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Performance optimization of lateral displacement estimation with spatial angular compounding

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ABSTRACT

Elastography provides tissue mechanical information to differentiate normal and disease states. Nowadays, axial displacement and strain are usually estimated in clinical practice whereas lateral estimation is rarely used given that its accuracy is typically one order of magnitude worse than that of axial estimation. To improve the performance of lateral estimation, spatial angular compounding of multiple axial displacements along ultrasound beams transmitting in different steering angles was previously proposed. However, few studies have been conducted to evaluate the influence of key factors such as grating lobe noise (GLN), the number of steering angles (NSA) and maximum steering angle (MSA) in terms of performance optimization. The aim of this study was thus to investigate the effects of these factors through both computer simulations and phantom experiments. Only lateral rigid motion was considered in this study to separate its effects from those of axial and lateral strains on lateral displacement estimation. The performance as indicated by the root mean square error (RMSE) and standard deviation (SD) of the estimated lateral displacements validates the capability of spatial angular compounding in improving the performance of lateral estimation. It is necessary to filter the GLN for better estimation, and better performance is associated with a larger NSA and bigger MSA in both simulations and experiments, which is in agreement with the theoretical analysis. As indicated by the RMSE and SD, two steering angles with a larger steering angle are recommended. These results could provide insights into the performance optimization of lateral displacement estimation with spatial angular compounding.

1. Introduction

Static elastography is an imaging modality to provide information about tissue elastic properties which can be used to differentiate normal and diseased tissues [1]. Typically, a static or quasi-static external compression is applied on the biological tissue to introduce a distribution of displacements and strains within the tissue. The displacements are estimated from ultrasound radiofrequency (RF) signals, typically using a cross-correlation algorithm [2,3], and are then used to derive the corresponding strains [4,5]. The strain distribution is usually used to indirectly and semi-quantitatively indicate the distribution of elastic properties. Subsequently, the shear modulus or Young's modulus can be reconstructed from the displacement and strain distribution by solving an inverse problem [6,7]. Apart from static elastography, other elastographic methods, such as transient elastography (TE) [8], acoustic radiation force impulse (ARFI) imaging [9], shear wave elastography (SWE) [10,11], thermal strain imaging (TSI) [12], and shear wave dispersion ultrasound vibrometry (SDUV) [13,14], have also been developed in the past two decades to quantify the elasticity (and viscosity) of tissues.

Typically, only axial displacement and strain (along ultrasound beam) are estimated in static elastography and other elastographic methods. However, most biological tissues are nearly incompressible, i.e., compression in the axial direction will lead to expansion in the lateral (and elevational) direction [15]. It is thus important to estimate lateral displacement and strain to provide more comprehensive information on the tissue mechanical properties. To this end, several two-dimensional (2D) estimation methods have been developed to estimate both axial and lateral displacements, for instance, the cross-correlation method using one-dimensional (1D) kernel with a 2D search range [16], using 2D kernel with a 2D search range [17], and recorrelation method [16,18]. Nevertheless, axial displacement and strain estimates are typically used in

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current clinical practice given that they are more accurate than lateral estimates (typically by about one order of magnitude) [19]. That is mainly because of the lower resolution of ultrasound and lack of phase information [i.e., oscillations in the point spread function (PSF)] [20] and lower sampling frequency (i.e., low line density) in the lateral direction [21]. As such, several methods have been proposed to improve the lateral displacement estimation, e.g., the transverse oscillation (TO) strategy which adds oscillations in the PSF in the lateral direction [20] and the method based on tissue incompressibility which estimates the lateral displacement from axial displacement [22]. However, these methods are limited by the deteriorated lateral resolution [20], and isotropic and plane strain assumption [22], respectively.

As an alternative, spatial angular compounding has also been proposed to improve the performance of lateral displacement, to reduce noise in axial displacement estimation, and to reconstruct the displacement vectors and strain tensors [23–32]. In this method, ultrasound beams are transmitted in different steering angles, and the axial displacements corresponding to different steering angles are estimated separately and then are combined together to reconstruct the lateral displacements. Three-dimensional (3D) displacement vectors have also been obtained with 2D array by using spatial angular compounding [33]. Although, this method has been applied in noninvasive strain imaging of vessel wall [34] and complex vector flow imaging [35,36], the optimal settings, including the number of steering angles (NSA), the maximum steering angle (MSA) and the associated grating lobe noise (GLN), remain unclear. The aim of this study is therefore to investigate the impacts of these factors on lateral displacement estimation using spatial angular compounding through both computer simulations and phantom experiments.

2. Methods

2.1. Simulations

To investigate the performance of lateral displacement estimation with or without filtering the GLN and with different steering angles (i.e., MSA and the NSA), four linear array configurations, i.e., pitch = 304.8 μm and f_0 = 5 MHz, pitch = 304.8 μm and f_0 = 7.5 MHz, pitch = 135 μm and f_0 = 7.5 MHz, pitch = 197.9 μm and f_0 = 6.25 MHz, were simulated in Field II [37,38], as shown in Table 1. To investigate the optimal settings in practice, these four configurations were chosen according to four typical linear array transducers in clinical practice. A homogenous phantom with a scatterer density of 12.5/mm^3 underwent a series of lateral (horizontal) rigid motions, from 0 mm to the pitch value with a step of one tenth of the pitch. At each step, the phantom was imaged by conventional focused imaging (transmit focal depth = 20 mm) and plane wave imaging, with 9 steering angles from −20° to 20° with an increment of 5°. The simulated signals of the phantom were confirmed to follow a Rayleigh distribution [39,40]. Fig. 1 shows the sketch of focused imaging without [Fig. 1(a)] and with steering the transmit beams [Fig. 1(b)], and plane wave imaging without [Fig. 1(c)] and with transmit steering beams [Fig. 1(d)], respectively. For each imaging setup, the channel data were acquired at a sampling frequency of 100 MHz and were beamformed using the delay-and-sum (DAS) method with rectangular apodization and an f-number of 1.5. For all simulations, a speed of sound of 1540 m/s was used.

2.2. Phantom experiments

A Verasonics Vantage 256 System (Verasonics Inc., WA, USA) equipped with an L12-5 transducer was used for experimental investigations. The configuration of this transducer was the same as configuration IV in Table 1. The transducer was fixed on a computer-controlled positioner (Winner Optics. Beijing, China) to image a homogenous region of an elasticity QA phantom (CIRS 049, Norfolk, VA, USA) placed on an optical table [Fig. 2(a)]. The surface of the phantom was covered with distilled water for acoustical coupling. The transducer surface was positioned below the surface of the distilled water without contacting the phantom. The transducer was laterally moved step by step, with a step size equal to one tenth of the pitch. The maximum movement was equal to the pitch. At each step, 9 plane waves steered from −20° to 20° with an angular increment of 5° were transmitted with 128 consecutive elements of the transducer (because only 128 independent channels at one transducer port can be used for transmitting). To obtain the largest possible field-of-view after compounding, three aperture positions were designed for the negative, zero, and positive angles, respectively [Fig. 2(b)]. The 128 elements at the central part of the transducer were used when the steering angle was zero, while the 128 elements on the right side (or left side) of the transducer were used when the angle was negative (or positive). The pulse repetition frequency (PRF) was 2 kHz and the imaging depth was 40 mm. For each steering angle, the channel data were acquired at a sampling frequency of 50 MHz and were beamformed offline as described in Section 2.1. Please note that only plane wave imaging was experimentally tested with taking the practical application into account, because high frame rate is helpful for elastography using spatial angular compounding, and also because of the limited memory space of the ultrasound system for channel data acquisition.

2.3. Displacement estimation with spatial angular compounding

In spatial angular compounding, lateral displacements are reconstructed from axial displacements of multiple steering angles. For both simulations and phantom experiments, assuming only in-plane motion, the displacements along the receive beam direction (i.e., axial displacements) at different steering angles were estimated by an RF based 2D normalized cross-correlation method [19] (window size = 1 × 1 mm^2, axial overlap = 80%). All RF data were corrected to the new data grid for skewness using linear interpolation, which is necessary for displacement estimation for steered imaging [41,42]. Thereafter, the vertical displacements u_{ver} and horizontal displacements u_{hor} (i.e., lateral displacements)

Table 1
Transducer configurations.

Parameter	Setting I	Setting II	Setting III	Setting IV
f_0 (MHz)	5	7.5	7.5	6.25
Wavelength (μm)	308	205.3	205.3	246.4
Pitch (μm)	304.8	304.8	135	197.9
Number of elements	128	128	256	192

f_0, central frequency.