Program Digest, 2019 IEEE International Ultrasonics Symposium (IUS)
Glasgow, Scotland, October 6-9, 2019

3D Fast Ultrasound Imaging Through Pulse Compression: An Experimental Study

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Abstract— **3D fast/ultrasound imaging has emerged in the last years in research but still suffers from its poor image quality. Indeed, using plane/diverge waves does not permit to insonify the medium with sufficient energy at each point to get a good signal-to-noise ratio, contrast-to-noise-ratio or even resolution. On the other hand, coded excitation is currently used to increase signal-to-noise ratio and penetration depth. In this work, the objective is to combine 3D fast/ultrafast imaging with coded excitation to achieve better image quality at a high acquisition rate. Promising experimental results are obtained from both wire and cyst phantoms using a chirp excitation signal. The contrast-to-noise ratio and signal-to-noise ratio were improved by 4 dB and 2 dB respectively by the proposed method in comparison to the conventional way to do 3D imaging using a standard transmit. The improvement of the axial resolution of about 17% is the third important result obtained by the developed method still in comparison with the classical method. Experimental results show that an effective implementation on a research scanner of 3D coded excitation using plane wave imaging is possible.**

Keywords—3D, Pulse compression, Plane wave, Chirp.

I. INTRODUCTION

2D ultrasound imaging permits to get good image quality but suffers from geometry/placing issue since organs can be of irregular geometry in 3D. Moreover, for motion estimation technique, estimates can be affected by out-of-plane motion [1]. In the last few years, 3D fast/ultrafast ultrasound has emerged in research [2]. Unlike 2D imaging, it permits to directly visualize the 3D geometry of organs (e.g. heart or aorta) without using multiple 2D scans. Nevertheless, this new method generally suffers from its poor image quality due to both imaging technique and technological issues. The elements being smaller, the energy sent into the medium is weaker compared to 2D imaging; also, there is no elevation focus like in 2D imaging. That technology being relatively new, the image quality is certainly not optimal like for 2D imaging with linear array probe. To get better image quality, we propose to revisit the 3D fast ultrasound imaging by using coded excitations [3-4]. Extending the length and frequency range of the transmitted signal (without increasing its peak-to-peak energy or mechanical index) should improve image quality. With coded excitation, a pulse compression [5-7] is needed in reception to restore the axial resolution of the system. The proposed method is compared to standard 3D ultrasound imaging using sinusoidal transmit.

II. MATERIAL AND METHOD

A. Coherent plane-wave compounding (CPWC)

In plane-wave imaging, the entire region of interest (ROI) is insonified using a single plane wave shot. Backscattered echoes are combined off-line [8] to form a low resolution radiofrequency image. To improve the image quality, Montaldo *et al.* [9] proposed the Coherent Plane Wave Compounding (CPWC) approach for 2D imaging. The technique consists of transmitting multiple steered plane waves, which cover a sector angle $[\alpha min; \alpha max]$. In terms of resolution, SNR, and CNR, to obtain the same image quality as in standard focused imaging at the focal depth z_{foc}, the number N_{pw} of plane waves should be equal to [9]:

$$N_{pw} = \frac{x_{lat}}{\lambda F_{\#}} \qquad (1)$$

where x_{lat} is the lateral aperture size, λ is the transmitted pulse wavelength, and $F_{\#}$ is the F-number, which is defined as:

$$F_{\#} = \frac{z_{foc}}{x_{lat}} \qquad (2)$$

For small values of α, the steering angle for each plane wave i can be approximated by:

$$\theta_i = \arcsin\left(\frac{i\,\lambda}{x_{lat}}\right) \qquad (3)$$
$$\text{with } i = -(N_{pw}-1)/2, \ldots, +(N_{pw}-1)/2$$

However, some recent works [10] have shown that the image quality in terms of spatial resolution, SNR and CNR improves slightly when the number of plane wave insonification exceeds 9 in 2D imaging. CPWC is also used for 3D imaging to get better image quality, it is almost mandatory in practice since unfocused 3D imaging suffers from poor image quality.

B. Pulse compression using chirp excitations

1. Chirp definition: Coded excitation can be done with a chirp transmit. The analytical formulation of a chirp signal is [5-7]:

$$a(t) = C(t)e^{j2\pi\phi(t)} \tag{4}$$

Where $C(t)$and $\phi(t)$ are respectively the amplitude and the phase modulation functions. $\phi(t)$ is defined as:

$$\phi(t) = \left(f_c + \frac{B}{2T}t\right)t \text{ with: } -\frac{T}{2} \leq t \leq \frac{T}{2} \tag{5}$$

where f_c is the center frequency, B is the sweeping bandwidth, T is the duration of the signal, and $f_c - (B/2)$ is the starting frequency. The instantaneous frequency is given by:

$$f_i(t) = \frac{d\phi(t)}{dt} = \frac{d\left[\left(f_c + \frac{B}{2T}t\right)t\right]}{dt} = f_c + \frac{B}{t}t \tag{6}$$

As seen in (6), the instantaneous frequency of the chirp is a linear function of time thus its name as linear frequency modulated signal. Although a contradiction can exist between the notion of instantaneous frequency and the Fourier theory (which considers that a signal must be infinite in time to be limited in band), in practice, the term f_i indicates the spectral band in which the signal energy is concentrated at the time instant t. The parameter $K = B/T$ is the frequency modulation rate. Thus, the signal sweeps linearly the frequencies in the interval $\left[f_c - \frac{B}{2}, f_c + \frac{B}{2}\right]$.

2. Pulse compression: After transmitting a linear chirp inside a medium, the received echoes must be compressed to restore a good resolution in the final image. This filtering step (also called pulse compression) is achieved by correlating the received signals with the transmitted chirp. Several pulse compression techniques with different advantages/disadvantage exist, and the most used of them are described in the following parts.
The theory of matched filter was developed in the 1950s to optimize target detection in radar systems. The matched filter is a linear time-invariant filter that works by correlating a known signal with an unknown signal to detect the presence of the known signal. A matched filter optimizes the probability of the detection by maximizing the echo Signal-to-Noise-Ratio (eSNR) in the presence of white Gaussian noise. Its impulse response is equal to the time reversal of the transmitted signal with a time shift τ_d. For the real signal $a(t)$, defined in (4), the impulse response of the matched filter is given by **[5-7]**:

$$h_{matched} = a(\tau_d - t) \tag{7}$$

The transfer function of the matched filter is the complex conjugate of the signal spectrum. It can be written as:

$$H_{matched}(f) = e^{-j2\pi f\tau_d} A^*(f) \tag{8}$$

The matched filter can be generalized taking into account the non-white noise in the imaging system. Let us call $N(f)$ the power density of this noise which depends on frequency. In this case, the expression of the transfer function becomes:

$$H_{matched}(f) = e^{-j2\pi f\tau_d} \frac{A^*(f)}{N(f)} \tag{9}$$

C. Quality metrics

1. Spatial resolution: The spatial resolution provided by an image can be evaluated in both directions of the image. Therefore, axial/lateral resolution is defined as the ability of the system to discern two scatterers in the parallel/perpendicular direction of the ultrasound beam. The axial resolution is strongly linked to the length (number of cycles, for sinusoidal signals) of the transmitted signal. Shorter is the pulse length, better is the axial resolution. The lateral resolution is also linked to the transmitted signal but also to the aperture and the geometry of the probe. The axial/lateral resolution is defined as the full width at half maximum (at -6 dB on a log-scale) of an axial/lateral line in the B-mode image. A smaller width at -6 dB corresponds to a better axial (lateral) resolution of the imaging system.

2. Contrast-to-noise ratio (CNR): The metric that quantifies the perceived difference between a target ROI of the image and its background is the CNR. It is defined as:

$$CNR = \frac{|\mu_{ROI} - \mu_{back}|}{\sqrt{\sigma_{ROI}^2 - \sigma_{back}^2}} \tag{10}$$

where μ_{ROI} (μ_{back}) and σ_{ROI} (σ_{back})represent the average and the variance intensity inside (outside) the target ROI, respectively.

3. Signal-to-noise ratio (SNR): The SNR is another image quality metric estimating the noise contribution which is defined as:

$$SNR = \frac{\mu_{ROI}}{\sigma_{ROI}} \tag{11}$$

In homogeneous anechoic ROIs of the ultrasound image, SNR could give a very good synopsis of the imaging system response to noise, since in such ROIs $\mu_{ROI}; \mu_{noise}$ and $\sigma_{ROI}; \sigma_{noise}$. Therefore, SNR will be a good indicator to quantify the relative proportion of signal and noise (including acquisition noise and artefacts: reverberation, cross-talk, beamforming effects, *etc.*) in the obtained

B-mode images. To calculate both SNR and CNR, the log-compressed B-mode images (with the mentioned dynamic range mentioned) were used.

4. Axial/Lateral Auto-Correlation Length (AACL/LACL): The AACL and LACL quantify (respectively, in the axial and lateral direction) the coherence of the interference and thus of the speckle in the final B-mode image. Those metrics are defined as [7]:

$$AACL[n,a] = \sum_{i=-\infty}^{+\infty} I[a,i+n]\, I[a,i] \qquad (11)$$

$$LACL[l,n] = \sum_{i=-\infty}^{+\infty} I[i+n,l]\, I[i,l] \qquad (12)$$

where I is the beamformed image (before envelope extraction). a and l are the indexes of the vertical and horizontal image lines. Since the decay of the autocorrelation main lobe depends on the speckle size, main lobe's full width at half maximum (FWHM) is a good indicator of the speckle size. In other words, smaller FWHM of the *AACL (LACL)* implies smaller speckle in the axial (lateral) direction of the image.

III. Experimental results

A. Experimental setup

Four Verasonics Vantage™ 256 system (Verasonics, Redmond, Washington, United States) were used in this study [11]. Each system is in charge of controlling 8-by-32 elements of a total of 32-by-32, 3 MHz Vermon ultrasound probe (Vermon, Tours, France). Acquisition parameters are described in Tab. I. Experiments were conducted using a Gammex phantom model 410 SCG (Gammex, Middleton, Wisconsin, United States) with cyst inclusions and wires. An acquisition scheme with a pulse repetition frequency of 100 Hz was used based on 7 angle transmissions on each axis, namely the lateral axis (x) and transverse axis (y) resulting in 49 angles in total (7 angles × 7 angles). Two different transmission signals are compared: (i) a standard 2.5 sinusoidal-cycles, and (ii) a 6.7 μs chirp transmission from 2.1 to 3.9 MHz (transducer bandwidth at -6dB).

Table I. Acquisition parameters

Parameter	Value
Matrix probe	
Number of elements	1024
Matrix size	32 × 32 elements
Pitch	300 μm
Central frequency	3.0 MHz
Bandwidth (-6 dB)	2.1-3.9 MHz
Transmission	
Angle range	[-10, 10°]
Number of angles	7 × 7 angles
Transmit pulse	Sinusoidal 3 MHz / chirp 2.1-3.9 MHz
Signal duration	2.5-cycles/20-cycles
Sampling frequency	12 MHz
Pulse repetition. frequency	100 Hz

B. Results

As displayed in Fig. 1, the results show a significant improvement of the image quality with the proposed approach when compared to the classical one. Indeed, the B-mode image obtained using the chirp excitation shows a better overall contrast in the three planes of space.

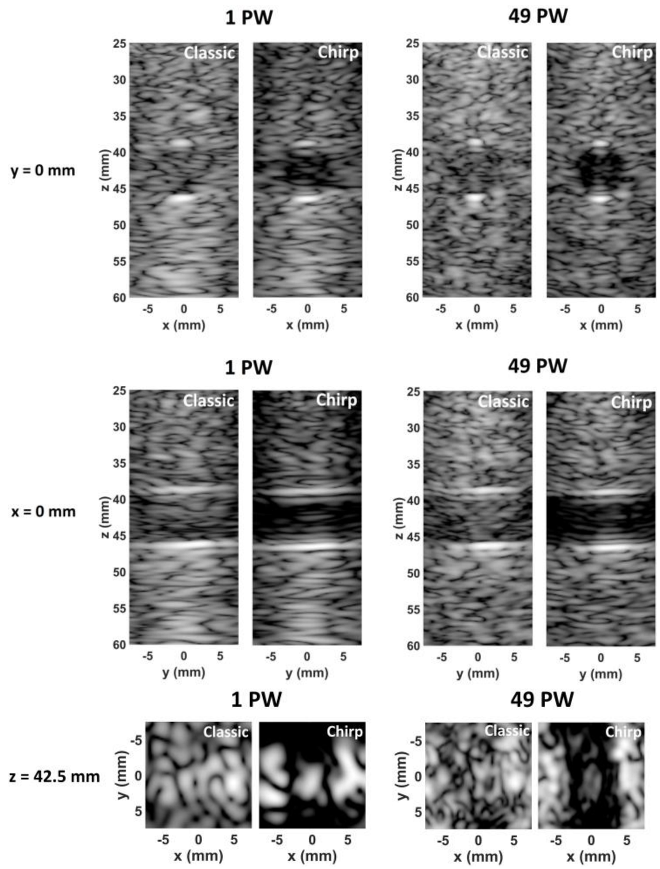

Figure 1: Obtained B-mode images in the three planes of space for both techniques using 1 plane wave (left) and 49 plane waves (right). All images have a dynamic range of 40 dB.

In order to quantify this observation, a study of the SNR and CNR as a function of the number of transmitted plane waves is carried out (Table II). Both imaging techniques yield better image quality in terms of CNR and SNR as the number of plane waves increases. For all forty-nine compounded plane waves, the developed method improves the CNR by 5 dB maximum for the considered cyst compared to the classical method. An improvement going up to 2 dB of the SNR is also provided by the new approach. It can also be noticed that only one plane wave with coded excitation provides better CNR and SNR values than forty-nine classical plane waves (Table II). Indeed, one chirp plane wave improves the CNR by 4 dB and the SNR by 1 dB in comparison to forty-nine classical plane waves. In other words, the frame rate can be increased 49 times using the 3D coded excitation method when the CNR and SNR metrics are considered.

Fig. 2 shows the B-mode images obtained from a wire phantom using 49 plane waves for both approaches. As it can be noticed, the wires and the speckle are thinner in the image obtained with the coded excitation technique. The measurement of the axial resolution at –6 dB confirms this visual aspect. Indeed, 17% enhancement is obtained when the proposed method is used (Classical: 0.66 mm, Chirp: 0.55 mm). The estimation of the axial autocorrelation length, which gives an idea on the speckle size in the axial direction, follows the same trend. The coded excitation method improves the AACL by 13 % in comparison to the classical approach (Classical: 0.78 mm, Chirp: 0.68 mm).

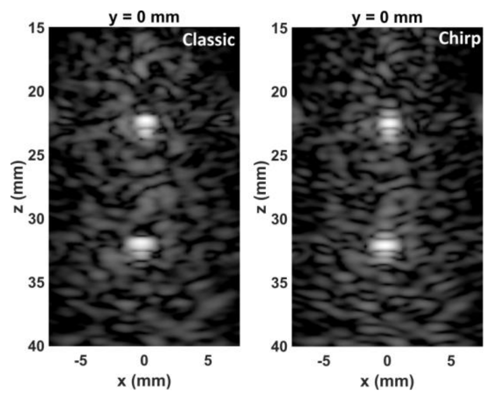

Figure 2: Obtained B-mode images in the plane (xz) for both techniques using 49 plane waves. Both images have a dynamic range of 40 dB.

The lateral resolution at -6dB and the lateral size of the speckle estimated (by the LACL) remain approximately the same regardless of the number of plane wave considered (Table II).

Table II. Results for the different metrics

	1 PW		49 PW	
	Classic	Chirp	Classic	Chirp
SNR (dB)	11,37	11,18	10,1	12,05
CNR (dB)	-18,49	-9,8	-13,92	-9,01
AACL (mm)	0,74	0,68	0,78	0,68
LACL (mm)	1,24	1,26	0,98	0,96
Axial Resolution (mm)	0,64	0,59	0,66	0,55
Lateral Resolution (mm)	1,48	1,52	1,18	1,16

IV. Conclusion

In this work, chirp transmission was implemented with a 3D plane wave sequence and compared against standard transmit. Thanks to pulse compression, the contrast-to-noise ratio is highly improved using coded excitation. Indeed, chirp excitation permits to obtain better contrast using only one plane wave than using 49 sinusoidal compounded plane waves. Such a technique could, in 3D imaging, allow better detection of hyper/hypoechoic region in tissue with a better image quality or a higher frame rate. Currently, 3D fast imaging is not widely used in medicine because of poor image quality. However, this approach applied in clinics would allow a more accurate detection of pathologies and volume estimation of tumors.

Acknowledgment

The Verasonics system was cofounded by the FEDER program, Saint-Etienne Metropole (SME) and Conseil General de la Loire (CG42) within the SonoCardioProtection Project headed by Prof. Pierre Croisille and Dr. Magalie Viallon, principal investigators. This study was also conducted as part of the LABEX CELYA (ANR-10-LABX-0060) and LABEX PRIMES (ANR-10-LABX-0063) projects of the Université de Lyon, within the "Investissements d'Avenir" (ANR-11-IDEX-0007) program operated by the French National Research Agency (ANR).

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