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Short Communication

Comparison of the performance of different tools for fast simulation of ultrasound data

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ABSTRACT

Simulation of ultrasound data is often performed for developing new ultrasound data processing techniques. The spatial impulse response method (as implemented in FieldII) has typically been used as the gold standard due to its excellent accuracy in the linear domain. When scatterer numbers become significant and when 3D volumetric data sets need to be computed, calculation time can become an issue however. In order to solve this problem, two alternative methods have recently been proposed both of which are based on the principle of convolving a set of point scatterers with a point spread function. "FUSK" operates in the frequency domain while "COLE" runs in the spatio-temporal domain. The aim of this study was to directly contrast both methodologies in terms of accuracy and processing speed using FieldII as a reference.

1. Introduction

Ultrasound imaging continues to expand its role in medical diagnostics and therapy guidance as it is an inexpensive, non-invasive and compact imaging modality that is readily available. The simulation of ultrasound data sets is regarded as a standard tool when designing new ultrasound systems or when validating new ultrasound processing techniques. An easy and fast methodology, that has been used extensively, is the so-called convolution model [1–3]. In the 1D implementation of this model, the point spread function (PSF) of the imaging system (i.e. the transmitted pulse) is spatially convolved with a Dirac-train which represents the position and the reflectivity of discrete scattering sites along the image line. This method has been used to generate 2D and 3D US data sets by extending the convolution to multiple dimensions and distributing the Dirac-functions in 2D/3D space. However, convolutions in 3D can become rather demanding in terms of memory requirements and computation time.

As an alternative simulation approach, the spatial impulse response method (as implemented in FieldII) has typically been used as the 'gold standard' due to its excellent accuracy in the linear domain [4,5]. However, when scatterer numbers become significant and when 3D volumetric data sets need to be computed, calculation time can become an issue. In order to get a better trade-off between accuracy and processing time, two alternative methods named "COLE" and "FUSK" were recently introduced [6,7]. The aim of this study was to directly contrast both methodologies in terms of their accuracy and processing speed using FieldII as a reference.

2. Methods

2.1. Simulation approaches

"COLE" and "FUSK" have previously been presented [6,7]. In brief, both methods are based on the principle of convolving a set of point scatterers with a point spread function. "FUSK" operates in the frequency domain while "COLEs" runs in the spatio-temporal domain.

COLE produces US data sets by convolving the transmitted ultrasound pulse with the projected amplitudes of all the scatterers along one image line in the spatial domain [6]. During the projection process the distance of the scatterer from the image line in combination with the lateral/elevation PSF characteristics is taken into account. In this way, the conventional 2D/3D convolution problem reduces to multiple 1D convolutions (one for each line) by un-coupling the axial from the azimutal/elevation characteristics of the PSF. Besides of analytical expressions of the PSF, COLE allows the integration of various simulated or measured beam profiles as a lookup table (LUT). As such, it generates cardiac synthetic ultrasound data sets with more realistic properties.

FUSK is a method to simulate 2D/3D ultrasound images in a two-step simulation process [7]. First, the scatterers are low pass filtered and fitted to a grid which does not need to be larger than the beam-space grid. Then, the object is converted to its *k*-space

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counterpart by a 3D Fourier transformation. The point spread function of the imaging system is calculated using the Fraunhofer-approximation which roughly states that the ultrasound field in the focal point equals the Fourier transform of the aperture for each frequency of the transmitted pulse. This approximation has been tested to perform well with the typical setting where the transmit aperture is smaller than the receive aperture [7]. The baseband *k*-space representations of the object and the point spread function are multiplied, corresponding to a spatial convolution, and subsequently inverse Fourier transformed, to yield the final simulated image.

2.2. Simulation settings

All simulations were performed using an ultrasound system with typical settings in echocardiography as shown in Table 1. A 64 crystal phased array transducer measuring $10 \times 14\ \text{mm}^2$ and transmitting a 2.5 MHz Gaussian pulse with a −6 dB relative bandwidth of 60% was used for all simulations.

2.3. Image reconstruction

As COLE, FUSK and FieldII do not give absolute reflection amplitudes, the simulated RF data sets were normalized prior to further processing and comparison. Given that all simulation models are linear, the reflected energy of a single point scatterer positioned in the focus point was hereto calculated and used as a normalization factor. Subsequently, the RF signals were envelope detected using a Hilbert transform, log-compressed and shown in a scan converted gray scale image with a dynamic range of 40 dB.

2.4. Quantitative comparison of simulated data sets

2.4.1. Speckle statistics

(1) *First order statistics:* In order to test the first order statistical properties of the simulated images, homogeneous phantoms ($40 \times 40\ \text{mm}^2$) with scatterer density (600 scatterers per mm^2) well above the limit for fully developed speckle were generated and simulated using COLE, FUSK and FieldII with an unfocused beam in transmit and receive. It is known for such a set-up that the signal to noise ratio (SNR) reaches a theoretical limit of 1.92 and that the amplitude of the envelope signal follows Rayleigh statistics [8]. Therefore, the SNR as well as the sum-of-squared-difference (SSD) between the measured histogram and its Rayleigh fit were calculated. In order to take the stochastic nature of these measurements into account, these simulations were repeated 10 times.

(2) *Second order statistics:* An often used second order statistical image property is based on the gray level co-occurrence matrix (GLCM – $C(i, j)$) [9]. This is actually a two-dimensional histogram where each element $C(i, j)$ is the joint probability occurrence of pixel pairs (p, q) at a certain pixel distance $(\Delta x, \Delta y)$ having gray level values i and j in the gray scale image I:

$$C(i,j) = \sum_{p=1}^{n} \sum_{q=1}^{m} = \begin{cases} 1, & \text{if} \quad I(p,q) = i \text{ and } I(p+\Delta x, q+\Delta y) = j \\ 0, & \text{else} \end{cases}$$

The parameters contrast (CONTRA), correlation (COR) and angular second moment (ASM) have been defined as [9]

$$\text{ASM} = \sum_{i=0}^{v-1} \sum_{j=0}^{v-1} C(i,j)^2$$

$$\text{COR} = \sum_{i=0}^{v-1} \sum_{j=0}^{v-1} \frac{(i-u_i)(j-u_j)C(i,j)}{\sqrt{\sigma_i^2 \sigma_j^2}}$$

$$\text{CONTRA} = \sum_{h=0}^{v-1} h^2 \left\{ \sum_{|i-j|=h} C(i,j) \right\}$$

where

$$u_i = \sum_{i=0}^{v-1} i \sum_{j=0}^{v-1} C(i,j)$$

and

$$\sigma_i^2 = \sum_{i=0}^{v-1} \sum_{j=0}^{v-1} (i-u_i)^2 C(i,j)$$

with v the number of gray levels. ASM is a measure of the homogeneity of the image, while COR represents how correlated a pixel is to its neighbor over the whole image. Finally, CONTRA expresses the amount of local variations present in the image.

Ten standard software phantoms ($70 \times 20\ \text{mm}^2$) were generated using a scatterer density of 500 per mm^2. The lower scatterer density was used to avoid excessive CPU time. In order to add three cystic regions (20, 10 and 5 mm in diameter) these scatterers' scattering amplitude was reduced to 33% of the amplitude of the other scatterers. Each phantom was scanned by COLE, FUSK and FieldII using a beam focused at a depth of 80 mm during transmit and dynamically focused in receive. In order to describe the second order statistics of the resulting images, the GLCM (pixel distance 1) along the axial and lateral direction was calculated locally (mask 23×23 pixels) using a sliding window approach. Subsequently, the parameters (ASM, CONTRA, COR) were calculated and an average root-mean-squared-error (RMS) was extracted using the FieldII parameters as ground truth.

2.4.2. PSF evaluation

In order to study the difference of the local PSF profile, simulated RF signals and gray scale images of individual scatterers positioned at polar grids (14×5 measuring $70\ \text{mm} \times 36°$) were generated using COLE, FUSK and FieldII. The beam was focused at a depth of 80 mm during transmit and dynamically focused in receive. In order to express the similarity of the PSF generated by FUSK or COLE and the reference PSF generated by FieldII, the average SSD and the normalized cross-correlation (NCC) of the resulting images were calculated.

2.4.3. Computation time

A standard phantom ($70 \times 20\ \text{mm}^2$) containing three cystic regions (20, 10 and 5 mm in diameter) was generated 10 times at different scatterer densities (10–100–500 per mm^2) resulting in a total of 30 phantoms. Images were simulated of each phantom using FieldII, FUSK and COLE using identical ultrasound system

Table 1
Parameters for the simulation of the ultrasound data sets.

Transducer frequency	2.5 MHz
Sampling frequency	50 MHz
Ultrasound velocity	1540 m/s
Tilt angle	0°
Start depth	10 mm
Depth	100 mm
Size of ROI	45°
Number of lines	60