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Ultrasonics 42 (2004) 739–743
Ultrasonics
www.elsevier.com/locate/ultras

Feasibility of 3D harmonic contrast imaging

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Abstract

Improved endocardial border delineation with the application of contrast agents should allow for less complex and faster tracing algorithms for left ventricular volume analysis. We developed a fast rotating phased array transducer for 3D imaging of the heart with harmonic capabilities making it suitable for contrast imaging. In this study the feasibility of 3D harmonic contrast imaging is evaluated in vitro. A commercially available tissue mimicking flow phantom was used in combination with Sonovue. Backscatter power spectra from a tissue and contrast region of interest were calculated from recorded radio frequency data. The spectra and the extracted contrast to tissue ratio from these spectra were used to optimize the excitation frequency, the pulse length and the receive filter settings of the transducer. Frequencies ranging from 1.66 to 2.35 MHz and pulse lengths of 1.5, 2 and 2.5 cycles were explored. An increase of more than 15 dB in the contrast to tissue ratio was found around the second harmonic compared with the fundamental level at an optimal excitation frequency of 1.74 MHz and a pulse length of 2.5 cycles. Using the optimal settings for 3D harmonic contrast recordings volume measurements of a left ventricular shaped agar phantom were performed. Without contrast the extracted volume data resulted in a volume error of 1.5%, with contrast an accuracy of 3.8% was achieved. The results show the feasibility of accurate volume measurements from 3D harmonic contrast images. Further investigations will include the clinical evaluation of the presented technique for improved assessment of the heart.

Keywords: 3D; Echocardiography; Harmonic imaging; Contrast agent

1. Introduction

Ultrasound contrast agents (UCA), consisting of microbubbles, are used to enhance the echoes from a blood pool to which they are added [1]. In fundamental (B-mode) imaging, UCA increase the backscattering of blood to a level close to that of tissue. Discrimination between blood and tissue is therefore compromised which can be quantified with the so-called contrast to tissue ratio (CTR). Microbubbles, however, reveal a non-linear oscillation when insonified with a sufficiently high acoustic pressure wave [2,3]. As a result the backscatter signal of UCA contain the transmitted frequency and also its multiples or harmonics. Since tissue does not comprises a non-linear scattering behaviour of the same magnitude this offers an opportunity to distinguish between the backscatter signal from UCA and tissue. Unfortunately, harmonic frequencies are also generated by propagation of the ultrasonic wave through tissue and blood [4,5]. The backscatter of these generated harmonics from tissue will hamper the distinction of harmonics radiated by the microbubbles and reduce the CTR. However, it has been shown that CTR levels for second harmonic imaging are much higher than those of fundamental imaging [6].

Most array transducers used for medical imaging exhibit a wide bandwidth that covers the fundamental transmit frequency as well as its second harmonic. We developed a wideband fast rotating phased array transducer for harmonic 3D imaging of the heart suitable for contrast imaging [7].

Since it provides information not only from a single cross-section, as with 2D-imaging, but from the complete volume under investigation the advantages of 3D

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doi:10.1016/j.ultras.2003.12.012

echo are numerous. A few of these advantages are: accurate quantification of cardiac properties, extraction of 2D any plane images and dynamic 3D views of cardiac structures [8,9]. 3D contrast imaging can add to these advantages since tissue often exhibits non-homogeneous scattering resulting in drop out. This is especially problematic when border tracing algorithms are used for quantification of the recorded echo data. When used at an appropriate concentration, UCA give a homogeneous and complete opacification. This characteristic could be beneficial for the development of less complex and faster tracing algorithms through improved endocardial border delineation with UCA. In addition, current commercial UCA feature a longer lifetime (up to 15 min) compared to former generations. This makes the use of infusion unnecessary even for the longer acquisition duration needed for 3D echo and therefore improves the feasibility of 3D contrast imaging.

In this study the optimal scanning settings with maximum CTR for 3D harmonic imaging are explored. The optimal settings are extracted from in vitro investigation. Using these settings the feasibility of 3D harmonic contrast imaging is evaluated with volumetric in vitro measurements.

2. Material and methods

2.1. Fast rotating ultrasound transducer

The fast rotating ultrasound (FRU)-transducer consists of three major parts: a DC motor that drives the array, a slip-ring device with 82 contacts, establishing signal transfer to and from the rotating array, and a linear phased array [10].

The DC motor drives the array at a rotation speed ranging from 240 to 480 rpm and is connected to an external control system with a manual setting for the rotation speed. A 4D-dataset (three spatial dimensions and time) can be acquired in a short time (e.g. 1 s), resulting in a sparse sampling of the time–volume space. Longer acquisitions yield a much denser sampling. The typical acquisition time is approximately 10 s, which has proven to be convenient for clinical application.

The array of the transducer, custom made by Delft Instruments (Delft, The Netherlands), contains 64 elements with a pitch of 0.21 mm and is tapered into an octagonal shape, approximating a circle with a radius of 7 mm. It has a fractional bandwidth of 80% with a center frequency of 3 MHz. The fixed focus of the acoustic lens in the elevation direction was set at 60 mm.

The transducer is connected to a General Electric/VingMed (Horten, Norway) Vivid 5 system. For this study the third prototype of the transducer is used which is shown in Fig. 1. Improvements of this third prototype can be found in the wider bandwidth, allowing improved harmonic imaging, and the more ergonomic design of the transducer.

Fig. 1. Third prototype of the fast rotating ultrasound transducer.

2.2. Contrast to tissue ratio measurements

The contrast to tissue ratio (CTR) is defined as the ratio of the scattered power by the contrast to that of the tissue. A commercially available tissue mimicking flow phantom (ATS Laboratories Inc., Model 524, Bridgeport, Connecticut, USA) was used to measure the CTR at different frequencies ranging from 1.66 to 2.35 MHz. This was facilitated by the availability of the radio frequency (RF)-data from the scanner. The excitation amplitude of the array was identical for all frequencies and was set at a magnitude at which no contrast destruction was visual. On average the non-derated mechanical index (MI, peak negative pressure in MPa over the square root of the frequency in MHz) of the transmitted acoustic pulse was 0.4 (613 kPa). The transmitted pulse contained 2.5 cycles and RF-data was recorded at a frame rate of one frame per second. Sonovue (Bracco, Genéve, Switzerland) was used at a dilution of 1 over 1000 in NaCl and was flowing at a constant rate of 90 ml/min through the phantom. Two-way frequency spectra were calculated from two regions of interest (ROI), one in the tissue and the other in the flow area, at equal depth (approximately 5 cm). The excitation frequency resulting in the highest CTR was selected as the optimal excitation frequency. The same procedure was used to determine the optimal pulse length for the optimal excitation frequency. Pulse lengths of 1.5, 2 and 2.5 cycles were investigated. For reasons of resolution preservation larger pulse lengths than 2.5 cycles were not used. Ultimately the receive filter of the echo system was set according to the obtained CTR results.

2.3. Volume measurements

An agar phantom was constructed with the shape of a left ventricle (LV) and with carborundum as scatter particles (see Fig. 2). The volume of the phantom cavity was 131 ml. The backscatter level from the LV-phantom 'tissue' was higher than that of the flow phantom (56.8 and 52.6 dB, respectively at a MI of 0.4 (503 kPa)). A 3D harmonic contrast recording was made using the optimized settings for the excitation frequency, the pulse length and the receive filter. Sonovue was used at a dilution of 1 over 2000 in water. The amplitude of the transmitted acoustic signal was again set at a non-destructive magnitude. For the other recording, without contrast, the cardiac imaging settings of the scanner manufacturer were used (an excitation frequency of 1.82 MHz and a pulse length of 1.5 cycles). A frame rate of 59 and 65 frames per second and a MI of 0.2 (341 kPa) and 1.4 (1.9 MPa) were used for the recordings with and without contrast, respectively. The rotation speed for both recordings was set at 360 rpm. Approximately 2 s of each recording was used to reconstruct the phantom cavity using self-developed post processing software [11]. After reconstruction the phantom cavities were extracted with an advanced threshold-based method. For this method depth dependent bimodal histogram thresholding, a closing procedure with a diamond structuring element of size 3 and volume extraction using six connectivity were employed [12]. Finally the extracted volumes were compared with the real volume of the phantom cavity.

Fig. 2. Left ventricular (LV)-phantom which consists of two parts. The actual LV shaped part is shown on the left and the assisting part is shown on the right of the picture. Both parts are made from agar with carborundum as scattering particles. The assisting part contained a stirring device to assist a homogeneous distribution after UCA injection.

3. Result and discussion

To calculate the CTR for different frequencies the RF-data from the contrast and tissue ROI were selected as shown in Fig. 3A. Two-way frequency spectra from the two ROI were calculated of which an example is shown in Fig. 3B. Subtraction of the two spectra yielded

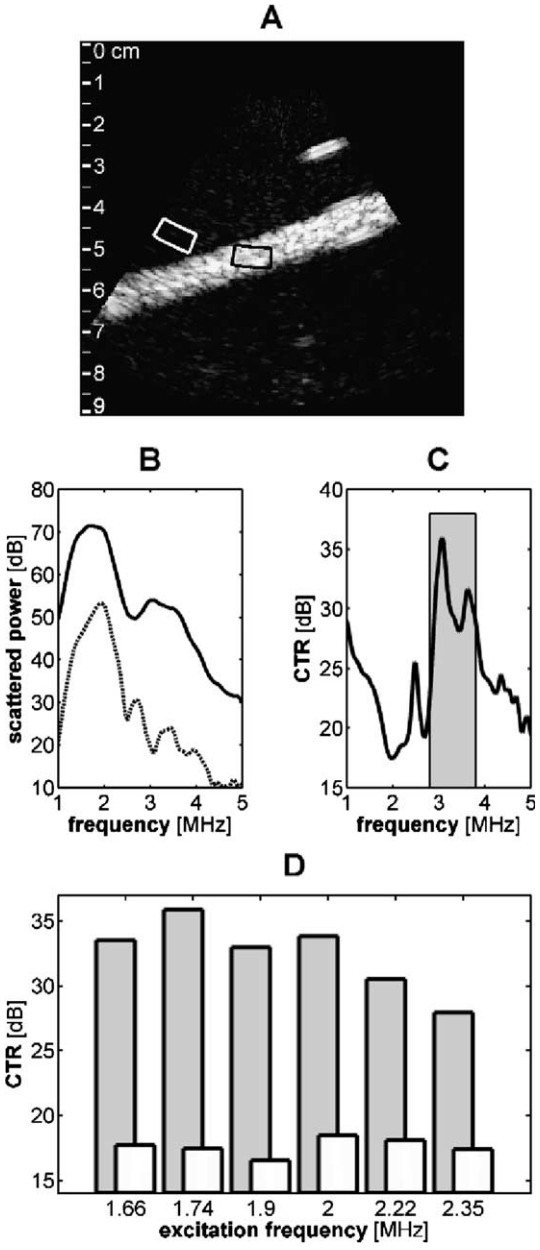

Fig. 3. B-mode image from the tissue-mimicking phantom with contrast in the flow area showing the regions of interest used for the power spectra calculations (panel A). Power spectra from the tissue (dotted line) and contrast (dashed line) region (panel B) and their difference (panel C) defined as the contrast to tissue ratio (CTR). Fundamental (open bars) and harmonic (filled bars) CTR as function of frequency showing the optimal excitation frequency at 1.74 MHz for harmonic contrast imaging (panel D). The receive filter settings for the optimal excitation frequency is indicated with a rectangle in panel C.

the CTR as function of frequency (see Fig. 3C). The highest CTR was found close to the second harmonic of the transmitted frequency. From Fig. 3C it can be seen that the CTR of the second harmonic is more than 15 dB higher than at fundamental frequency. Fig. 3D shows the harmonic and fundamental CTR for different excitation frequencies. An excitation frequency of 1.74 MHz was found to give the highest harmonic CTR. Pulse length variations at the optimal excitation frequency of 1.74 MHz resulted in an optimal pulse length of 2.5 cycles. For the optimal excitation frequency the receive filter was configured with a center frequency of 3.3 MHz and a bandwidth of 1 MHz giving the maximum harmonic amplitude (see Fig. 3C).

The optimized settings of the scanner for harmonic contrast imaging were used for a 3D recording of the LV-phantom with and without contrast. The extracted phantom cavity from the recording with contrast is shown in Fig. 4 along with the reconstruction of phantom 'tissue' from the recording without contrast. The volume of the extracted phantom cavity from the recording without contrast was 129 ml, resulting in a volume error of 1.5%. This is equal to the accuracy of our previous reported in vitro volume measurements [4]. The volume from the recording with contrast was 136 ml, resulting in a volume error of 3.8%. Although the accuracy of both extracted volumes is high the volume from the recording with contrast seems significantly higher than that from the recording without contrast. This can be explained by the phantom being slightly porous enabling contrast bubbles to invade the tissue. Another explanation can be found in a larger point spread characteristic of contrast backscatter. Both explanations however, need to be investigated in more detail.

Fig. 5 shows an example of left ventricular harmonic B-mode images with and without contrast made with a commercial scanner and its standard cardiac array transducer. In the image without contrast especially the apical segment on the left shows a vague delineation of the endocardial border while this segment is clearly visualized in the image with contrast. The same apical delineation improvement was observed by Kasprzak et al. [13]. Other studies involving 3D harmonic contrast imaging report promising reconstructions of kidney and liver vascularities [14,15].

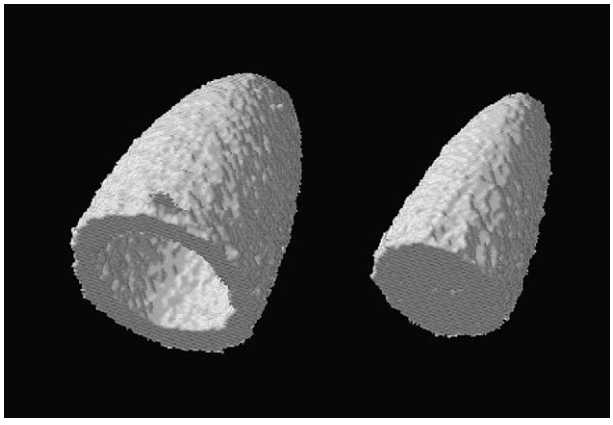

Fig. 4. 3D reconstructions of the left ventricular phantom 'tissue' and cavity. The 'tissue' was obtained from the recording without contrast and the cavity from the recording with contrast.

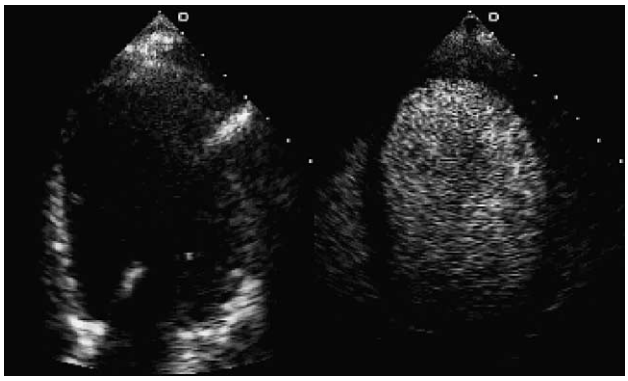

Fig. 5. Harmonic B-mode images from a commercial scanner, recorded with its standard cardiac transducer, without (left) and with contrast (right). In the recording with contrast the delineation of the apical segments is clearly improved.

In the preceding section accurate volume measurements from harmonic contrast recordings of the FRU-transducer have been proven feasible. Currently, the only commercial 3D echo system for cardiac imaging with harmonic capabilities is the Sonos 7500 with its ×4 ×MATRIX transducer from Philips. However, no record has been found on its harmonic contrast imaging performance yet.

4. Conclusion and future

Many advantages of 3D echo have been reported throughout the last decade. The introduction of 3D echo in the daily clinical practice has however only recently started. To make 3D echo a mature and widely spread technique more modalities other than fundamental and harmonic B-mode imaging have to be added. The results presented in this paper shows the feasibility of 3D harmonic contrast imaging as a possible addition to the existing clinical applications. As reported by Bouakaz et al. further improvements on 3D contrast imaging can be expected when harmonics higher than the second harmonic will be made available for imaging [6]. Application of these super harmonics along with a clinical evaluation of the currently presented technique will be the issue of our future efforts. By then it can be investigated whether the improved delineation of the endocardial border from harmonic contrast recordings is beneficial for (semi)-automated border tracing algorithms.

Acknowledgements

The authors would like to thank W.J. van Alphen, L. Bekkering and F.C. van Egmond for their significant contribution in the construction of the fast rotating ultrasound transducer. This work has been supported by the Technology Foundation STW (RKG.5104).

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