basarab et al. [37]. It is based on the phases of such images, which offers high computational speed. The feasibility of the proposed method has been investigated and presented in [38]. In the present study, a more complete simulation and in vitro validation of the full 3-D TO method are presented.

The paper is organized as follows. Section II provides the TO concept and the proposed 3-D extension in 2-D linear scanning as well as pyramidal scanning. Section III describes 3-D phase-based motion estimation (PBME). Section IV presents two standard and state-of-the-art methods that were implemented for comparison purposes. Simulation and experimental results are presented in Section V and Section VI, respectively. Before the conclusion given in Section VIII, the results and parameters of our estimation are discussed in Section VII.

II. The 3-D Transverse Oscillation Method

Instead of designing the 3-D TO fields as proposed in [35], [36], creating oscillations in the axial and transverse directions one at a time (first lateral and then elevation), we propose to build a 3-D TO field with oscillations in the axial and in the two transverse directions simultaneously. This section describes how the 3-D TO field is obtained in case of 2-D linear scanning and 2-D sectorial scanning.

A. 3-D Apodization Function for 2-D Linear Scanning

To obtain 3-D ultrasound RF images with TO, the point spread function (PSF) must show oscillations in the two transverse directions. Moreover, the beamforming strategy should be able to produce PSFs with the same shape in the entire imaged medium to get consistent motion estimates. Let \( h(x, y, z) \) be the 3-D PSF, with \( z \) the axial direction along the beam axis, and \( x \) and \( y \) the lateral and elevation directions, respectively, both perpendicular to the beam axis. In this paper the PSF is approximated as a separable PSF both spatially and in transmit/receive, as given in (1). The transverse radiation patterns are equal to the pattern in transmit multiplied by the pattern in receive as in [39] and [40].

\[
    h(x, y, z) = h(x) \times h(y) \times h(z) \\
    = h_t(x) \times h_t(x) \times h_t(y) \times h_t(y) \times h(z),
\]

where \( h_t(x) \), \( h_t(y) \), and \( h_t(z) \), are the transmit and receive transverse profiles and \( h(z) \) is the pulse echo axial profile. Given that the TO method mainly consists in introducing oscillations in the transverse dimensions, we only consider the transmit/receive separation of the transverse profile in this model. As a result, \( h(z) \) is considered to be the pulse echo axial profile. As is the case in 2-D, 3-D TO PSFs are based on axial and transverse modulations limited by axial and transverse windows, as given by

\[
    h(x, y, z) = \prod_{k=\{x,y,z\}} \text{mod}_k(k),
\]

where \( \text{mod}_d() \) are \( \cos() \) or \( \sin() \) functions and the amplitude-modulating function varies from one author to another. In this paper we will consider

\[
    h(x, y, z) = \prod_{k=\{x,y,z\}} e^{-\pi(k/\sigma_k)^2} \cos\left(\frac{2\pi k}{\lambda_k}\right),
\]

where \( \lambda_x \) (respectively, \( \lambda_y \) and \( \lambda_z \)) and \( \sigma_x \) (respectively, \( \sigma_y \) and \( \sigma_z \)) are the lateral (respectively, elevation and axial) oscillation wavelength and standard deviation of the Gaussian envelope.

The modulation naturally present in the axial PSF is related to the excitation pulse and the impulse response of the transducer elements. The transverse profile of the PSF can be controlled in transmit and receive by adapting the delays between the elements and the weighting coefficients applied to each element.

The Fraunhofer approximation is classically used to design these aforementioned parameters. Under this approximation, the transverse profile of the PSF at the focusing depth is related to the apodization function by a Fourier transform [41]:

\[
\begin{align*}
    h(x) &\sim \text{FT}\left\{ w \left( \frac{x}{\lambda_x} \right) \right\} \\
    h(y) &\sim \text{FT}\left\{ w \left( \frac{y}{\lambda_y} \right) \right\},
\end{align*}
\]

where FT represents the Fourier transform, \( w \) is the apodization function, and \( x_s \) and \( y_s \) are the lateral and elevation distances in the aperture domain.

Another approach to design the apodization function leading to transverse oscillations was presented in [42]. In this paper, the approach to obtain TO is based on the emission of a plane wave, which is assumed to affect the transverse PSF profiles as little as possible to control the transverse PSF profiles only during receive. In this case, one transverse profile of the PSF, represented in Fig. 1(a), can be obtained by amplitude-modulating a cosine with a Gaussian, as given by

\[
\begin{align*}
    h(x) &= h_t(x) \times h_t(x) = 1 \times \cos(2\pi x/\lambda_x) \times e^{-\pi(x/\sigma_x)^2} \\
    h(y) &= h_t(y) \times h_t(y) = 1 \times \cos(2\pi y/\lambda_y) \times e^{-\pi(y/\sigma_y)^2}.
\end{align*}
\]
Fig. 2. Apodization functions in receipt. Each small square represents a lateral and elevation directions.

A. Algorithm Description

In this section, the description of the 3-D phase-based block matching method is proposed. This technique is the 3-D extension version of the method proposed by Basarab et al. [22]. A pair of 3-D RF volumes of interest with TO, \(i_1(x, y, z)\) and \(i_2(x, y, z)\), representing the same medium before and after the application of a 3-D translation, are considered. The relation between the two volumes is

\[
i_1(x, y, z) = i_2(x + d_x, y + d_y, z + d_z),
\]

where \(d_x\), \(d_y\), and \(d_z\) are the spatial shifts to be estimated in each pixel \((x, y, z)\).

B. Phase of Analytical Signal

In contrast with classical block matching, the proposed method uses phase images instead of amplitudes, and is motivated by the fact that phase information is more stable under changes in intensity and contrast than am-

Fig. 1. Representation of one transverse PSF profile along the \(x\) direction (a) and the corresponding apodization function (b).

with depth. To avoid this behavior, apodization must be adapted dynamically. The receive apodization function can be calculated using (4) and (5). This results in a bimodal Gaussian function with two distinct peaks [(6) below and Fig. 1(b)].

\[
w_{x,p}(x) = \frac{1}{2} \left( e^{-\pi(x-x_0/\sigma_x)^2} + e^{-\pi(x+x_0/\sigma_x)^2} \right),
\]

\[
w_{y,p}(y) = \frac{1}{2} \left( e^{-\pi(y-y_0/\sigma_y)^2} + e^{-\pi(y+y_0/\sigma_y)^2} \right),
\]

where

\[
\begin{align*}
x_0 &= \lambda_\theta / \lambda_r, \quad y_0 = \lambda_\phi / \lambda_r, \\
\sigma_x &= \sqrt{2}\lambda_\theta / \sigma_{\theta}, \quad \sigma_y = \sqrt{2}\lambda_\phi / \sigma_{\phi},
\end{align*}
\]

and \(\lambda\) is the wavelength of the transmitted pulse, \(z\) is the depth of interest, \(\pm x_0\) and \(\pm y_0\) are the positions of the two peaks in lateral and elevation directions, respectively, and \(\pm \sigma_x\) and \(\pm \sigma_y\) are the standard deviation of each peak in lateral and elevation directions. \(w_{x,y}(x, y)\) represents the 2-D apodization function resulting from the multiplication of both apodization functions (Fig. 2).

B. 3-D Apodization Function for Pyramidal Scanning

The 3-D TO method requires a 2-D array transducer. The main issue with 2-D linear scanning using the current 2-D transducers is the small volumes that can be acquired due to both the limited footprint and number of array elements. For a larger field of view, each transverse direction can be scanned using a sector approach, also called pyramidal scanning. However, the actual field of view is still limited and the insonation volume is determined by the size of the aperture.

The TO beamforming framework introduced above can be easily adapted to pyramidal scanning using the same apodization function as the example given in Fig. 2. Indeed, the transformation of the model parameters from Cartesian to polar coordinates is straightforward. This is done by setting

\[
\begin{align*}
\lambda_\theta &= \lambda \vartheta, \quad \lambda_\phi = \lambda \varphi, \\
\sigma_\theta &= \sigma \vartheta, \quad \sigma_\phi = \sigma \varphi, \\
z &= r
\end{align*}
\]

where \(\lambda_\vartheta\) and \(\lambda_\varphi\) are the expected wavelengths of the transverse PSF profiles in radians, \(\sigma_\theta\) and \(\sigma_\phi\) are the stan-

\[
\begin{align*}
x_\vartheta &= \lambda \vartheta / \lambda_r, \quad x_\varphi = \lambda \varphi / \lambda_r, \\
\sigma_\vartheta &= \sqrt{2} \lambda \vartheta / \sigma_\theta, \quad \sigma_\varphi = \sqrt{2} \lambda \varphi / \sigma_\phi,
\end{align*}
\]

where \(\lambda\) is the wavelength of the transmitted pulse. In pyramidal scanning, \(x_\vartheta\), \(x_\varphi\), \(\sigma_\vartheta\), and \(\sigma_\varphi\) become independent of depth and are only linked to the values of the lateral wavelength and standard deviation of the Gaussian envelope in radian (7). This is a significant difference with linear scanning. Therefore, in the polar coordinate system corresponding to the acquisition, a 2-D dynamic apodization function is not required to obtain uniform PSF profiles throughout depth.

III. THE 3-D PHASE-BASED MOTION ESTIMATION METHOD
plitude information. Several methods to extract the 2-D local phase images can be found in the literature. The first method was introduced by Hahn [43] where 2-D complex signals (the 2-D analytic signals) are calculated in the complex Fourier domain. In addition, a second method was introduced by Bülow and Sommer [44], where 2-D hypercomplex signals (the quaternionic signals) are calculated in the hypercomplex Fourier domain using the 2-D quaternionic Fourier transform and which are conducted to the monogenic signal introduced by Felsberg and Sommer [45]. Considering that the volumes of interest defined in this paper feature sinusoidal profiles in the three spatial directions, the 3-D local phase images are extracted using the 3-D analytic signal approach [46].

The proposed method uses six phase volumes, which are obtained using three single-octant analytical signals calculated from $i_1$ and $i_2$ and defined in the Fourier domain as:

$$I_{n1}(f_x, f_y, f_z) = I_n \cdot (1 + \text{sgn}(f_x))(1 + \text{sgn}(f_y))(1 + \text{sgn}(f_z))$$

$$I_{n2}(f_x, f_y, f_z) = I_n \cdot (1 + \text{sgn}(f_x))(1 - \text{sgn}(f_y))(1 + \text{sgn}(f_z))$$

$$I_{n3}(f_x, f_y, f_z) = I_n \cdot (1 + \text{sgn}(f_x))(1 - \text{sgn}(f_y))(1 - \text{sgn}(f_z)),$$

with $n = \{1, 2\}$, and where

$$\text{sgn}(u) = \begin{cases} 1, & \text{if } u > 0 \\ 0, & \text{if } u = 0 \\ -1, & \text{if } u < 0 \end{cases}$$

$f_x$, $f_y$, and $f_z$ correspond to the modulation frequencies in the respective direction, and $I_1$ and $I_2$ are the 3-D Fourier transforms of the ultrasound volumes $i_1$ and $i_2$. Then the analytic signals $I_{n1}$, $I_{n2}$, and $I_{n3}$ are obtained by applying the inverse Fourier transform to $I_{n1}$, $I_{n2}$, and $I_{n3}$, respectively.

### C. Motion Estimation

As explained in the introduction, our PBME is designed to estimate motion with 3-D TO RF images. This allows us to consider that the 3-D signals extracted locally from images $i_1$ and $i_2$ follow the model:

$$i_1(x, y, z) = \omega_{x1}(x, y, z) \cos(2\pi f_x(x - d_{x1}))$$

$$\times \cos(2\pi f_y(y - d_{y1})) \cos(2\pi f_z(z - d_{z1}))$$

$$i_2(x, y, z) = \omega_{x2}(x, y, z) \cos(2\pi f_x(x - d_{x2}))$$

$$\times \cos(2\pi f_y(y - d_{y2})) \cos(2\pi f_z(z - d_{z2})), (13)$$

where $f_x$, $f_y$, and $f_z$ are the spatial frequencies on each direction assumed to be known; $\omega_{x1}$ and $\omega_{x2}$ are two 3-D Gaussian envelopes as defined in Section II-B; and $d_{x1}$, $d_{x2}$, $d_{y1}$, and $d_{y2}$ are the relative displacement between the image $i_x$ and a point of reference.

Local phases are extracted using the single octave analytical signals in (11) of the volumes $i_1$ and $i_2$ defined in (13) and denoted as $\phi_{11}$, $\phi_{12}$, $\phi_{21}$, $\phi_{22}$, and $\phi_{23}$. Using elementary trigonometry identities, we can show that these local phases can be written as follows:

$$\phi_{n1}(x, y, z) = 2\pi \times (f_x(x - d_{x1}) + f_y(y - d_{y1}) + f_z(z - d_{z1}))$$

$$\phi_{n2}(x, y, z) = 2\pi \times (f_x(x - d_{x2}) - f_y(y - d_{y2}) + f_z(z - d_{z2}))$$

$$\phi_{n3}(x, y, z) = 2\pi \times (f_x(x - d_{x3}) - f_y(y - d_{y3}) - f_z(z - d_{z3}))$$

with $n = \{1, 2\}$.

Let us denote by $\phi_1$, $\phi_2$, and $\phi_3$ the phase differences between the same type of complex signals corresponding to the signal $i_1$ and to the signal $i_2$.

$$\phi_1(x, y, z) = \phi_{11}(x, y, z) - \phi_{21}(x, y, z)$$

$$\phi_2(x, y, z) = \phi_{12}(x, y, z) - \phi_{22}(x, y, z),$$

$$\phi_3(x, y, z) = \phi_{13}(x, y, z) - \phi_{23}(x, y, z)$$

$d_{x}$, $d_{y}$, and $d_{z}$ are the relative 3-D spatial shift between volumes $i_1$ and $i_2$ that have to be estimated, defined as:

$$d_x = d_{x2} - d_{x1}$$

$$d_y = d_{y2} - d_{y1}$$

$$d_z = d_{z2} - d_{z1}.$$ (16)

In this paper, we consider that the two volumes $i_1$ and $i_2$ are delayed (in each direction) by less than half an oscillation period.

Taking into account the equations in (14), (15), and (16) and using elementary calculations, the form of the three phase differences $\phi_1$, $\phi_2$, and $\phi_3$ is obtained:

$$\phi_1(x, y, z) = 2\pi (f_x d_x + f_y d_y + f_z d_z)$$

$$\phi_2(x, y, z) = 2\pi (f_y d_y - f_x d_y + f_z d_z).$$

$$\phi_3(x, y, z) = 2\pi (f_x d_x - f_y d_y - f_z d_z). (17)$$

As explained in [30], the phases in (14) are not linear for this entire interval, but present phase jumps. We observe that $\phi_1$, $\phi_2$, and $\phi_3$ do not depend on the spatial variables $x$, $y$, and $z$. Thus, the three phase differences are assumed to be constant on the entire definition interval of volumes $i_1$ and $i_2$. This makes it possible to easily estimate $x$, $y$, and $z$ values corresponding to phase jumps. Knowing the frequencies $f_x$, $f_y$, and $f_z$ and considering that the spatial displacement between the volumes is smaller than half their period, we can easily threshold $\phi_1$, $\phi_2$, and $\phi_3$ to eliminate phase jumps. Moreover, we note $\bar{\phi}_1$, $\bar{\phi}_2$, and $\bar{\phi}_3$ as the mean spatial values of $\phi_1$, $\phi_2$, and $\phi_3$ after thresholding. The equations in (17) can be written as follows:

$$\begin{bmatrix} \bar{\phi}_1 \\ \bar{\phi}_2 \\ \bar{\phi}_3 \end{bmatrix} = \begin{bmatrix} 2\pi f_x & 2\pi f_y & 2\pi f_z \\ 2\pi f_x & -2\pi f_y & 2\pi f_z \\ 2\pi f_x & -2\pi f_y & -2\pi f_z \end{bmatrix} \begin{bmatrix} d_x \\ d_y \\ d_z \end{bmatrix}. (18)$$
This system of three equations with three unknowns allows us to analytically estimate the relative 2-D spatial shift between $i_1$ and $i_2$:

$$
\hat{d}_x = \frac{\phi_1 - \phi_2}{4\pi f_x} ; \quad \hat{d}_y = \frac{\phi_2 - \phi_3}{4\pi f_y} ; \quad \hat{d}_z = \frac{\phi_3 + \phi_1}{4\pi f_z}.
$$

(19)

Note that the PBME estimator has several advantages, as will be discussed later in the paper, but it cannot estimate a displacement larger than half a wavelength of the oscillations in each direction. This limitation is discussed in Section VII.

IV. Comparison With Other Existing Methods

The accuracy of our 3-D phase block matching was compared with two other motion estimation methods.

A. Block Matching

The first one is a classical block matching algorithm using the maximum of the cross-correlation function as a local estimator and referred to as BM, for block matching with a normalized cross-correlation. Thus, in contrast to the phase block matching proposed herein, this method uses the amplitude images $i_1$ and $i_2$ to estimate motion. Because a sub-pixel precision of the local estimation is required with our application, a Gaussian peak-fitting strategy was employed.

B. Optical Flow

The second is a 3-D extension of the Lucas-Kanade optical flow method [47]. The calculation of the apparent velocity field is based on the estimation of the spatio-temporal changes in pixel intensities throughout an image sequence. The basic idea is to relate time differences to spatial differences.

In contrast to PBME, which is applied on RF images, these two methods are usually applied on the RF envelopes. In this paper, these two conventional methods were additionally tested on RF data with TO. As a result, the PBME was compared with four methods: block matching on B-mode image (BMBM), block matching on TO images (BMTO), optical flow on B-mode (OFBM), and optical flow on TO images (OFTO).

V. Simulation Results

Simulated data were used to validate the implementation of 3-D beamforming on a 2-D phased array and to evaluate the proposed motion estimation method. First, the 3-D TO beamforming strategy was analyzed with a simulated PSF. Afterwards, the PBME was validated on a 3-D moving simulated medium. The simulated RF signals were corrupted with a Gaussian noise to evaluate the robustness of the PBME. The simulations were repeated 10 times for statistical purposes.

A. Simulation Description

The simulations were performed using the ultrasound simulation program Field II [48], [49]. In this section, all results were obtained with the same 2-D phased array; the simulation parameters are reported in Table I. Pyramidal 10 mm × 30° × 30° volumes were acquired using a 0.5° azimuth, elevation angular sampling step, and a 0.10-mm axial sampling step.

The beamforming parameters were adjusted to obtain the same spatial wavelength equal to 0.13 rad under the Gaussian envelopes, in both transverse directions, with $x_{\varphi_0} = x_{\varphi_0} = 4$ mm, and $\sigma_{\varphi_0} = \sigma_{\varphi_0} = 0.9$ mm.

B. 3-D Beamforming Validation

To investigate how well defined the 3-D transverse oscillations are, one 3-D PSF, at a depth of 30 mm, was calculated with Field II with the specified 2-D apodization function (see Fig. 3). For both transverse directions, the theoretical transverse spatial wavelength was 0.13 rad. The mean measured wavelength was 0.136 rad. We obtained a bias of 4.6% between the theoretical and simulated transverse spatial wavelengths.

C. PBME Method Validation

To validate the PBME method, a 3-D trajectory was investigated. A random reflector map was generated with a 3-D motion corresponding to an elliptical trajectory. The axial, lateral, and elevation maximum amplitudes of displacement were 0.4, 1, and 1 mm, respectively. We choose a grid of $20 \times 15 \times 15$ voxels in the volume of interest, which corresponded to a volume of $0.2 \times 3.9 \times 3.9$ mm$^3$ at a depth of 30 mm. The PBME was evaluated on 10 different simulations of the same trajectory. Our motion estimator was then applied to each consecutive pair

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Transducer</td>
<td>2-D phased array</td>
</tr>
<tr>
<td>Number of elements</td>
<td>64 × 64</td>
</tr>
<tr>
<td>Center frequency</td>
<td>3 MHz</td>
</tr>
<tr>
<td>Pitch</td>
<td>$\lambda/2$</td>
</tr>
<tr>
<td>Kerf</td>
<td>$\lambda/100$</td>
</tr>
<tr>
<td>Sampling frequency</td>
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<tr>
<td>Speed of sound</td>
<td>1540 m/s</td>
</tr>
<tr>
<td>No. of transmit cycles</td>
<td>2</td>
</tr>
<tr>
<td>Transmit focus</td>
<td>Plane wave</td>
</tr>
<tr>
<td>Receive focus</td>
<td>Dynamic</td>
</tr>
<tr>
<td>Transmit apodization</td>
<td>2-D Hanning windows</td>
</tr>
<tr>
<td>Receive apodization</td>
<td>4 distant peaks (Fig. 2)</td>
</tr>
<tr>
<td>Number of lines</td>
<td>64 × 64</td>
</tr>
<tr>
<td>Volume width</td>
<td>$30' \times 30'$</td>
</tr>
</tbody>
</table>
of 3-D RF images. The results were compared with those obtained with the conventional methods described in Section IV (see Fig. 4). Then, to study the behavior of the methods under controlled noise, the simulated RF images were corrupted with Gaussian noise. As a result, three additional image sequences with increasing noise levels were used: a sequence with SNR equal to 25 dB, a sequence with SNR equal to 20 dB, and a sequence equal to 15 dB. The corresponding mean and standard deviation of relative error between the estimated and actual displacements are summarized in Table II.

Regarding the axial direction, the five methods gave close results, even if a slight improvement was noticed with the proposed method. There was no clear improvement for the classical methods when applied to TO images. The use of TO images, however, allowed better motion estimates for the two classical methods, in the two transverse directions. Nevertheless, the proposed algorithm outperformed the conventional techniques and returned the best trajectory estimate.

With regard to the classical methods (block matching and optical flow), according to the results presented in Table II, we noticed that the use of TO images allows a better motion estimation (on average the error is reduced by a factor of 1.5). However, errors obtained with classical methods remained higher than those obtained by our method. Indeed in axial, lateral, and elevation directions, and for each sequence (not corrupted and SNR of 25, 20, and 15 dB), the lowest errors were obtained with the proposed PBME. Moreover, one can notice that the errors obtained by the presented PBME method on noise corrupted sequences remain smaller than those obtained with
the other methods on uncorrupted sequences even with the lowest SNR (15 dB).

VI. EXPERIMENTAL RESULTS

In this section, the RF-3-D method was implemented and evaluated on a research scanner and the accuracy of our 3-D motion estimator was tested with experimental images.

A. Acquisition Description

The proposed beamforming strategy, designed to generate, in real time, 3-D-RF images with TO in both transverse directions, was implemented on a research scanner (Ultrasound Advanced Open Platform, ULA-OP) [50], equipped with a 2-D phased array composed of 8 × 8 elements. All results were obtained with the same parameters shown in Table III.

B. 3-D Beamforming Validation

In this subsection, the 3-D specific beamforming method was validated on a PSF. To simulate a PSF in 3-D, we positioned a 0.1-mm beads in a degassed gel to get a PSF at 20 mm depth. The results were compared with a Field II simulation using the same parameters as those implemented in the research scanner. The simulated and acquired PSFs were very similar (Fig. 5), which approved the experimental implementation of our specific TO beamforming.

C. PBME Method Validation

In this section, the accuracy of the motion estimation was tested with experimental images acquired in two different scenarios: a linear translation and a more complex 3-D trajectory. The results were compared with those obtained with the methods introduced in Section IV. The size of the region of interest (ROI) used for the estimation was defined at 24 × 16 × 16 voxels for the two comparison methods and at 24 × 10 × 10 voxels for the PBME method. The influence of the ROI size on the estimation is not negligible and is discussed in Section VII.

A homogeneous homemade gelatin phantom, approximately 10 × 10 × 10 cm³ in size, was constructed and used in the experiments. The phantom was fixed at the bottom of a large water tank and remained stationary throughout the experiments. The RF-3-D experimental images were acquired during a 3-D known displacement achieved by moving the transducer with the aid of a 3-D motor-controlled translation system (OWIS GmbH, Staufen im Breisgau, Germany). For the two next experiments, 30 estimations were made using different ROI locations to provide an indication of the robustness of the methods.

TABLE II. Trajectory Estimation Errors.

<table>
<thead>
<tr>
<th>Method</th>
<th>No. corrupted</th>
<th>SNR (dB)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>25</td>
</tr>
<tr>
<td>Block matching (B-mode)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lateral</td>
<td>10.07 ± 1.69</td>
<td>10.56 ± 6.87</td>
</tr>
<tr>
<td>Elevation</td>
<td>10.28 ± 1.59</td>
<td>10.76 ± 6.34</td>
</tr>
<tr>
<td>Axial</td>
<td>1.64 ± 0.6</td>
<td>2.04 ± 0.8</td>
</tr>
<tr>
<td>Block matching (TO)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lateral</td>
<td>5.95 ± 2.79</td>
<td>6.44 ± 2.85</td>
</tr>
<tr>
<td>Elevation</td>
<td>6.06 ± 2.85</td>
<td>6.54 ± 2.89</td>
</tr>
<tr>
<td>Axial</td>
<td>2.18 ± 0.7</td>
<td>2.58 ± 1.2</td>
</tr>
<tr>
<td>Optical flow (B-mode)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lateral</td>
<td>14.8 ± 2.17</td>
<td>17.61 ± 8.95</td>
</tr>
<tr>
<td>Elevation</td>
<td>15.12 ± 2.19</td>
<td>17.61 ± 8.64</td>
</tr>
<tr>
<td>Axial</td>
<td>3.2 ± 0.78</td>
<td>5.32 ± 2.6</td>
</tr>
<tr>
<td>Optical flow (TO)</td>
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<td></td>
</tr>
<tr>
<td>Lateral</td>
<td>10.03 ± 1.2</td>
<td>12.84 ± 5.51</td>
</tr>
<tr>
<td>Elevation</td>
<td>10.42 ± 1.1</td>
<td>12.91 ± 5.37</td>
</tr>
<tr>
<td>Axial</td>
<td>2.68 ± 0.66</td>
<td>4.8 ± 2.4</td>
</tr>
<tr>
<td>PBME</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lateral</td>
<td>2.93 ± 2.3</td>
<td>3.06 ± 2.54</td>
</tr>
<tr>
<td>Elevation</td>
<td>2.97 ± 2.4</td>
<td>3.17 ± 2.38</td>
</tr>
<tr>
<td>Axial</td>
<td>0.88 ± 0.6</td>
<td>0.92 ± 0.64</td>
</tr>
</tbody>
</table>

Boldface indicates the minimal errors.

TABLE III. Parameter Setting for the Experimental Results.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Transducer</td>
<td>2-D phased array</td>
</tr>
<tr>
<td>Number of elements</td>
<td>8 × 8</td>
</tr>
<tr>
<td>Center frequency</td>
<td>3.9 MHz</td>
</tr>
<tr>
<td>Pitch</td>
<td>0.37 × 0.37 mm</td>
</tr>
<tr>
<td>Kerf</td>
<td>0.03 × 0.03 mm</td>
</tr>
<tr>
<td>Sampling frequency</td>
<td>50 MHz</td>
</tr>
<tr>
<td>Speed of sound</td>
<td>1540 m/s</td>
</tr>
<tr>
<td>No. of transmit cycles</td>
<td>3</td>
</tr>
<tr>
<td>Transmit focus</td>
<td>Plane wave</td>
</tr>
<tr>
<td>Receive focus</td>
<td>Dynamic focusing</td>
</tr>
<tr>
<td>Transmit apodization</td>
<td>2-D Hanning windows</td>
</tr>
<tr>
<td>Receive apodization</td>
<td>4 distant peaks</td>
</tr>
<tr>
<td>Number of lines</td>
<td>32 × 32</td>
</tr>
<tr>
<td>Volume width</td>
<td>64° × 64°</td>
</tr>
</tbody>
</table>

TABLE III. Parameter Setting for the Experimental Results.
1) First Experiment—Linear Translation: This first result compares the accuracy of the five methods under the same values of displacement between two consecutive images along a linear trajectory. The transducer was moved 15 times in the three directions simultaneously, from 0 to 0.42 mm in 0.03-mm steps (i.e., 0.03, 0.06, . . . , 0.39, 0.42 mm) for each of the two transverse directions (x, y), and with an increment of 0.015 mm over a distance of 0.21 mm (i.e., 0.015, 0.03, . . . , 0.21 mm) for the axial direction (z). The mean and standard deviation of relative error between the 30 different trajectory estimations and the real displacement are reported in Table IV.

As for errors in the axial direction, as shown in Table IV, the combination of PBME and TO images performed slightly better than block matching and optical flow algorithms, whether on B-mode or TO images. Additionally, an improvement was observed in both transverse directions. Indeed, as even if the classical methods got better results on TO images (as suggested previously in [30]), we obtained a better accuracy in the lateral and elevation directions with the PBME method.

2) Second Experiment—3-D Trajectory: The transducer was moved according to a more complex 3-D trajectory, and 40 volumes were acquired during the motion. The transducer followed an elliptical trajectory with 2.5, 1.5, and 1.5 mm amplitude in the axial, lateral, and elevation directions, respectively. Motion between each consecutive pair of volumes was estimated and accumulated to recover the full 3-D trajectory of the transducer (Fig. 6). The axial, lateral, and elevation displacement errors (mean and standard deviation) measured in millimeters, along the entire trajectory, are reported in Fig. 7. For purpose of clarity of the figures, error bars represent ±1/10 of the standard deviation.

This finding showed the interest of TO images for motion estimation in transverse directions. Indeed, as shown by Figs. 6 and 7, transverse trajectories estimated with the optical flow and BM methods were more accurate with TO than B-mode images, whereas all classical methods returned roughly same estimation in the axial direction. However, consistent with the simulations, the proposed PBME algorithm returned the best displacement estimates in the transverse directions.

![Fig. 5. Experimental 3-D PSF with TO (right) compared with a 3-D PSF simulated by Field II (left) with identical parameter setting. The axial/lateral view is on the top, and the elevation/lateral view is on the bottom. The corresponding lateral and elevation profiles are also reported. The experimental and simulated 3-D PSF show very similar results.](image)

<table>
<thead>
<tr>
<th>Method</th>
<th>Lateral</th>
<th>Elevation</th>
<th>Axial</th>
</tr>
</thead>
<tbody>
<tr>
<td>Block matching (B-mode)</td>
<td>14.21 ± 7.63</td>
<td>15.31 ± 8.26</td>
<td>4.54 ± 3.61</td>
</tr>
<tr>
<td>Block matching (TO)</td>
<td>12.22 ± 8.67</td>
<td>11.51 ± 8.7</td>
<td>4.63 ± 3.7</td>
</tr>
<tr>
<td>Optical flow (B-mode)</td>
<td>10.33 ± 3.29</td>
<td>12.39 ± 3.96</td>
<td>4.5 ± 1.87</td>
</tr>
<tr>
<td>Optical flow (TO)</td>
<td>9.06 ± 3.61</td>
<td>9.74 ± 4.56</td>
<td>4.43 ± 2.07</td>
</tr>
<tr>
<td>PBME</td>
<td>7.99 ± 3.85</td>
<td>8.59 ± 3.66</td>
<td>3.72 ± 1.97</td>
</tr>
</tbody>
</table>

Boldface indicates the minimal errors.
One can also notice the correlation between error and displacement amplitude for the classical methods. For both, optical flow and BM, Fig. 7 shows that the higher the displacement, the greater were the errors. The bias reported by our method, however, seemed to be less correlated with the amplitude of the displacement, making our algorithm more efficient.

VII. DISCUSSION

A full 3-D TO beamforming method and its corresponding phase-based motion estimation method have been proposed and validated in simulation and phantom experiments.

The results confirm that the proposed method is a serious candidate for determining full 3-D motion from volumetric ultrasound images. The results from this research show that 3-D PBME can accurately determine displacements from 3-D RF images that contain oscillations in the transverse directions.

Although the estimation errors of block matching and optical flow on B-mode images were comparable with those found in the literature [18]–[20], our study shows that they can be improved using 3-D TO RF images. We have also shown that despite this improvement our method is still able to estimate the transverse motions with more accuracy.

However, three important parameters have not been discussed and should be in this section: the maximum displacement that can be estimated, the influence of the ROI size (we will see that both depend of the oscillation frequency contained in the image), and the computation load.

A. Computation Load

The comparison of the computation time between the three methods is illustrated in Table V. Three different ROI sizes were investigated using an Intel(R) core i7–3720QM CPU at 2.6 GHz. In terms of computation load, the optical flow was the most expensive method, espe-
cially with a large ROI. In contrast, the proposed PBME method was the fastest. Note that no interpolation is used in the block matching approach, the sub-pixel estimation is made by Gaussian fitting and the computation load time would be considerably increased if the interpolation method was used to access sub-pixel estimation.

B. Influence of ROI Size

The influence of the ROI size was tested only in the lateral direction, changing only the lateral size of the ROI, but the results presented in Fig. 8 can be extrapolated in the other two directions. The PBME method was applied on three single lateral motions (0.1, 0.3, and 0.5 mm of tissue phantom) using the same parameters as presented in Section VI. The lateral ROI size was changed within a range of [8], [28] voxels with an increment of two voxels. Fig. 8 shows that, regardless of the displacement, the motion was correctly estimated when the lateral size of the ROI was around 10 and 20 voxels, which corresponded to one and two lateral wavelengths. This can be attributed to the fact that PBME uses the Matlab (The MathWorks Inc., Natick, MA, USA) fast Fourier transform (FFT) algorithm. If no interpolation is used, i.e., zero padding the FFT correctly works when the signal contains an entire number of oscillations, which is the case when the lateral ROI size is 10 and 20. To summarize, the PBME method works well when the size of the ROI corresponds to an entire number of wavelengths produced by the 3-D TO method.

C. Maximum Estimable Displacement

The theory claims that the maximum displacement that can be estimated in one direction corresponds to half the wavelength of the oscillations present in this direction. To better see this limitation, the PBME method was tested on several single transverse motions of a PSF simulated with Field II at a 15 mm depth, using the same parameters as those presented in Section V. The PSF was translated with a pure lateral displacement, within the range [0; 1.5] mm with a 0.1-mm increment. A grid of 20 × 15 × 15 voxels in the volume of interest was used, which corresponded to a volume of 0.2 × 1.95 × 1.95 mm³ (z, x, y) at 15 mm. The resulting displacement estimations are presented in Fig. 9.

Motion was correctly measured by the PBME method with a mean relative error of 0.5% up to a displacement around 1.0 mm (Fig. 9), which corresponded to the mid-wavelength, at a depth of 15 mm. In each direction, the maximum estimable displacements depend on the central frequency of the probe; consequently, our method is able to estimate a very small motion. For example, in the axial direction with a central frequency of 7 MHz and speed.
of sound of 1540 m/s, we can estimate a motion smaller than 110 µm. This method can be very useful in very high frame-rate imaging, where the movement that must be estimated is very small.

In the transverse directions, this limitation can be examined if we succeed in changing the wavelength of the transverse oscillations versus time. This can be done by filtering the RF images instead of using a specific apodization function. This TO filtering method has already been done in 2-D [51], and in 2-D high-frame-rate imaging [52], by our group, and will be extended in 3-D in a future study.

VIII. Conclusion

We have proposed a full 3-D TO beamforming method with a specific 3-D phase-based motion estimation algorithm. The method is able to estimate a 3-D motion over successive volumetric acquisitions. The 3-D beamforming and PBME method have been tested and validated with Field II simulations and in vitro on a 3-D trajectory of a gelatin phantom. The results have been compared with those obtained with two other conventional estimation methods, block matching and optical flow, and we have seen that the proposed method outperforms them both, especially in the transverse direction. Moreover, this paper has shown that our method is able to achieve accurate transverse motions, notwithstanding the small size of the probe used in the study. Further studies are required with larger probes and for the in vivo evaluation of this method.

Even if 3-D ultrasound is still not the standard in clinics, it is accepted that 3-D imaging has the potential to overcome many limitations of conventional 2-D imaging, such as user dependence or out-of-plane motion of the organ studied as well as the need for the clinician to mentally reconstruct a 3-D scene observed in 2-D. Because the proposed 3-D PBME method cannot estimate a displacement larger than half a wavelength in each direction, this technique can be used in all applications requiring 3-D motion estimation provided it is used with ultrafast imaging (plane or circular wave). Note also that although some applications such as cardiac imaging necessitate a large investigation depth using a limited footprint probe, using the pyramidal scanning enables maintaining constant TO in the polar coordinate system without further separating the peaks of the apodization function. Motion correction in high-intensity focused ultrasound therapy could benefit from this new technique, as could cardiac ventricular function evaluation, which relies on the 3-D motion estimation of the cardiac muscle. This method is even more useful when trying to estimate the tissue motion of a very homogeneous tissue along perpendicular directions of the beam axis such as the 3-D motion of the carotid artery wall and more particularly its longitudinal kinematics, which have been recently identified as an early potential marker of atherosclerosis.

References

Sebastien Salles was born in Bagnolet, France, in 1987. He is a Ph.D student at the CREATIS laboratory of the Lyon National Institute of Applied Sciences. He obtained his electrical engineering degree, coupled with a master’s degree in acoustics from the Lyon National Institute of Applied Sciences in 2012. His thesis focuses on the motion estimation of the carotid artery wall. His main research interests are image formation techniques, image processing, and tissue motion estimation.

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image formation techniques and motion estimation. Since 2014, he has been an associate member of the IEEE Bio Imaging and Signal Processing (BISP) technical committee and a member of the IEEE International Ultrasonics Symposium technical program committee.

Damien Garcia was born in Paris, France. He obtained the DEA and engineer degree from the École Centrale de Marseille in 1997, and the M.Sc. and Ph.D. degrees in biomedical engineering from the University of Montreal in 2003. He was a post-doctoral fellow from 2006 to 2008 in the Department of Echocardiography, Gregorio Marañón Hospital, Madrid, Spain. Dr. Garcia is director of the Research Unit of Biomechanics & Imaging in Cardiology (RUBIC) at the University of Montreal Hospital Research Centre (CRCHUM), and assistant professor in the Department of Radiology, Radio-Oncology and Nuclear Medicine at the University of Montreal. His research interests are in cardiovascular ultrasound imaging, Doppler echocardiography, image processing, mathematical and biomechanical modeling, fluid dynamics, and flow imaging. Damien Garcia holds a research scholarship from the Fonds de Recherche en Santé du Québec (FRSQ). A detailed list of his publications is available from www.biomecardio.com.

Didier Vray is currently a Professor of Signal Processing and Computer Sciences at INSA-Lyon, France. Since he joined the research laboratory CREATIS, his main research interest focus has been on ultrasound medical imaging. His research includes vascular imaging, flow imaging, tissue motion estimation, and bimodality ultrasound/optical imaging. He is the author of more than 100 scientific publications and 2 issued patents in this field.