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Multiperspective Ultrasound Strain Imaging of the Abdominal Aorta

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Abstract— Current decision-making for clinical intervention of abdominal aortic aneurysms (AAAs) is based on the maximum diameter of the aortic wall, but this does not provide patient-specific information on rupture risk. Ultrasound (US) imaging can assess both geometry and deformation of the aortic wall. However, low lateral contrast and resolution are currently limiting the precision of both geometry and local strain estimates. To tackle these drawbacks, a multiperspective scanning mode was developed on a dual transducer US system to perform strain imaging at high frame rates.

Experimental imaging was performed on porcine aortas embedded in a phantom of the abdomen, pressurized in a mock circulation loop. US images were acquired with three acquisition schemes: Multiperspective ultrafast imaging, single perspective ultrafast imaging, and conventional line-by-line scanning. Image registration was performed by automatic detection of the transducer surfaces. Multiperspective images and axial displacements were compounded for improved segmentation and tracking of the aortic wall, respectively. Performance was compared in terms of image quality, motion tracking, and strain estimation.

Multiperspective compound displacement estimation reduced the mean motion tracking error over one cardiac cycle by a factor 10 compared to conventional scanning. Resolution increased in radial and circumferential strain images, and circumferential signal-to-noise ratio (SNRe) increased by 10 dB. Radial SNRe is high in wall regions moving towards the transducer. In other regions, radial strain estimates remain cumbersome for the frequency used.

In conclusion, multiperspective US imaging was demonstrated to improve motion tracking and circumferential strain estimation of porcine aortas in an experimental set-up.

Index Terms— Ultrasound, Multiperspective, Abdominal aorta, Strain imaging, Elastography.

I. INTRODUCTION

An abdominal aortic aneurysm (AAA) is a localized increase in the diameter of the abdominal aorta. Degeneration of the elastic properties and bulging of the

arterial wall leads to adaptive growth, further deforming the local geometric structure and morphology [1]. When the wall stress exceeds the wall strength, rupture occurs and the subsequent hemorrhage is associated with a mortality rate of >80% [2]. Clinical intervention is performed when the estimated rupture risk exceeds the risk of the procedure. The current estimator for rupture risk in the clinic is based on the anterior-posterior diameter of the AAA. The guidelines for surgical intervention advise surgery at aortic diameters of ≥ 5.5 cm for male, and ≥ 5.0 cm for female subjects according to evidence-based medicine [3]. Diameters are typically measured and monitored with two-dimensional (2D) transabdominal ultrasound.

However, some large diameter (≥ 5.5 cm) AAAs remain stable for a long period of time, whereas the risk of rapid growth causes uncertainty regarding the management of AAAs in the range of 4.0 - 5.0 cm in diameter, despite their statistically low rupture risk [4]–[6]. For this reason, an additional, more patient-specific approach is required to improve patient-selection for clinical intervention. It has been found that the distensibility [7] and local weakening [8] of the wall play an important role in rupture risk assessment of AAAs. The assessment of these properties can be performed with the use of strain imaging, based on tracking the deformation of the arterial wall over the cardiac cycle. Next to geometrical and functional measurements, studies have demonstrated that the peak wall stress is an accurate, reproducible indicator for rupture risk of AAAs, which can be estimated by the use of finite element analysis (FEA) [9]–[11]. The input for these models requires the geometry of the AAA, the mechanical properties of the AAA wall of the subject, and blood pressure.

Typically, computed tomography (CT) imaging is used as the gold standard for geometry assessment of AAAs [9]–[11] in combination with population based material properties of AAA tissue for wall stress analysis [12]. However, the drawbacks of CT, such as the exposure to ionizing radiation and contrast agents limit its use for geometry assessment during longitudinal clinical trials. Moreover, CT imaging provides little or no temporal information, making it difficult to perform strain

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This paper has supplementary downloadable video clips available at <https://ieeexplore.ieee.org/document/9120073>. This includes mp4 files about the B-mode images, motion tracking, and strain estimations for all acquisition schemes.

Fig. 1. (a) Imaging plane of a cross-section of a blood vessel wall. Geometry and wall motion perpendicular to the ultrasound (US) beam (white regions) are more difficult to obtain due to low contrast and resolution. (b) By imaging the vessel from multiple perspectives with a second transducer, the geometry and motion can be estimated accurately for a larger segment of the vessel (orange regions).

imaging and estimate patient-specific material parameters for the AAA wall. Magnetic resonance imaging (MRI) has been used as an alternative imaging modality to obtain both the geometry for image based modeling [10] and cyclic wall strain estimates [13], [14]. The main disadvantages of MRI include the high costs and scanning time, moderate resolution, and motion and susceptibility related artefacts [15].

For this reason, US has been proposed as the imaging modality of choice, because it benefits from a high spatial and temporal resolution, it is cheap, and it has the ability to acquire real-time, three-dimensional (3D) images noninvasively. US has shown to be a reproducible method for geometrical measurements of AAAs [16], [17], and for functional measurements of AAA wall strain [18]–[20]. Moreover, with the use of time-resolved 3D-US, a more patient-specific wall stress analysis could be performed, whereby the global elasticity of the AAA wall was estimated by inverse FEA [21], [22].

The main limitation of US imaging is that the image contrast and resolution are lower in aortic wall segments that are aligned with the insonification angle. In these regions, the precision of AAA geometry and motion estimates is much lower and require heavy regularization [21]. Therefore, both ultrasound-based FE modeling and strain imaging of the aorta would significantly benefit from higher quality images and strain estimates.

The low lateral contrast and resolution in segments of the aortic wall could be improved by introducing multiperspective ultrasound, *i.e.*, imaging the aorta from multiple perspectives. This can be achieved by the addition of an extra transducer during the image acquisition to transmit and receive US beams from a different perspective as is illustrated in Fig. 1. Multiperspective US images can be compounded in order to increase the resolution, field of view, and contrast between the aortic wall and its surroundings [23]. Furthermore, similar to Hansen et al. [24], high precision axial displacement estimations from two perspectives can be combined to obtain a more accurate motion and strain estimation over larger circumferential segments of the aortic wall. The latter is not feasible with a single probe, since the overlapping region in compounding at large depths would be too low. Finally, the implementation of ultrafast US acquisition schemes, *i.e.*, coherent compounding of diverging unfocused beams, can minimize image decorrelation by decreasing the transmit time between transducers [25]. This makes dual transducer strain imaging feasible.

The purpose of this study is to explore the benefits of using an additional transducer for US strain imaging of aortic tissue. To investigate this, a method for multiperspective ultrafast US imaging is compared in performance to single perspective conventional focused line-by-line imaging and ultrafast imaging. The multiperspective scanning mode was developed on a dual-probe, experimental US system. Moreover, a method was developed to perform multiperspective motion and compound strain estimation. The method's performance was assessed experimentally on pressurized porcine aortas in a mock circulation loop and model of the abdomen. Furthermore, a qualitative and quantitative comparison was made in terms of image quality, motion tracking, and strain estimation precision between the conventional method, the current state-of-the-art, and the proposed method.

II. MATERIALS AND METHODS

A. Experimental set-up

For this study, thoraco-abdominal porcine aortas ($N = 3$) were used for experimental imaging due to their accessibility and their similarity in geometry and material properties to human aortas [26]. These aortas, originating from healthy young pigs between 5 and 7 months old and weighing between 100 and 120 kg, were obtained from the local slaughterhouse right after excision and stored in phosphate-buffered saline (PBS) solution. Most of the excessive tissue was removed and the aortas were individually stored in PBS in a -20° freezer, thereby preserving the mechanical properties of the tissue until experimental measurements [27], [28].

Prior to inflation testing, one aorta was thawed in lukewarm water ($\sim 30^\circ$ C) and remaining connective tissue was removed. Side-branches were closed firmly using sutures. Next, the aorta was tested for leakage and persisting leakages were patched using tissue glue (Locite superglue-3 PowerFlex, Henkel, Germany).

After preparation, the aorta was fixated in a mock circulation set-up. A longitudinal pre-stretch of 1.3 was applied, mimicking physiological conditions [29], [30]. The mock circulation set-up, shown in Fig. 2a, has been previously used and described by Mascarenhas et al. (2016) and van Disseldorp et al. (2020), and aims to mimic *in vivo* hemodynamics by inflating the aortas at a physiological pressure range [31], [32]. During the inflation experiments, the mock-circulation set-up was filled with saline solution of 9 g/L NaCl to obtain an osmolality in the same range as blood [33]. Physiological saline solution was pumped through the aorta by a piston pump with a pulse interval of 0.8 s. A mechanical valve before and after the pump ensured unidirectional flow. The distal side of the aorta was connected to a 3-element Windkessel afterload model to obtain physiological pressures within the aorta [34]. This model was realized by the addition of a vertical column storing saline solution as a compliant element (C), and an adjustable inlet (R_{in}) and outlet (R_{out}) resistance. The pressure was monitored with a pressure wire inside the aorta, connected to a PC with a LabView interface (National instruments, Austin, TX, USA). Resistances were adjusted such that the pressure in the aorta was 130 - 70 mmHg during the inflation measurements.

The porcine aortas were embedded in a phantom of the abdomen, designed to add realism (compared to inflation in

Fig. 2. (a) Schematic overview of the mock circulation set-up used for inflation testing of porcine aortas. The aorta is fixated between two cannulas inside a container, where the inflow side of the aorta is connected to a piston pump circulating physiological saline solution. The distal side of the aorta is connected to a three-element Windkessel model and the pressure is monitored using a pressure wire inside the aorta. A 3D printed spine and gelatin solution were added to mimic the surrounding tissues in the abdomen. Ultrasound data were acquired with two curved array transducers. (b) Abdomen phantom consisting of a gelatin-filled container embedding the aorta, spine, and ultrasound transducers.

water) and mimic the mechanical properties of the surroundings of the human abdominal aorta. This phantom was created by embedding the aortas in a gelatin-filled container including a recreated human spine (Fig. 2b). The addition of the spine will contribute to an inhomogeneous deformation pattern of the aorta [35], [36]. The 3D geometry of the spine was obtained by segmentation of the abdominal part of the human spine in a CT dataset using Mimics segmentation software (Mimics Research 17.0, Materialise, Leuven, Belgium). The spine was created using 3D printing (Polyamide 12, Shapeways Inc., Eindhoven, The Netherlands) and was positioned closely underneath the aorta. Inside the container, two notches were made for the positioning of two curved array transducers in the same plane at an angle of approximately 85° from each other.

After correctly positioning the spine, aorta, and transducers, a gelatin solution was added to the container, mimicking surrounding tissue. The gelatin solution was prepared by heating up tap water to 70°C on a hot plate stirrer while gradually adding 8 wt. % gelatin (300 bloom, FormX, the Netherlands). Once the gelatin was fully dissolved in the solution, 2 wt. % of silicon carbide (SiC) was added to the solution to function as an acoustic scatterer during the US acquisition, generating a speckle pattern in the images. A gelatin concentration of 8 % wt. was chosen such that the Young's modulus of the gelatin solution is approximately $25 \pm 10\text{ kPa}$ [37], similar to surrounding abdominal tissue [38]. Whilst being stirred, the solution was cooled down to 28°C and then poured into the container within the mock circulation set-up. The gelatin solution was left overnight to solidify, awaiting inflation experiments the next day.

B. Ultrasound acquisition

US image acquisitions were performed using a 256-channel Verasonics Vantage research ultrasound system (Verasonics Inc., Redmond, WA, USA). The system was equipped with two curved array transducers with a center frequency of 3.7 MHz (Verasonics C5-2v, 128 elements), a similar transducer type and frequency that is used for *in vivo* abdominal imaging. Three different acquisition schemes were designed: conventional

single perspective focused line-by-line imaging, single perspective ultrafast imaging, and multiperspective ultrafast imaging. These acquisition schemes are visualized in Fig. 3.

Multiperspective ultrafast imaging was performed by designing an interleaved scanning sequence between two curved array transducers, where each transducer performs a set number of transmits (N_{angles}) over different steering angles. In order to minimize decorrelation of US signal between the image acquisition of two transducers, high frame rates were required. This was realized by transmitting and receiving simultaneously with all elements of one transducer, thereby creating spherical waves. This allows for ultrafast imaging at the cost of lateral resolution in comparison to conventional scanning, whereby focused scanlines are transmitted line-by-line. To improve lateral resolution, multiple steered spherical waves were transmitted by implementing linearly increasing element-specific electronic delays and were coherently compounded. For each transducer, 11 steered spherical waves were transmitted and received ranging between -10° and 10° with respect to the transducer origin. The number of transmits was chosen according to a trade-off between image quality and frame rate [39]. The time between transmits was set to 0.25 ms to prevent interference between acquisitions. As a result, the frame rate of ultrafast multiperspective US acquisitions was 180 Hz for each transducer.

Single transducer image acquisitions were acquired with both ultrafast spherical wave imaging and focused line-by-line imaging, in the same perspective as the left-hand transducer. The image parameters for single probe ultrafast imaging were equal to the ultrafast multiperspective acquisition, using a frame rate of 180 Hz and 11 steering angles ranging from -10° to 10° . The image parameters for line-by-line imaging were set to closely mimic conventional, focused, line-by-line scanning as is typically performed with clinically used scanners. In this case, a moving sub-aperture of 14 elements ($\sim 7.1\text{ mm}$) was used to record 128 consecutive scanlines with a transmit focus positioned at the depth corresponding to the center of the aorta. The interval between focused scanline transmissions was set to

Fig. 3. Ultrasound acquisition schemes: (a) Single perspective conventional focused line-by-line imaging showing beam focus for number of lines (N_{lines}); (b) Single perspective ultrafast spherical wave imaging showing transmit field for number of angles (N_{angles}); (c) Multiperspective ultrafast spherical wave imaging showing an interleaved scanning sequence for two transducers that acquire N_{angles} for each frame.

0.25 ms, and acquisitions were performed at a frame rate of 30 Hz.

The received channel radiofrequency (RF) data were sampled at four times the transmit frequency and stored to perform image reconstruction after the acquisition. The RF channel data were reconstructed using the Verasonics system in polar coordinates on a sector shaped grid with respect to the transmit transducer. The speed of sound of the propagation medium was assumed to be 1510 m/s according to the gelatin concentration [40]. The pixel spacing of the reconstructed US images was equal to 0.05 mm in axial direction ($\sim 1/8 \lambda$) and 0.51 mm ($\sim$ transducer pitch) in lateral direction, increasing with imaging depth. The reconstructed data were stored as in-phase quadrature (IQ) data of which the real part was used for further processing in MATLAB R2019a (The MathWorks, Natick, MA, USA).

C. Image fusion

The images obtained with multiperspective ultrafast imaging were first registered to allow for image fusion. This was done by identifying the transducer surface of the non-transmitting transducer in each image using a shape-based feature extraction algorithm. The envelope detected images of both perspectives were binarized using locally adaptive thresholding, of which a resulting image is shown in Fig. 4a. A circular Hough transform was applied to the binarized images, where a circle is drawn and accumulated in a parameterized matrix (Hough space) for each positive pixel, having a radius r of 49.6 mm, equal to the radius of the curved array transducers. The parameterization of the 2D Hough space can be described by the following equation with (a, b) the coordinates of the Hough space, (x, y) the coordinates of the pixels in the binarized image, and the angle φ ranging between -180° to 180° :

$$a = x - r \cos(\varphi), \text{ and } b = y - r \sin(\varphi). \quad (1)$$

The spacing of φ was set to 0.1° ; the resolution of the Hough space was set to 0.13 mm x 0.18 mm (axial x lateral). Fig. 4b shows a visualization of the Hough space, described. The coordinates (a, b) of the highest voted pixel in the Hough space trace back to the midpoint of the detected curvature, in this case the transducer origin shown as an overlay in Fig. 4c. Using the

Fig. 4. (a) Binarized ultrasound image after envelope detection of RF data. The top right curvature shows the transducer surface of the non-transmitting probe. (b) 2D circular Hough space, where the maximum value traces back to the origin of the non-transmitting transducer. (c) B-mode image with an overlay showing the detected origin of the transducer surface.

transducer locations in both perspectives, the rotation and translation were derived.

The multiperspective images were incoherently compounded using a weighted compounding method to improve the segmentation of the aortic wall. After registration, the pixel coordinate systems of both imaging perspectives were aligned by performing a rigid rotation and translation of the coordinates of a single image. The pixel spacing remained the same after this transformation. To benefit from the high axial resolution of US imaging, a new Cartesian pixel grid was defined, with an axial and lateral pixel spacing equal to the axial pixel spacing of a single perspective reconstructed US image, being $1/8 \lambda$. The envelope images of both transducers were then interpolated onto this new grid using a scattered interpolant function.

A mask was designed to perform weighted compounding of the interpolated images in order to optimize the contrast-to-noise ratio between the aortic wall and its surroundings. The ultrasound resolution and contrast are optimal when the wave propagation is perpendicular to the surface of acoustic impedance transition between two different media. For this reason, the angle θ from the midpoint of the aorta with respect to the transducer origin was calculated. Fig. 5 provides a simple illustration of this angle. Using the angle θ for every pixel (i, j)

Fig. 5. (a) Illustration of the angle Θ , which represents the angle between each pixel and the transducer origin seen from the midpoint of the aorta. $\Theta = 0$ represents the segment of the wall that is perpendicular to the wave propagation. (b) The wall was divided into eight regions (R1-R8) for the analysis of image quality, motion, and strain in different wall segments.

in the interpolated images of each transducer, two masks were created using the following periodic function:

$$M(i, j) = \frac{1}{2} \cos(2\theta(i, j)) + \frac{1}{2}. \quad (2)$$

Next, the masks were made complementary by dividing each mask with the sum of both. The interpolated images were then multiplied with its corresponding mask and summed to obtain a weighted compound envelope image. Compound B-mode images were obtained after log compression and normalization by subtraction of the maximum pixel intensity of the dataset, similar to the display of conventional US imaging.

D. 2D displacement estimation

The aortic wall was segmented in the compounded B-mode image in end-diastole by manually selecting the border between the inner wall and the lumen of the vessel. Using these selected points, a mesh of the aortic wall was created with a uniform wall thickness of 1.7 mm. As speckle tracking works more accurately with RF data, the coordinates of this mesh were transformed back to the coordinate systems of each transducer to perform motion tracking using both the envelope and reconstructed RF data, obtained with the individual transducers, *i.e.*, before fusion. The pixels in and surrounding the segmented aorta were tracked for the duration of one cardiac cycle. In the single perspective ultrafast and focused ultrasound images, the segmentation was obtained in a similar fashion.

The motion tracking was performed with a two-dimensional (2D) coarse-to-fine displacement estimation algorithm [41]. Using normalized cross-correlation as an index of similarity between frames, initial coarse displacements were estimated using kernels of envelope data within a confined search area (SA), followed by a fine displacement estimation of RF data with the use of more confined kernel size and SA. Please note that lateral kernel sizes are depth-dependent due to the diverging lines in the polar coordinate system of the RF-data. The size of the image kernels is a trade-off between tracking accuracy and precision and was based on image resolution and the thickness of the aortic wall. Based on the length of the transmitted waveform, two period cycles, the axial resolution is equal to 1λ , being 0.41 mm. The lateral resolution is estimated to be at least 1.4-2.9 mm, based on the aperture size used ($= 7.1$ mm) for a line-by-line acquisition at an imaging depth between 25 and 50 mm.

Fig. 6. Masks used for the radial displacement compounding estimated by multiperspective ultrasound acquisitions. On the left, the mask used on the displacement fields of transducer 1 is shown. Right shows the complementary mask used on the displacement fields of transducer 2. The dotted lines represent $\Theta = 0$, where the midpoint of the aorta connects to the transducer origin.

The kernel size and SA consisted of a fixed number of pixels. The coarse axial kernel size was set to cover the full wall thickness, while the fine kernel size was set to approximately half the wall thickness to be able to measure radial strain. The lateral kernel size was set to be at least as large as the estimated lateral resolution at the highest tracking depth, to include enough characteristic image features for motion tracking. For ultrafast US acquisitions, image kernels of 2.6 mm x 4-5.3 mm (axial x lateral) were used in a SA of 3.1 mm x 4.8-6.3 mm to estimate coarse displacements, and image kernels of 0.8 mm x 2.4-3.2 mm were used in a SA of 1.0 mm x 3.2-4.2 mm to estimate fine displacements. For focused US acquisitions, similar kernel sizes were used and the SA was increased to 3.7 mm x 7.1-9.5 mm for coarse displacements, and 1.3 mm x 4-5.3 mm for fine displacements. Axial and lateral displacements were filtered using a median filter of 0.6 mm x 4-5.3 mm (11 x 5 pixels).

E. Strain imaging

The resulting axial displacements of the multiperspective ultrafast imaging acquisitions were combined with weighted compounding masks to improve the tracking of the aortic wall. Estimated lateral displacements were not used, due to their comparatively low accuracy compared to axial displacements. First, axial displacements μ_{ax} were converted to radial displacements μ_{rad} using the angle from the midpoint of the aorta with respect to the transducer origin Θ .

$$\mu_{rad} = \frac{\mu_{ax}}{\cos(\theta)}. \quad (3)$$

The radial displacement fields of each perspective were aligned and interpolated on a new grid. Next, a mask was designed that combines the most accurate displacement estimations of each transducer, which is at wall regions where the displacement aligns with the insonification angle. To achieve this, a higher weight was assigned to the most accurate estimated displacements at $\Theta = 0^\circ$, yet rejects the $70 - 110^\circ$ regions where displacements can be overestimated because of the division by a small number in the conversion from axial to radial displacements. This two-dimensional mask M was described for the pixel positions (i, j) for $-180^\circ \leq \Theta \leq 180^\circ$. The coefficient of $18/7$ creates a half period of 70° and

was chosen to obtain smoother transitions at the cutout regions at $|\theta| = 70^\circ$ and $|\theta| = 110^\circ$.

$$M(i, j) = \begin{cases} \frac{1}{2} \cos(\frac{18}{7}\theta) + \frac{1}{2}, & 0^\circ \leq |\theta| \leq 70^\circ \\ 0, & 70^\circ \leq |\theta| \leq 110^\circ \\ \frac{1}{2} \cos(\frac{18}{7}\theta + 40) + \frac{1}{2}, & -180^\circ \leq \theta \leq -110^\circ \\ \frac{1}{2} \cos(\frac{18}{7}\theta - 40) + \frac{1}{2}, & 110^\circ \leq \theta \leq 180^\circ. \end{cases} \quad (4)$$

Each mask was then divided by the sum of both masks to create complementary masks. These resulting masks, also shown in Fig. 6, were multiplied with the interpolated radial displacements of each perspective and summed to obtain compounded radial displacement fields. Compounded radial displacements were converted back to axial and lateral displacements used to track the position of the segmented aortic wall between each frame. The conversion was performed using the following equations:

$$\mu_{ax} = \mu_{rad} \cos(\theta), \text{ and } \mu_{lat} = \mu_{rad} \sin(\theta). \quad (5)$$

Axial and lateral strains of the aortic wall were estimated for the single- and multiperspective datasets by means of a 2D least-squares strain estimator (2DLSQE) [42]. A strain kernel with a fixed size of 5 radial x 9 circumferential mesh points (1.7 mm x 4.6 mm) was used for single perspective displacements. A smaller strain kernel of 3 x 3 mesh points (1.0 mm x 1.5 mm) was used for the compounded displacements, to achieve higher precision at the possible cost of accuracy. The resulting axial and lateral strains were converted to radial and circumferential wall strain for further analysis with the use of the following equation [43]:

$$\begin{bmatrix} \epsilon_{rad} & \epsilon_{\gamma} \\ \epsilon_{\gamma} & \epsilon_{circ} \end{bmatrix} = R \begin{bmatrix} \epsilon_{xx} & \epsilon_{xy} \\ \epsilon_{yx} & \epsilon_{yy} \end{bmatrix} R^T, \quad (6)$$

where

$$R = \begin{bmatrix} \cos(\theta) & \sin(\theta) \\ -\sin(\theta) & \cos(\theta) \end{bmatrix}. \quad (7)$$

ϵ_{rad} and ϵ_{circ} are respectively the radial and circumferential strain, calculated from axial (ϵ_{yy}) and lateral wall strain (ϵ_{xx}) with rotation matrix R . ϵ_{γ} and ϵ_{xy} and ϵ_{yx} represent shear strain in polar and Cartesian coordinates, respectively.

F. Analysis of results

A qualitative and quantitative comparison was made between the three acquisition schemes in terms of image quality, motion tracking, and strain estimation. For comparison of different wall segments, the wall was divided into eight regions based on the incidence of the ultrasound wave, as illustrated in Fig. 5b. The increase in visibility of the aortic wall was quantified in the envelope detected images by the contrast-to-noise ratio (CNR) according to the following equation:

$$CNR = 20 \log_{10} \frac{|\mu_w - \mu_l|}{\sqrt{\sigma_w^2 + \sigma_l^2}}, \quad (8)$$

where μ_w and μ_l denote the mean signal of the wall and lumen respectively, and σ_w and σ_l the standard deviation of the signal

in the wall and lumen. To compare the motion estimation performance for each image acquisition scheme, the mean tracking drift error was estimated between the estimated positions of the middle wall layer before and after inflation. The drift error should decrease with a higher precision in the estimated displacements. The mean error (ME) was calculated using n segmented pixel positions (x_i, y_i) in the middle layer of the segmented wall in two frames; one at begin systole (bs) and one at end diastole (ed):

$$ME = \frac{1}{n} \sum_{i=1}^n \sqrt{(x_{i,bs} - x_{i,ed})^2 + (y_{i,bs} - y_{i,ed})^2}. \quad (9)$$

The precision of the strain estimations was quantified by calculating the elastographic signal-to-noise ratio (SNRe) using the mean (μ) and standard deviation (σ) of the end-systolic radial or circumferential strain, as measured in the middle layer of the segmented wall:

$$SNRe = 20 \log_{10} \frac{\mu}{\sigma}. \quad (10)$$

The middle wall layer was chosen as metric since the strain magnitude is expected to gradually increase towards the inner wall layer [44]. To verify the magnitude of circumferential wall strain, the global circumferential strain was calculated by the change in circumference between manual segmentation in begin systole and end systole.

III. RESULTS

A. Image quality

The B-mode images of one of the aortas embedded in the abdomen phantom, acquired with line-by-line, ultrafast, and multiperspective ultrafast US acquisitions is shown in Fig. 7a-c, respectively. The images show a cross-sectional view of the porcine aorta, positioned above the spine, resembling the illustration in Fig. 2b. The surface of the non-transmitting transducer is visible in the top part of the images, which was successfully used in both perspectives for the registration of the multiperspective ultrasound acquisitions of every acquired dataset.

The mean and standard deviation of the contrast-to-noise ratio of every aorta ($N = 3$) are listed in Table I. The single perspective images (Fig. 7a-b), both show normal bright reflections of the aortic wall at 10 and 5 o'clock positions (R1, R5), where the wall surface is perpendicular to the US propagation of the transmitting left-hand transducer. Shadow regions appear at the 3 and 7 o'clock positions (R3, R6), due to refraction and the absence of reflections at the gelatin-wall transition. The ultrafast image in Fig. 7b also shows beam width artefacts in the lumen artefact at the height of the spine as a result of the use of unfocused spherical wave transmissions. For this reason, the line-by-line acquisition outperforms the ultrafast acquisition in terms of image contrast, as supported by a 1 dB higher overall CNR (Table I), at the cost of a lower frame rate.

The fused images of the multiperspective ultrafast acquisitions, displayed in Fig. 7c, show bright reflections over a larger segment of the aortic wall compared to single

Fig. 7. B-mode images of the abdomen phantom visualizing the porcine aorta above the spine embedded in gelatin. Image acquisitions were obtained with (a) line-by-line imaging, (b) ultrafast imaging, and (c) multiperspective ultrafast imaging.

TABLE I

QUANTITATIVE RESULTS OF THE IMAGE QUALITY, MOTION ESTIMATES, AND STRAIN ESTIMATES FOR EACH REGION OF THE VESSEL WALL (R1-R8)

Measure *	Acquisition **	Total	R1	R2	R3	R4	R5	R6	R7	R8
CNR (dB)	SP Line-by-line	-8 ± 2	-4 ± 3	1 ± 1	-4 ± 4	-5 ± 6	-1 ± 1	-5 ± 5	-6 ± 8	-4 ± 5
	SP ultrafast	-9 ± 2	-4 ± 2	-2 ± 2	-8 ± 6	-3 ± 1	-1 ± 1	-10 ± 8	-12 ± 7	-10 ± 9
	MP ultrafast	-7 ± 1	-4 ± 2	-4 ± 0	-5 ± 1	-8 ± 1	-2 ± 1	-8 ± 3	-1 ± 1	-11 ± 6
ME (mm)	SP Line-by-line	0.52 ± 0.07	0.35 ± 0.15	0.54 ± 0.33	0.57 ± 0.06	0.75 ± 0.28	0.53 ± 0.09	0.54 ± 0.26	0.45 ± 0.19	0.34 ± 0.11
	SP ultrafast	1.37 ± 0.13	1.85 ± 0.22	0.99 ± 0.08	1.26 ± 0.35	1.88 ± 0.1	1.31 ± 0.73	1.26 ± 0.14	1.29 ± 0.28	1.15 ± 0.14
	MP ultrafast	0.05 ± 0.01	0.03 ± 0.02	0.04 ± 0.01	0.04 ± 0.01	0.03 ± 0.02	0.05 ± 0.02	0.07 ± 0.05	0.06 ± 0.03	0.06 ± 0.02
SNRe ^{cir} (dB)	SP Line-by-line	-3.0 ± 3.1	-12.3 ± 0.7	3.0 ± 3.0	9.1 ± 2.0	-4.1 ± 2.9	-7.5 ± 6.3	-11.2 ± 21.9	8.1 ± 7.0	7.6 ± 2.5
	SP ultrafast	-10.2 ± 3.6	-19.1 ± 11.5	1.8 ± 3.7	2.6 ± 12.3	-11.1 ± 10.9	3.2 ± 4.4	-9.1 ± 2.9	6.0 ± 4.7	-7.4 ± 9.1
	MP ultrafast	6.8 ± 1.0	23.8 ± 5.2	31.1 ± 4.2	30.3 ± 4.0	4.3 ± 18.5	17.1 ± 4.3	9.0 ± 2.7	19.5 ± 4.8	29.2 ± 11.7
SNRe ^{rad} (dB)	SP Line-by-line	-6.0 ± 11.6	10.1 ± 8.2	-0.5 ± 9.8	-1.6 ± 11.5	5.6 ± 16.1	7.0 ± 8.4	-2.7 ± 4.8	10.4 ± 2.3	1.7 ± 2.6
	SP ultrafast	-7.5 ± 2.8	2.5 ± 0.3	-5.6 ± 13.7	-2.5 ± 6.1	-5.1 ± 2.0	-1.3 ± 8.7	-3.9 ± 13.9	1.4 ± 9.9	-8.3 ± 12.9
	MP ultrafast	-7.5 ± 9.5	5.8 ± 3.0	-8.2 ± 16.4	5.3 ± 8.5	-0.3 ± 4.9	18.5 ± 3.3	-8.6 ± 21.7	0.7 ± 18.2	-0.7 ± 5.1

* CNR = Contrast-to-Noise Ratio, ME = Mean Error, SNRe^{cir} = Circumferential Elastographic Signal-to-Noise Ratio, SNRe^{rad} = Radial Elastographic Signal-to-Noise Ratio. ** SP = Single perspective, MP = Multiperspective. Values are reported as mean $\pm$ 1 standard deviation.

perspective images. At the top of the image, each transducer surface is visible and coincides with the field of view of each other. Although some beam width artefacts from multiperspective images are suppressed by the weighted image fusion, artefacts are still visible as clutter in parts of the lumen and at the depth of the spine. Despite the existing artefacts, the CNR of multiperspective ultrafast images surpassed the CNR of single perspective ultrafast images and, in lesser extent, that of focused line-by-line images.

Noise in the fused image can be expressed more strongly compared to the single perspective ultrafast image. This effect originates from the normalization step of envelope to B-mode data, where the log compressed envelope data is scaled to the highest pixel intensity value. The highest pixel intensity is typically lower in compounded images due to averaging, which thereby decreases the dynamic range of the B-mode image as shown in Fig 7c. This can also result in higher signal levels of the backscattered US in the gelatin.

One might argue that the occurrence of image artefacts originates from grating lobes, as beam steering of $-10^\circ - 10^\circ$ was performed with a wavelength / transducer pitch ratio of 0.81. To investigate this, we compared the single perspective 11-angle coherent compound image to a single transmit image without beam steering. These images are displayed in Fig. 8 and show that the image artefacts visible in the ultrafast images do

Fig. 8. Top-down B-mode images of the abdomen phantom comparing (a) an ultrafast coherent compound image of 11 angles $-10^\circ - 10^\circ$ beam steering with (b) a single transmit ultrafast image without beam steering. Beam width artefacts resulting from unfocused wave transmits remain and do not originate from grating lobes.

not disappear when no beam steering is applied. In fact, artefacts at acoustic boundary transitions appear brighter without the use of coherent compounding of unfocused wave transmits, demonstrating the presence of beam width artefacts.

B. Motion estimation

Fig. 9 visualizes the tracking error (drift) over a full cardiac cycle, also quantified by the mean error shown in Table I. The wall contours of the line-by-line acquisitions (Fig. 9a) show that both in systole and diastole, the wall coordinates are more uniformly distributed in the upper and lower wall segments than at the sides. This agrees with the expectation that motion tracking performs better in the axial direction, i.e., the

Fig. 9. Segmentation and 2D motion tracking results of the aortic wall in systole and diastole. (a, b) show the positions of the wall obtained by tracking line-by-line images and single perspective ultrafast images, respectively. (c) shows the wall positions tracked by compounded axial displacements of the multiperspective ultrafast images. Top icon serves as transducer position marker for regional analysis.

propagation of the ultrasound wave. The error drift increases by 0.1-0.4 mm at the sides of the aorta, where the wall motion is mainly in the lateral direction. The contours of the ultrafast motion tracking (Fig. 9b) have a similar shape but wall coordinates are more dispersed in the lateral direction, resulting in larger mean errors as noted in Table I. Fig. 9c shows how motion tracking performs using compounded axial displacements. The contours show uniformly distributed wall coordinates in systole and diastole, and the mean error in diastole decreases to 0.05 mm. In systole, an underestimation appears at the 6 o'clock wall segment, where the displacements tracked by both transducers weigh equally to the resulting estimated wall motion. The shadow regions at 4 and 8 o'clock ($80^\circ < |\text{top-down angle}| < 115^\circ$) are left out of the analysis. These regions benefit the least of the current dual-probe strategy due to the inter-probe angle and the luminal artefacts.

C. Strain imaging

For each acquisition scheme of one dataset, the circumferential and radial strain images are visualized as an overlay on top of the B-mode images in Fig. 10. The size of the strain kernels used is indicated by a red outline on the aortic wall meshes. The SNRe was calculated for circumferential and radial strain of every aorta and the means and standard deviations are summarized in Table I. The global circumferential strain for the displayed aorta was measured to be 0.08 ± 0.01 .

In the single perspective line-by-line image in Fig. 10a, an average circumferential strain was measured of 0.04 ± 0.06 . The circumferential strain pattern is most uniform at the lateral sides of the aorta (R3, R6), where the direction of strain is equal to the US wave propagation and SNRe reaches up to 9 dB. A similar trend is seen in the circumferential SNRe of ultrafast images, although the overall magnitude is lower due to a larger error drift and lower image contrast. When performing strain imaging using the compounded axial displacements from multiperspective ultrafast acquisitions, the overall circumferential SNRe increases from -3.0 ± 3.1 dB to 6.8 ± 1.0 dB. Regional analysis shows that strain precision is highest at wall regions that move into the direction of the US propagation from the left-hand (R1, R5) and right-hand (R3, R7) transducer. In the compound strain image in Fig. 10c, an average circumferential strain was measured of 0.07 ± 0.03 , agreeing with the globally measured circumferential strain. The difference in strain magnitude between the upper and lower

Fig. 10. (a-c) Circumferential and (d-f) radial strain images of the aortic wall in systole obtained with (a, d) line-by-line imaging, (b, e) ultrafast imaging, and (c, f) multiperspective ultrafast imaging. The strain kernel size is depicted by a red outline on the wall mesh. Top icon serves as transducer position marker for regional analysis.

wall layer is caused by the spine, inducing inhomogeneous wall motion during inflation, which is also seen in the wall contours in Fig. 9.

In the radial strain imaging performance of line-by-line images, regional analysis also shows that strain precision is higher where the strain direction is equal to the US wave propagation (R1, R5). R4 and R7 also have a high SNRe, however the total variation between datasets is high with a SNRe of -6.0 ± 11.6 for the entire wall. The total SNRe of the single perspective ultrafast strain estimates is lower with a value of -7.5 ± 2.8 . The variation in SNRe is reduced, however the wall mesh and corresponding strain pattern (Fig. 10e) look irregular due to lateral motion drift. After weighted compounding of axial displacements, negative radial wall strains are measured at wall segments where the strain direction coincides with the US wave propagation (Fig. 10f). High SNRe is found at R1 and R5, and in lesser extent at R3 and R7. In the segments between the two transducers, the expanding movement of the inner wall layer is estimated to be lower than that of the outer layer, thereby resulting in local overestimations of radial strain. As a result, the total radial SNRe does not exceed that of other acquisition schemes. At non-overlapping regions where wall motion was aligned with the sound propagation of each transducer, difference in magnitude between wall layers can be distinguished in radial and circumferential strain estimates.

IV. DISCUSSION

In this study a dual transducer multiperspective US acquisition scheme was developed and used to image and estimate the wall strain of porcine aortas in a controlled experimental set-up. The value of using an additional transducer was demonstrated as well as the feasibility of incoherently compounding multiperspective images and axial displacements as a means for improving the imaging and characterization of aortic tissue. Moreover, a comparison was made with single perspective line-by-line and ultrafast imaging in terms of image quality, motion tracking, and strain estimation.

The incoherent compounding of multiperspective ultrafast

images increased the visualization of the aortic circumference, providing more information for the manual segmentation of the aortic wall. In the experimental phantom, the image contrast increased despite the occurrence of beam width artefacts in the lumen and the gelatin-aorta boundary. Although grating lobes were not the origin of image artefacts, the use of a transducer with a smaller element pitch would allow for beam steering at larger angles, suppressing these artefacts. Increasing the current number of steering angles might benefit the image quality but would also extend the acquisition time and thereby lead to more decorrelation of the US signal between transducers.

In motion tracking, it was shown that tracking single perspective line-by-line images outperformed the tracking of ultrafast images in terms of precision. This can be explained by the poorer lateral resolution of transmitting spherical waves and the higher number of frames during one cardiac cycle, in which potential block matching errors are accumulated. After the weighted compounding of axial displacement estimations from two perspectives, the error drift over one cardiac cycle decreased from 0.5 mm to 0.05 mm compared to conventional scanning. Although the exclusion of lateral displacement estimates decreases motion tracking errors, it is challenging to track the wall in shadow regions spanning a total of 19% on the circumference of the vessel wall. This was both due to the lack of US signal because of refraction and the present beam width artefacts. Acquisition time versus image quality is also a trade-off here, since large acquisition times could pose a problem in displacement compounding when each transducer visualizes a different part of the cardiac cycle, where displacements are not equal.

In strain estimation using single perspective line-by-line images, strain estimates were most precise at wall segments where the strain directions were in the same direction as the ultrasound propagation, as also described in Hansen (2009) and Hansen (2012) [43], [44]. Single perspective ultrafast strain estimates were lower in precision, with a SNRe drop of 7.2 dB circumferentially and 1.5 dB radially compared to line-by-line estimates. Multiperspective strain imaging improved circumferential strain estimates with a SNRe increase of 9.8 dB opposed to line-by-line estimates, while using a smaller 2DLSQE kernel size of 3x3 pixels. The use of a smaller strain kernel size improves strain resolution, making it possible to distinguish strain magnitudes between different wall layers. This is on the condition that the kernel size for motion tracking is smaller than the wall thickness [42]. This effect was demonstrated most strongly in the non-overlapping wall segments where radial wall displacements were in alignment with the ultrasound propagation of each transducer. In these regions radial and circumferential strain magnitudes gradually decreased towards the outer border in agreement with theoretical strain patterns in homogeneous tubes with a concentric lumen [44].

In wall segments between transducers, overestimations were observed, especially in radial strain estimations. This occurs quickly, when performing vascular strain imaging using a relatively low transmit frequency of 3.7 MHz. In most cases, the tracking of the outer layer of the upper wall was affected by

the beam width artefact caused by the reflection of the gelatin-wall boundary, leading to a small overestimation of axial motion displacement. The motion tracking in the lower wall segment was affected by acoustic shadowing of the right-hand transducer, causing a small bulge in the inner wall layer. For this reason, radial strain SNRe from multiperspective images did not exceed that from conventional US-imaging in this abdomen phantom. The radial SNRe could be increased by using a larger strain kernel size but at the cost of the strain resolution.

Assessing the accuracies of the estimated strains would be essential to measure more exactly how much strain imaging in the aortic wall has improved. The lack of such verification is a common problem in experimental strain imaging studies and not specifically for this study. Verification experiments could include uni- or bi-axial tensile tests [45], ultrasound simulations, or the use of other imaging modalities for validation.

The created phantom of the abdomen allowed for the first acquisitions and processing of multiperspective abdominal strain images under dynamic conditions, mimicking physiological pressures and the mechanical surroundings of the abdominal aorta [32]. In terms of acoustics, an *in vivo* situation was recreated by use of curved array transducers at a clinically used transmit frequency, and a 3D printed spine and a scattering surrounding medium were added. The limitations of our current set-up were the relatively shallow depth of the aortas (~ 4 cm) and the homogeneity of the surrounding medium. The latter minimized speed of sound inhomogeneities which will be encountered *in vivo* but also resulted in more distinct transitions in acoustic impedance, causing beam width artefacts in ultrafast acquisitions. These artefacts affecting our image quality, motion tracking, and strain estimations, are expected to be less prevalent in *in vivo* measurements, where transitions in acoustic impedance appear more gradually and sound is scattered differently.

Because new acquisition schemes were developed and tested on an experimental US scanner, *in vivo* measurements on volunteers were yet beyond the scope of this study, due to safety regulations on medical devices and ethics approval. The next step will be to apply these techniques *in vivo* and investigate if strain imaging of the abdominal aorta can also improve with the benefits of multiperspective ultrafast imaging. However, the translation to *in vivo* will also introduce new challenges. User operability will become more difficult with handling two transducers. A solution could be to fixate one probe and use the other one freehand, to find a good viewing angle. However, in 2D it would be difficult to remain in the same imaging plane, requiring an extension to volumetric imaging. A good probe holder device or the use of an arch above the patient could also be used, providing additional information on the relative angle and distance between the transducers. The viewing angles will depend on the anatomy of each person, but we expect that especially less echogenic patients can benefit from additional perspectives as field-of-view is limited. Differences in speed of sound will likely introduce mismatches in the compounding of images and displacements, which could be resolved with the

use of speed-of-sound mapping [46]. The time-of-flight information obtained with a multi-transducer approach could also provide additional information about speed-of-sound relating to the transducer element positions and vice-versa.

In this study, image registration was performed by localization of the transducer surfaces through a shape-based feature extraction algorithm. The method performed well in this set-up; however, it is unconfirmed whether curved array transducer surfaces will be sufficiently visible *in vivo*. Still, detecting image features in multi-perspective images can be promising for image registration. Examples of useful features could include the shape of the spine and aorta, but also patient-specific artefacts such as bowel gas and calcifications.

Moving towards the use of multiperspective US imaging in the clinic could further increase the precision of vascular strain imaging of AAAs and/or FE analysis to analyze rupture risk. The increase in wall surface visibility, and precise motion and strain estimates, could decrease the regularization of the wall segmentation and use larger segments of the vessel wall for the estimation of global wall stiffness [21]. Also, the increase in strain resolution would allow for the estimation of local wall strains and stiffness, creating a more realistic simulation of wall stress in AAAs containing thrombus and calcifications [22].

V. CONCLUSION

In conclusion, multiperspective ultrasound imaging improves the performance of motion tracking and circumferential strain estimations after compounding of axial displacement fields. Optimization of ultrafast ultrasound imaging in terms of image acquisition and reconstruction is recommended to make more use of the multi-transducer acquisition set-up proposed. To make the step towards multiperspective strain imaging of AAAs *in vivo*, more research is required. Recommendations for future research include *in vivo* measurements, the testing development of image registration and speed of sound mapping techniques, and an extension towards multiperspective 3D imaging.

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