2-D Arterial Wall Motion Imaging Using Ultrafast Ultrasound and Transverse Oscillations

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Abstract—Ultrafast ultrasound is a promising imaging modality that enabled, inter alia, the development of pulse wave imaging and the local velocity estimation of the so-called pulse wave for a quantitative evaluation of arterial stiffness. However, this technique only focuses on the propagation of the axial displacement of the artery wall, and most techniques are not specific to the intima–media complex and do not take into account the longitudinal motion of this complex. Within this perspective, this paper presents a study of two-dimensional tissue motion estimation in ultrafast imaging combining transverse oscillations, which can improve motion estimation in the transverse direction, i.e., perpendicular to the beam axis, and a phase-averaged motion estimation. First, the method was validated in simulation. Two-dimensional motion, inspired from a real data set acquired from a human carotid artery, was applied to a numerical phantom to produce a simulation data set. The estimated motion showed axial and lateral mean errors of 4.2 ± 3.4 µm and 9.9 ± 7.9 µm, respectively. Afterward, experimental results were obtained on three artery phantoms with different wall stiffnesses. In this study, the vessel phantoms did not contain a pure longitudinal displacement. The longitudinal displacements were induced by the axial force produced by the wall’s axial dilatation. This paper shows that the approach presented is able to perform 2-D tissue motion estimation very accurately even if the displacement values are very small and even in the lateral direction, making it possible to estimate the pulse wave velocity in both the axial and longitudinal directions. This demonstrates the method’s potential to estimate the velocity of purely longitudinal waves propagating in the longitudinal direction. Finally, the stiffnesses of the three vessel phantom walls investigated were estimated with an average relative error of 2.2%.

I. INTRODUCTION

ULTRAFAST ultrasound is a promising imaging modality with several potential clinical applications [1]. For blood flow imaging, this technique has been used to develop ultrafast blood vector velocity imaging [2], which has been applied to different cardiovascular pathologies [3], [4]. In the domain of tissue elasticity imaging, high-frame-rate ultrasound enabled the implementation of real-time shear wave elastography [5], [6]. It has also made it possible to visualize and quantify the natural electromechanical wave in the heart [7]. More generally, ultrafast ultrasound can be used to examine extremely fast phenomena that occur in the human body that have never been observed or analyzed before with any medical imaging modality.

Over the last decade, researchers have shown interest in studying artery wall motion, especially the motion of the carotid intima–media complex (IMC), to demonstrate its significance as a marker of certain cardiovascular pathologies [8], [9] which are among the leading causes of death. Axial motion, i.e., the cross-sectional diameter change, has been widely studied in combination with pulse pressure to assess arterial distensibility and compliance [10], [11]. Several groups [12]–[14] demonstrated that the 2-D motion of the IMC showed a cyclic and reproducible longitudinal motion. According to clinical trials [15], [16], longitudinal motion seems to reflect cardiovascular status and it has been demonstrated that its amplitude can be related to cardiovascular risk.

At the same time, ultrafast imaging has been used to image carotid and aorta artery wall motion [17]–[25]. With the development of pulse wave imaging (PWI), the velocity of the so-called pulse wave (PWV) could be estimated, providing a quantitative evaluation of arterial stiffness. Because the PWV is directly related to arterial stiffness, it has been shown to be an efficient indicator of cardiovascular risk [26], [27].

However, these studies focused only on the propagation of the axial displacement of the artery wall, and most of them were not specific to the IMC. Consequently, they did not take into account the longitudinal motion of the IMC. Neglecting this component of motion can lead to errors in
axial motion estimation [17]. Moreover, it is probable that phenomena that are unobservable with conventional imaging techniques are happening in the longitudinal direction of this complex and might be imaged and quantified with specific imaging techniques.

With this double objective of improving pulse wave imaging and observing the longitudinal behavior of the IMC more precisely, we have developed an ultrafast imaging method for the quantitative evaluation and imaging of the two-dimensional motion of the carotid artery IMC. This development represents a significant challenge because of the difficulty of estimating small lateral displacements. In conventional ultrasound imaging, several motion estimation techniques have been proposed to track the axial and longitudinal motion of the IMC, based on speckle tracking [12], echo-tracking [8], Kalman filtering [28], and finite impulse response filtering [29]. All of these techniques were confronted with the inherent limitation of conventional US imaging, namely the rather coarse resolution cell of the ultrasound images along the lateral direction (i.e., perpendicular to the beam axis) [30]. Moreover, when imaging blood vessels, the longitudinal profiles of the tissues are rather homogeneous because the anatomical layers are aligned along the lateral direction and do not present any significant landmarks.

For these reasons, the longitudinal component of the wall motion is particularly challenging to extract in conventional US imaging. Therefore, a different approach is required to meet the challenge of accurate longitudinal displacement assessment of the IMC.

We previously developed vector motion estimation methods based on transverse oscillation (TO) techniques. The TO method, initially developed for vector flow imaging at the same time by two different groups (Jensen and Munk [31] and Anderson [32]), has been adapted for tissue vector motion estimation by our group [33], [34] and others [35], [36]. The idea behind TOs is to produce a pressure field featuring oscillations not only in the axial direction, i.e., along the ultrasound beam axis, but also in the perpendicular direction. The objective is to generate ultrasound images suitable for longitudinal motion estimation. Given their nature, TO schemes are usually combined with phase-based motion estimation (PBME) techniques [33], [34]. This method and its feasibility have already been validated in several papers by our group [37]–[40]. This technique, however, is limited to displacements shorter than the mid-wavelength. Fortunately, in ultrafast imaging the motion amplitude between successive frames is substantially reduced, making it compatible with the mid-wavelength constraint of the phase-based estimation techniques.

We therefore propose to combine ultrafast ultrasound imaging, TOs, and a PBME algorithm. The objective of this paper is to develop a high-frame-rate, precise, and robust 2-D motion estimation algorithm to estimate the local PWV and arterial stiffness, taking into account the longitudinal component of the motion. The proposed technique is validated through a controllable and repeatable motion with simulations and in vitro experiments conducted on artery phantoms. Moreover, as discussed in the following paper, because the wall longitudinal displacements of the artery phantom were induced by the axial force produced by wall axial dilatation (induced by pulsatile flow), the estimation of the axial and longitudinal motion propagation velocity must match closely.

An initial version of this study was presented at the 2014 IEEE International Ultrasonics Symposium and is extended here [41]. In particular, the study is extended to phantoms with different stiffnesses, and a quantitative evaluation of the phantom stiffness is provided.

In the following, Section II briefly reviews the principles of ultrafast imaging, TO imaging, and how it has been modified to combine it with plane wave imaging, the principles of our PBME algorithm, and finally the mechanical model used to estimate wall stiffness. The principle of the estimation algorithm is then given. Sections III and IV present the validation of this method on simulated RF images and the experimental results of our technique on in vitro experiments, respectively.

II. Plane Wave TO Imaging

In this section, the plane wave transverse oscillation imaging method is presented. First, the plane wave imaging principle is briefly reviewed, then the conventional transverse oscillation method using the double Gaussian apodization function is explained, and finally the filtering methods used in this manuscript to obtain the transverse oscillations are described. Note that the use of a specific apodization function during the beamforming process amounts to spatially (in the transverse direction) filtering the information [42]. As a result, the conventional transverse oscillations method and the filtering method presented are approximately the same technique; however, the first is carried out during the beamforming and the second after the beamforming process.

A. Plane Wave Imaging

One common way to achieve ultrafast imaging is to produce broad and plane waves by exciting all the probe elements simultaneously. After a plane wave has been transmitted, the raw RF signals received by the probe elements must be coherently coalesced to construct an ultrasound image. Several methods can be used to construct (beamform) the image as a low-resolution image. One of the simplest methods is given by the delay-and-sum approach used by Montaldo et al. in [43], performed in the spatial domain. There is another family of methods for recovering images from the raw RF signals after plane-wave transmission, which work in the Fourier domain. The main step of this process is remapping the Fourier coefficients. This step can differ depending on the authors and the method used to derive the remapping equations [44]–[47].
Combining low-resolution images into high-resolution images at a frame-rate equal to the pulse-repetition frequency (PRF), is a technique borrowed from radar called synthetic aperture. Synthetic aperture imaging, can be carried out in different fashions; for instance, by using plane- or spherical waves. Tanter and Fink’s group introduced the term coherent compounding, which employs multiple plane waves emitted from different angles before summed to form a high-resolution image [43]. Jensen’s group used spherical waves for obtaining high-frame-rate vector velocity imaging [48]. The two techniques both use synthetic aperture imaging, where low-resolution images from different insouification angles are summed. A third way of obtaining ultrafast imaging is using a single-emission scheme in which no summing of low-resolution images is carried out—that is what is used in this paper.

B. Conventional TOs

The aim of TO imaging is to generate a sound field featuring oscillations in the transverse direction, i.e., perpendicular to the ultrasound beam axis. A beamformer specifically developed for TO is used to obtain this pressure field \(h(x)\). First, the conditions of Fraunhofer approximation are reached by focusing the received beam. Then the appropriate apodization (weighting) of the raw signals received by the probe elements, \(w_i(x)\), must be chosen. As stated by the Fraunhofer approximation, \(w_i(x)\) and \(h(x)\) can be approximated as the Fourier transforms of each other and the TO apodization function is typically composed of two peaks (see also Fig. 1):

\[
w_i(x) = \frac{1}{2} \left( e^{-\pi(x-x_0)/\sigma_0} + e^{-\pi(x-x_0)/\sigma_0} \right)
\]

\[
h(x) = e^{-\pi(x/\sigma_x)^2} \cos\left( \frac{2\pi x}{\lambda_x} \right),
\]

where \(x_0 = \lambda z/\lambda_x\) and \(\sigma_0 = \sqrt{2}\lambda z/\sigma_x\). \(x_i\) represents the position of the transducer element, \(z\) is the depth of interest, \(\lambda\) is the wavelength of the transmitted pulse, \(\lambda_x\) is the expected lateral wavelength, and \(\sigma_x\) is the full-width at half-maximum of the Gaussian envelope. When this beamforming approach is used, one obtains 2-D RF images with both axial and transverse oscillations. Note that the combination of conventional TO and ultrafast imaging has been proposed in [49] for flow estimation.

C. TOs by Filtering

As can be seen in (1), one limitation of the conventional TO approach mentioned previously is that the TO wavelength depends on the depth. As a consequence, if the same TO wavelength is expected in the entire image, which is usually the case, the apodization function must be dynamically adapted as a function of depth. Moreover, TOs are created during beamforming, and the TO parameters cannot be changed without beamforming the whole set of raw data once more. Changing the TO wavelength could, however, be very useful, for example in a multi-resolution approach [50]. To overcome these two limitations and ensure optimal control of the TO parameters, the approach proposed in our method is to produce the TO by filtering the beamformed RF image.

It can simply be demonstrated and observed that the 2-D Fourier transform of a TO image must clearly exhibit four identified spots whose positions and width correspond to the axial and TO wavelengths and bandwidths. We thus develop a filtering-based TO imaging mode that has been shown to be able to produce images exhibiting four well-defined spots in the 2-D spectrum. Several techniques can be used to obtain the expected oscillations [51]. The convolution method used here seems the best adapted and most versatile approach. It is conducted only along the lateral direction. The axial oscillations are already present in the columns of the RF image, and we wish to affect the axial oscillations as little as possible. To produce the TOs, each line of the RF images is convolved with the adapted filter. In this study, the spatial lateral filter given in (3) is the simple multiplication of a sinusoid and a Gaussian window:

\[
\omega(x) = G(x) \times \cos\left( \frac{2\pi x}{\lambda_{0x}} \right),
\]

where \(x\) represents the position of the transducer element and \(\lambda_{0x}\) represents the desired TO wavelength, and we have

\[
G(x) = \frac{1}{\sigma_x \sqrt{2\pi}} e^{-x^2/(2\sigma_x^2)},
\]

with
where $\sigma_x$ is chosen to be of the same order as the width of the axial frequency spectrum. Much as the 2-D filter can be applied on a small region of the RF images, in this study it was applied to the entire 2-D RF signal to create lateral modulation. This technique is able to produce TOs with any lateral wavelength value, but in an attempt not to keep only the noise of the image, the value chosen must be coherent with the frequency content of the initial RF images.

The filtering method to produce the TO images from conventional RF images is available as a Matlab (The MathWorks Inc., Natick, MA, USA) graphical user interface at http://www.creatis.insa-lyon.fr/ius-special-issue-2014/.

**D. Phase-Based Motion Estimation**

Once TO images have been obtained, it becomes possible to estimate the vector motion using the phase-based technique described in [34]. The principle of this local method is to create complex image blocks based on a 2-D extension of the analytical signal using the Hann approach [37]. Basically, an image block of this type is generated by keeping only one quadrant of its 2-D Fourier spectrum. Two different analytical images are obtained by keeping two different quadrants of the 2-D Fourier spectra. The 2-D translation vector $(D_x, D_z)$ between two successive images from a sequence is then deduced from the phases $\phi_{11}$, $\phi_{12}$, $\phi_{21}$, and $\phi_{22}$ according to

\[
D_x = \frac{(\Phi_1 - \Phi_2) \lambda_{0z}}{2\pi},
\]

\[
D_z = \frac{(\Phi_1 + \Phi_2) \lambda_{0z}}{2\pi},
\]

with

\[
\Phi_1 = \phi_{11} - \phi_{12},
\]

\[
\Phi_2 = \phi_{21} - \phi_{22},
\]

where $\lambda_{0z}$ is the axial wavelength of the TO RF images corresponding to the wavelength of the signal transmitted by the probe, $f_{0z}$ is the lateral wavelength of the TO RF images produced by filtering, $\phi_{11}$ and $\phi_{12}$ are the phases of two analytical images from the first image, and $\phi_{21}$ and $\phi_{22}$ are the phases of two analytical images from the next consecutive image. Because the axial and lateral displacements are related to the axial and lateral wavelengths, we understand the advantage of fully controlling the TO wavelength. For further detail on this motion estimation technique, the reader can refer to [38].

To work perfectly, the axial and lateral region of interest (ROI) sizes must correspond to entire numbers of wavelengths; this parameter is discussed in Section V.

This estimation method is available in a Matlab function at http://www.creatis.insa-lyon.fr/ius-special-issue-2014/.

**E. Wall Stiffness Estimation**

The pulse wave results from the wall axial displacement propagation generated by the pulsatile flow. Their propagation speed and pattern are related to the underlying mechanical properties (e.g., arterial stiffness). In a straight elastic tube containing a nonviscous liquid, the disturbances in pressure, tube diameter, or axial wall displacement (and axial wall velocity, propagate as waves along the tube, at a certain velocity [52]. The Young’s modulus of the tube is related to the axial wave velocity through the modified Moens-Korteweg equation as follows [53]:

\[
c = \sqrt{\frac{E\mu}{2R(1-\nu^2)}}
\]

where $c$ is the axial wave velocity, $E$ is the Young’s modulus of the conduit wall, $h$ is the wall thickness, $\rho$ is the density of the wall, $R$ the inner radius of the tube, and $\nu$ is the Poisson ratio ($\nu \approx 0.5$ for soft quasi-incompressible tissues).

We assume that the axial displacement induced two longitudinal motions that are opposed to each other, as illustrated in Fig. 2.

Fig. 2 shows a schematic diagram of the propagation of the pulse wave along the wall. At time $t_1$, the pulse wave (represented by the red arrow) reaches the longitudinal position $d_1$. At time $t_1 + \Delta t$, the pulse wave reaches the longitudinal position $d_1 + \Delta d$. The velocity of the pulse wave was calculated as the incremental displacement $\Delta d$ divided by the time interval $\Delta t$. Because we assume that the wall’s axial displacement produced the wall’s longitudinal displacements, the PWV can be estimated following the axial and longitudinal displacement propagation.

**III. SIMULATION RESULTS**

First, to evaluate its feasibility, the method presented was applied to simulated RF images. In this section, se-
quences of RF images were produced with Field II [54], [55], validating the TO technique presented as well as the entire method on fully controlled trajectories of simulated data.

A. Simulation Setup

The medium investigated was a homogeneous numerical phantom composed of 5000 scatterers distributed uniformly over a 1 × 20 mm rectangle whose top was positioned 15 mm deep. The scattering amplitudes followed a unit normal distribution. The scatterers composing the medium were moved according to a trajectory obtained from an ultrasound sequence acquired on a healthy subject. This trajectory was extracted by block matching and was low-pass filtered to remove noise. It is composed of 240 consecutive axial and longitudinal displacements. The trajectory was composed of axial and lateral displacements ranging from 0.04 to 36 µm and 0.02 to 32 µm, respectively, and trajectory amplitude approximately 1 mm.

The parameters of the simulated probe corresponded to those of the L523 linear probe from Esaote S.p.A. (Genoa, Italy); see Table I. The raw RF signals were beamformed using the Stolt’s method for plane wave imaging [47].

B. TOs

As represented in Fig. 3, the frequency spectrum of the RF image from the simulated medium by Field II is very broadband in the lateral direction and many different wavelengths can be chosen. In this study, two different lateral wavelengths were chosen arbitrarily and investigated, 1.5 mm and 1 mm; for each case σx was equal to 0.2 mm.

Fig. 3 shows that using our TO filtering method, one can very easily obtain TO images with different wavelengths. In these examples, RF images are produced with lateral wavelengths of 1.5 mm and 1 mm.

C. Tracking Results

Tracking was done by selecting 30 regions of interest (ROIs) on four data sets simulated with Field II as described in Section III-A. Our method was tested with TO wavelengths equal to 1.5 mm and 1 mm. The axial and lateral ROI sizes were equal to two axial and lateral wavelengths, respectively. The mean errors between each estimated trajectory and the real trajectory are calculated and reported in Table II.

The results illustrated in Fig. 4 show that the technique was accurate in tracking the 2-D motion of a simulated homogeneous medium. However, a small difference can be noted between the 1.5-mm and the 1-mm examples. As explained in Section V, the axial and lateral ROI sizes should correspond to entire numbers of wavelengths. As a result, because the motion estimation results from the mean of the ROI estimate motion, the bias is worse for the 1-mm case because the ROI is smaller than in the 1.5-mm case.

IV. EXPERIMENTAL RESULTS

The simulation study given in the previous section has shown that the method presented was able to accurately estimate a 2-D trajectory composed of small displacements. To further evaluate its feasibility, we suggest applying this technique to experimental phantom data.

A. Experimental Setup

The experiments were conducted on artery phantoms made in polyvinyl alcohol (PVA) cryogel. The setup is depicted in Fig. 5.

1) Artery Phantom Design and Preparation: The phantom was fabricated using the lost-core molding technique [56]. The outer and inner diameters of the phantom were 9 and 7 mm, respectively, and its length was 260 mm. An in-house formula was used to make this phantom, with 83.7% distilled water, 15.0% PVA, 1.0% 0.5- to 10-µm silicon dioxide scatterers, and 0.3% potassium sorbate preservatives. Because the phantom material was in the form of PVA cryogel, the wall elasticity could be

![Fig. 2. Illustration of the wave propagation along the wall. It is assumed that the axial displacement (red arrow) induced two opposed longitudinal displacements (blue arrows). Small black squares connected by a dotted line represent the same part of the wall before and after the passage of the wave.](image)
modified by controlling the number of freeze–thaw cycles [57]. Each freeze–thaw cycle consisted of a 24-h freeze period at $-20^\circ C \pm 0.5^\circ C$ and a 24-h thaw period at $4^\circ C \pm 0.5^\circ C$. Three phantoms with different numbers of freeze–thaw cycles were prepared: stiff, medium, and soft phantoms with three, two, and one freeze–thaw cycles, respectively.

2) Young’s Modulus Measurement: The Young’s moduli for the PVA phantoms were measured in the circumfer-

Fig. 3. TO images produced by filtering. The frequency spectrum of a classic beamformed RF image is multiplied by a mask described in Section II-C. Two lateral wavelengths, 1.5 mm and 1 mm, were investigated.
ential direction. To derive this quantity, we have gauged the change in pressure (to obtain the stress) and the corresponding change in wall thickness (to obtain the strain). A pressure change was created by stepwise injecting water (in 5-mL increments using a syringe pump) through the inlet into the phantom’s inner lumen which was pre-filled with water, thereby creating a lumen expansion. Such pressure change was detected by placing a pressure sensor at the phantom outlet (which was sealed beforehand to avoid water leakage). To determine the corresponding change in wall thickness, the known expanded volume was first used to derive the change in lumen diameter using the standard formula for cylindrical volume ($V = \pi d^2/4h$) and the ratiometric relation $V' / V_0 = (d'/d_0)^2$. The change in the outer wall radius was then estimated based on an assumption that PVA volume was preserved during the inflation process. Lastly, the strain was derived based on the change in wall thickness, and it was used to compute the Young’s modulus accordingly.

3) Pulsatile Flow System: Concerning the pulsatile flow system, an in-house-developed arbitrary flow system was used for this work, with a gear pump controlled by programmable data acquisition (DAQ) and the Labview software (National Instruments Corp., Austin, TX). The square flow was produced with a peak flow rate of 8 mL/s and a duty cycle of 10%.

4) Plane-Wave Data Acquisition: For data acquisition, the Ultrasonix SonixTouch US system (Richmond, BC, Canada) with the data acquisition device Sonix-DAQ (Richmond, BC, Canada) was used, the ultrasound probe was positioned parallel to the vessel direction. As in the simulation, no compounding was performed and only a single plane wave insufflation at 90° was performed. With this setup, the frame rate reached 10 kHz. The linear L14-5W/60 array used and the acquisition parameters are described in Table III. Each sequence was composed of 32000 RF images (~3.2 s).

5) Tracking Parameters: The beamformed RF images were filtered with the TO filtering described in Section II-C, the lateral wavelength was set at 1.5 mm. The tracking was done using the motion estimation described in Section II-D, for each pixel along the wall phantom located at 30 mm of depth. A so-called spatiotemporal profile was obtained by drawing the trajectory of each pixel as the lines of a single image. The axial and lateral ROI size was equal to two axial and lateral wavelengths, respectively.

B. Tracking Results

The estimated trajectories were reproducible all along the vessel wall, as represented in Fig. 6(a), which corre-
responds to the results obtained with the soft phantom. The line color corresponds to the position of the ROI on the vessel wall at $t = 0$ s. A light line color corresponds to an estimation made on the left side of the vessel and a dark line color corresponds to an estimation made on the right side of the vessel wall. A zoom is shown at the beginning of the cycle, where the propagation of the pulse wave was visible in both axial and lateral directions. (b) 2-D vessel wall velocity vectors of the soft phantom at six different times during the beginning of the cycle, with a time interval of 3 ms. Only the posterior wall motion is shown. The velocity vectors are color coded according to the velocity amplitude and overlaid onto the B-mode images.

The axial and lateral amplitudes of the estimated trajectories were 0.5 and 1 mm, respectively, with an interframe displacement estimated between 0.1 and 10 $\mu$m for the axial trajectory and between 0.1 and 6 $\mu$m for the lateral trajectory. This shows that a method able to estimate very small displacements is required. As expected, the amplitude of axial and lateral trajectories increased at the mechanical pulse and decreased until the end of the sequence. Some oscillations are visible during the decreasing phase, which can be explained by the fact that the phantom was fixed on both sides, leading to reverberations of the motion profiles.

Note that no pure longitudinal displacements (bulk wave) were present in these experiments; indeed, only axial displacements were introduced into the phantom by the pulsatile flow. The longitudinal displacements present and measured during these experiments were due to the axial dilatation of the wall.

Fig. 6(a) illustrates a zoomed-in view at the beginning of the cycle when the vessel wall begins to undergo downward motion. The magnification was performed in the axial and lateral trajectories. Again, to see the displacement propagation better, the line color was set according to the position of the estimated trajectory. The propagation of the axial and longitudinal motion was visible through the zooms. Concerning the axial trajectory, the movement took about 16 ms to spread from the beginning to the end of the phantom wall, the beginning of the wall (left) started to increase at 249 ms and the end of the wall (right) started to increase at 265 ms. These propagation values can also be observed in the lateral direction. This 2-D motion propagation is visible in Fig. 6(b), which shows the 2-D wall velocity vectors of the soft phantom at six different time instants at the beginning of the cycle, with a 3-s time interval between two plots. Only the posterior wall motion is represented. The velocity vectors were color-coded according to the velocity amplitude and overlaid onto the B-mode images. The wall velocities were obtained by multiplying the estimated incremental displacements (i.e., between two consecutive images) by the frame rate. Regarding the orientation of the vectors more precisely, we observed that the axial motion exerted a force on the wall and introduced opposed longitudinal motions on the both sides.

The movie from which the six images were extracted is available at http://www.creatis.insa-lyon.fr/ius-special-issue-2014/.

Two-dimensional spatiotemporal images were used to calculate PWV and the longitudinal propagating wave velocity (LWV) corresponding to the axial and longitudinal displacements, respectively. Fig. 7 illustrates the spatiotemporal variation of the axial and lateral wall velocities in the soft phantom. The horizontal axis represents time; the vertical axis represents the longitudinal positions of the phantom artery (i.e., distance from left to right). Note that the lateral velocity map clearly shows the propagation of two opposite longitudinal motions (i.e., red and blue) that are introduced by the axial motion.

The black line indicates the foot of the waves at different lateral positions. The foot (i.e., fiduciary point) of the wall velocity waveform was defined as the inflection point at which the temporal derivative of the velocity (i.e., wall acceleration) attains its maximum. The time of the foot was plotted against the distance traveled by
the pulse wave. A linear regression fit was applied on the time–distance plot. The wave velocities were calculated as the reciprocal of the linear regression slope. To provide an indicator of the robustness of our method, the wave velocities were calculated on each back-and-forth motion (ten waves): as the vessel phantom was fixed in the water tank, the waves made round trips along the vessel wall.

The velocity of the pulse wave estimated along the axial and lateral directions is reported in Table IV for the stiff, medium, and soft vessel phantoms. As expected, the PWV increases as the stiffness of the vessel wall increases. Moreover, the same wave velocity values were found for the pulse wave along the axial and lateral directions. Because the longitudinal displacements were produced by the axial displacement, it is consistent to find the same velocity values. This validates the capability of our methods to estimate the Young’s modulus accurately.

The corresponding Young’s moduli were calculated using the modified Moens–Korteweg equation (9). The stiffness values were compared with those obtained by mechanical testing (Table V). The Young’s moduli were calculated from the PWV measured and by assuming a wall density of \( \rho = 1050 \text{ kg/m}^3 \). The average relative error between the mechanical testing and the PWV-based Young’s modulus was found to be 2.2%, illustrating the capability of our methods to estimate the Young’s modulus accurately.

V. Discussion

In this paper, two innovations are presented: the first innovation is the combination of ultrafast imaging, TOs, and a PBME. To produce high-frame-rate imaging sequences, plane waves are produced by exciting all the probe elements simultaneously. Note that in this study, no compounding scheme was implemented. The RF images are created by coherently coalescing the raw RF signals received from the transmission of only one plane wave. In this way, the RF images are not affected by motion artifacts during the compounding procedure, which provides a better estimation of the 2-D motion inside the phantom. TOs are used to improve the motion estimation in the lateral direction (i.e., perpendicular to the beam axis). In this study, TOs are produced using a filtering method. This technique controls the TO parameters better, especially the TO wavelength, which must be chosen carefully. The method is able to produce TOs with any lateral wavelength value, but so as not to keep only the noise of the image, the chosen value must be coherent with the frequency content of the initial RF images. In this study, the TO wavelength was chosen manually according to the frequency spectrum of the RF image used, and keeping the most energetics frequency spectrum part. This can be considered as a limitation of the method presented. In this paper, a 1.5-mm TO wavelength was used. The motion is estimated using a phase-based method. Traditional methods are usually based on spatial information to assess motion; the estimations are performed directly on pixel intensity values. Conversely, the approach presented aims at a subpixel motion estimation, with no interpolation, by exploiting the phase information of the RF images. Several methods have been reported in the literature to extract the phase of an RF image. Considering that the ROI defined in this paper features oscillating profiles in the two spatial directions, the 2-D local phase images are extracted using the 2-D analytic signal approach. Because the PBME algorithm uses the Matlab fast Fourier transform (FFT) algorithm, the axial and lateral ROI sizes should correspond to entire numbers of wavelengths. Here they are equal to two axial and lateral wavelengths. Indeed, if no interpolation is used, i.e., zero-padding, the FFT works correctly when the signal contains an entire number of oscillations. This approach combining ultrafast imaging, TOs, and PBME was applied to artery-mimicking phantoms, which leads us to the second innovation.

The second innovation is related to the visualization and the quantification of mechanical wave propagation.
in the lateral direction. We have shown that the propagation of longitudinal displacements could be observed along the vessel wall. In this study, the vessel phantom did not contain a pure longitudinal displacement. The lateral displacements measured were induced by the axial force produced by the axial displacement. In reality, the longitudinal motion is an image of the axial motion. It is therefore not possible to investigate and evaluate the velocity of a pure longitudinal wave (bulk wave) with the current experimental setup. However, we succeeded in finding the same propagation velocity for the axial and lateral directions, which allows us to be extremely confident in the capability of this method to estimate the propagation of a pure longitudinal wave. Concerning the estimation of vessel phantom wall stiffness, we have shown that the method can estimate the wall’s Young’s moduli using the estimation of the PWV. The results presented can be compared with those obtained by another group [18] on five different vessel phantoms, whose estimation of the Young’s modulus was obtained with average relative errors of 38%, which is around 17 times higher than the average relative errors found in this study. Three different points can explain why these results show better agreement with mechanical testing than those in [18]. First, the results from [18] were obtained with a frame rate of 500 fps, whereas our results were obtained with a frame rate of 10000 fps, and therefore the propagation of the pulse wave is much better described in the latter case. Second, classical 1-D speckle tracking to estimate axial motion was used in the reference, whereas the presented phase-based motion estimation, known to be very efficient in estimating small displacements, was used in our study. Finally, the results in reference [18] were obtained using a 1-D estimator, whereas a 2-D estimator was used in the study presented herein. The authors showed that in this vessel study the wall moves in two directions. Moreover, even if motion is sought in only one direction, it is not possible to perfectly estimate the motion in one direction without taking into account the motion in the other direction. In a future study, and for in vivo cases, a pure longitudinal wave is expected to be highlighted and investigated with this technique. If such a wave exists, it will be extremely useful to investigate it as a marker of cardiovascular risk. However, the propagation of this longitudinal wave is very fast, and it will be interesting to determine whether such waves can be visualized and quantified. The longitudinal wave velocity can be expressed as

\[ V_L = \sqrt{\frac{E(1 - \nu)}{\rho(1 + \nu)(1 - 2\nu)}} \]  

where \( V_L \) is the longitudinal wave velocity or the bulk wave velocity, \( E \) is the Young modulus of the medium, \( \rho \) is the density of the medium, and \( \nu \) is the Poisson ratio. For a healthy carotid, we can assume \( E = 100 \text{ kPa} \) and \( \nu = 0.495 \); the resulting longitudinal wave velocity is about 60 m/s. With this information, a 500 fps frame rate ensures the reliability of the measurement of a wave velocity about 10 m/s [18]. This reinforces our belief that the velocity of the propagation of the longitudinal motion on the carotid wall can be estimated in a future study.

Note that the estimation of the longitudinal motion should also better estimate the axial motion and, as a result, lead to better imaging of the more conventional axial pulse wave.

VI. CONCLUSION

In this paper, we have presented a new approach to estimate 2-D tissue motion in ultrafast imaging. It has been demonstrated that this method is able to estimate the propagation of longitudinal displacement. This method combines ultrafast imaging, TOs, and a phase-based motion estimation algorithm. The TOs are obtained by a filtering approach to better control the oscillation wavelength. By proceeding directly in the Fourier domain, we have shown that the 2-D FFT of the TO image can be easily controlled by a mask, producing RF images with TOs. First, the method was validated on simulation data. The 2-D motion, inspired from a real data set acquired on a human carotid artery, was applied to a numerical phantom to produce a simulation data set. The estimated motion showed axial and lateral mean errors of 4.2 ± 3.4 μm and 9.9 ± 7.9 μm, respectively. The experimental results obtained on three artery phantoms, with different wall stiffnesses, showed that the method presented is able to perform 2-D motion estimation very accurately even if the displacement values are very small and even in the lateral direction. Then, we showed that the approach is able to estimate the pulse wave velocity both in the axial direction and the longitudinal direction, which demonstrated the method’s ability to estimate the velocity of waves propagating in the lateral direction. Finally, the wall vessel phantom stiffness was investigated. The Young’s moduli of three vessel phantoms were estimated with an average 2.2% relative error. In a future study, this approach will be applied to in vivo cases.

REFERENCES


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