High-frame-rate volume imaging using sparse-random-aperture compounding

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Abstract

High-frame-rate volume imaging (HFR-VI) aims to provide high-quality images with high-temporal information. Despite its potential, HFR-VI translation into clinical applications has been challenging due to the high cost of the equipment required to drive matrix probes with a large number of elements. The goal of this study is to introduce and test sparse-random-aperture compounding (SRAC), a technique that allows use of matrix probes with an ultrasound system that has fewer channels while maintaining high frame rates.

Four scanning methods were implemented with a 256-channel system using a 4-to-1 multiplexer and a 3 MHz matrix probe with 1024 elements. These methods used three types of waves, either single-diverging waves (SDW), multiplane-diverging waves (MDW) or wide beams (WB); and were driven using one to four SRAC. All methods were also implemented in a 1024-channel multisystem. The main-lobe-to-side-lobe ratio (MLSLR) and the contrast ratio (CR) were studied using a string phantom and a CIRS phantom, respectively.

The results showed an increase in the MLSLR and CR as a function of the number of SRAC. The multisystem provided the best results for the MLSLR. However, four SRAC outperformed the multisystem with respect to CR. The method using SDW provided the highest frame rates (i.e. 1875 and 7500 Hz for four and one SRAC, respectively), however it provided the lowest image quality. The two methods using MDWs showed a good compromise between image quality and frame rate (i.e. 187 to 750 Hz for four and one SRAC). WB provided the best image quality at the expense of frame rate (i.e. 18 to 75 Hz for four and one SRAC).

Our results suggest that SRAC in combination with the tested scanning methods can provide a low-channel count alternative for HFR-VI systems and allows a tunable tradeoff between image quality and frame rate guided by the desired application.

1. Introduction

In recent years, high-frame-rate volume imaging has gained a lot of interest because of its potential for studying complex, high velocity motion phenomena occurring in three dimensions (Fenster et al 2001, Huang and Zeng 2017). This technique allows correlation of phenomena in time and space, promoting the evolution of former 2D applications, such as functional imaging or elastography, into 3D applications (Provost et al 2014, Gesnik et al 2017, Correia et al 2018, Rau et al 2018, Papadacci et al 2019). This evolution was possible because of the development of plane-wave imaging, which led to a major paradigm shift in high frame rate ultrasound imaging in the last two decades. The development of plane-wave imaging was enabled by the ability to receive, store and reconstruct data from hundreds of transducer elements simultaneously (Tanter and Fink 2014). This technique was first described by Sandrin et al in 1999 and later further developed by Tanter et al in 2002 (1999, 2002). These studies demonstrated that thousands of frames per second (e.g. ~10,000 images/s) of ultrasound data can be acquired. Despite the frame rate advantages of plane-wave imaging, its main drawback is poor image quality due to the lack of transmit focusing leading to...
a wide point-spread function. However, in 2009 Montaldo et al introduced coherent angle compounding to increase the image quality of plane-wave imaging (Montaldo et al 2009). This breakthrough in frame rate and image quality allowed the study of mechanical properties of soft tissues using shear waves (Sandrin et al 2002, Bercoff et al 2004). Furthermore, in 2011 Macé et al demonstrated the use of high-frame-rate Doppler ultrasound imaging for measuring brain activity (i.e. functional imaging) (Macé et al 2011).

Plane-wave imaging and angle compounding were implemented by Provost et al (2014) for volume imaging in 2014. The authors used a 1024-channel ultrasound system to drive a 1024-element matrix probe with different scanning methods. In their study, they used diverging waves with different virtual sources and plane waves to create slanted wavefronts in the X-Z and Y-Z planes. They evaluated the contrast improvement and lateral resolution as a function of the number of virtual sources. In their work Provost et al, demonstrated the use of ultrafast volume imaging in elastography and Doppler applications. Later, in 2018 Correia et al presented a study on shear wave characterization in 3D (Correia et al 2018). Recently in 2019 Rabut et al implemented 4D functional ultrasound imaging of whole-brain activity in rodents (Rabut et al 2019). The authors studied the brain activity and connectivity during multiple sensory stimuli as well as during epileptic events. Also in 2019, Papadacci et al, implemented a sequence to simultaneously acquire 3D tissue and blood flow Doppler imaging for cardiac applications (Papadacci et al 2019). All these authors used a system similar to the one used by Provost et al with 1024 channels to drive either a 3 MHz or an 8 MHz matrix probe with 1024 elements.

Despite the importance of the previously mentioned studies, the implementation of these volume imaging applications requires ultrasound systems with a very large number of channels (i.e. 1024 or more). The need for these highly specialized systems makes high frame rate volume imaging expensive and difficult to implement. Several techniques to decrease the number of channels required to drive matrix probes have been proposed by multiple authors in the past. In 1994, Davidsen et al studied different random aperture designs in order to decrease the number of elements that are necessary on an ultrasound probe (Davidsen et al 1994). Their results determined that random sparse arrays had similar pulse-echo main-lobe dimensions compared to the fully populated array. Later, in 1996, Goss et al published a similar study in which they compared beam plots for 2D arrays when focusing on-axis and off-axis (Goss et al 1996). Their study showed that fully populated arrays and sparse-random arrays had similar main-lobe to grating-lobe ratios for on-axis focusing. However, the sparse-random apertures outperformed the fully populated arrays in cases for off-axis focusing (steering).

In 2002, Austeng and Holm studied several configurations of apertures in simulation and their effect on the amplitude of grating-lobes (Austeng and Holm 2002). In their study they developed symmetric and non-symmetric apertures in order to suppress grating lobes. Their apertures were based on the concept of sparse periodic layouts. The results showed that depending on the patterns, their apertures could present stronger grating lobes (i.e. similar to Vernier arrays) (Brunke and Lockwood 1997) or have higher suppression of the lobes as for almost-dense arrays.

A more recent study by Roux et al (2018) used an ultrasound system with 1024 channels to drive a probe with 1024 elements with different configuration of random apertures. In their study they used the full aperture of the probe, a random selection of 256 elements, or an optimized random selection of 256 elements. Their study concluded that for diverging waves, random apertures are a suitable approach for volume imaging with fewer elements. The lateral resolution was better for the random distribution compared to the other two cases. However, contrast ratio (CR) was better in the case of the optimized random distribution (i.e. by −4.0 dB) when compared to the full aperture.

Another approach to volume imaging includes the use of a multiplexer in which several elements from a matrix probe are connected to a single channel of the ultrasound system. Recently, Yu et al (2019) implemented a sequence using a 4-to-1 multiplexer that allowed them to control a probe with 1024 elements using a system with 256 channels. In this case, every transmitted wave needed to be repeated 16 times in order to transmit and receive with all combinations of the sub-apertures. Their scanning protocol consisted of 25 diverging waves repeated 16 times. The complete sequence accounted for 400 transmits and receives per frame, which resulted in a maximum frame rate of 30 volumes per second.

The goal of this study is to introduce and evaluate a methodology that we term sparse-random-aperture compounding (SRAC) for high-frame-rate volume imaging using a relatively low-channel-count ultrasound system. The SRAC concept uses multiple complementary random apertures to approximate the full aperture of a matrix probe. We evaluated the effect on image quality between compounding of one, two, three or four apertures with a 1024-element probe and a 256-channel Vantage ultrasound system (Verasonics, Inc. Kirkland, WA). This approach permits the user to select a compromise between image quality and frame rate that is suitable for their application. To evaluate the improvements in image quality, we implemented this technique in four different scanning methods using diverging, multiplane waves, and focused waves, combining some aspects of the studies by Roux and Yu. To quantify the effect of different number of SRAC,
for each of the scanning methods we measured the main-lobe-to-side-lobe ratio (MLSLR) and the contrast ratio (CR). Furthermore, we compared the results from SRAC (i.e. using the multiplexer) with the results for the same scanning protocols implemented in a multisystem with 1024 channels using all the elements of the matrix probe in transmit and receive (i.e. full aperture). The full aperture setup is similar to the one used by Provost and Roux.

2. Methods

2.1. Ultrasound system and probe
The probe used in this study was a Vermon 3 MHz matrix probe (Vermon S.A., Tours, France). This transducer consists of 1024 elements that are organized in a $32 \times 32$ pattern. The probe has four sub-apertures, each arranged in a $32 \times 8$ distribution (i.e. 256 elements). Figure 1(E) shows the four sub-apertures with a gap row in the y direction between them. The probe has a $300 \mu m$ pitch, therefore, the aperture is $9.6 \text{ mm} \times 10.5 \text{ mm}$ because of the missing rows.

The probe was connected to a Vantage 256 system (Verasonics Inc. Kirkland, WA, USA) using a 4-to-1 multiplexer (1024-MUX UTA from Verasonics Inc. Kirkland, WA, USA). The 1024-MUX UTA has 256 mux switches. Each switch has four states that connects a group of four elements in the probe (i.e. one on each sub-aperture) to one channel of the ultrasound system. For example, the elements 1, 257, 513, and 769 of the probe can be connected to channel 1 of the system. A set of four random apertures was created by taking the MUX-switch constraints into account.

To determine the element distribution for a random aperture, we first selected a random number between 1 and 1024. If that element number or an element connected to the same switch was already chosen, a new element number was randomly drawn until a total of 256 elements were assigned to a particular aperture. This procedure was continued until all four random apertures were determined, creating one set of complementary apertures. The four random apertures were mutually exclusive, which means that no element was repeated among the four random apertures. Figures 1(A)–(D) shows the four random apertures, while figure 1(E) shows the superposition of all four apertures, which recovers the full aperture.

One hundred sets of these random apertures were created (i.e. a total of 400 apertures). Each set was used for a transmit and receive event. For those scripts that had multiple transmits (i.e. for 10-angle compounding), 10 sets of random apertures were used (i.e. one set per angle). Changing apertures between transmit events and between transmit and receive actions was possible since switching of the multiplexer takes approximately $2 \mu s$.

A multisystem consisting of four synchronized Vantage-256 systems with a total of 1024 channels was used to drive the full aperture of the probe and compare with the performance of SRAC. The same scanning methods that were developed for the multiplexer (i.e. the SRAC approach) were then modified to use the full aperture in transmit and receive. However, the voltage used to drive the multisystem was $1/4$ of the voltage used for the Vantage 256, because the multisystem uses four times more elements for each transmit. This was done in order to normalize the transmits and to better compare between the two acquisition systems.
Figure 2. Beam patterns of the four scanning methods. (A) SDW; (B) Cross; (C) Spl; and (D) WB. The representations of scanning patterns are shown at 150 \( \lambda \), while the virtual source for the diverging and focused beam are shown at \(-32\) and \(-100\ \lambda\), respectively. The black square at 0 \( \lambda \) represents the matrix probe.

In order to study the performance of high-frame-rate SRAC, we designed four scanning methods: single diverging wave (SDW), multiangle in two directions (Cross), multiangle spiral (Spl), and wide beam (WB). The SDW consisted of transmitting one diverging wave with a virtual source behind the probe at \(-32\) wavelengths (\( \lambda \), based on the 3 MHz center frequency of the probe and assuming a speed of sound in water of 1500 m s\(^{-1}\)) as can be appreciated in figure 2(A). The Cross method coherently combined the echoes of five diverging waves tilted between \(-20\) and 20 degrees from both the \( X \) and the \( Z \)-axis (i.e. 40 degrees scanning sector) in steps of 10° in each direction (i.e. 5 steps in \( X-Z \) and 5 steps in \( Y-Z \) direction). The position of the virtual source, placed at \(-32\ \lambda \), and the scanning pattern of the transmit beam described a cross is presented in figure 2(B). The Spl method coherently compounded signals from multiplane diverging waves originated from a virtual source at \(-32\ \lambda \). This source moved in the \( X-Y \) plane describing a spiral around the \( Z \)-axis (figure 2(C)). Ten steps were chosen for the virtual source to revolve twice around the vertical axis (i.e. in 80° increments) to describe the spiral pattern. The steering of each plane is determined by the angle between the \( Z \)-axis and the imaginary line that connects the virtual source to the center of the transducer.

For this method the maximum angle was 20° and the minimum was 0°. Finally, the WB method utilized 100 partially overlapping semi-focused beams with the origin at the probe center. In this case the virtual sources were a grid of 100 points at \(-100\ \lambda \) behind the transducer (figure 2(D)). The echo signals from multiple overlapping beams at a reconstruction point were combined coherently to produce the image point.

Backscattered signals were received and stored using the complementary random apertures. In this case the scripts allow selection of the number of sparse-random apertures that are used for compounding. For example, to compound three random apertures in the Cross method (i.e. 10 angles), would require 30 transmits (i.e. 10 angles x 3 random apertures). Each of the angles uses a set of random apertures for transmitting and receiving (i.e. 10 sets total). The transmit for each angle is repeated three times using the same random aperture of each set (i.e. the first random aperture). However, the receive events are performed with the first three random apertures of the set. Thus, the first event transmits and receives using the first random aperture. However, the second event transmits with the first random aperture of the set and receives with the second random aperture of the set. Lastly, the third event receives with the third random aperture. This way, when using a 3 SRAC, the fourth random aperture of each set is not used. The same sequence is
Figure 3. String and CIRS 054 GS phantom setups for the study of the MLSLR (A) and CR (B). (A) represents the setup for the on-axis MLSLR of the X-Z plane.

repeated for the subsequent angles using the corresponding set of random apertures. After the transmit and receive sequence, the RF data of each angle and aperture compounding were used to reconstruct a single frame using the Verasonics pixel-oriented sum and delay algorithm (Daigle 2013) at a voxel resolution of 1 λ.

The multisystem used the full aperture for both the transmit and the receive (i.e. all elements of the 1024 matrix probe). In this case no synthetic-aperture compounding was performed.

2.2. Experiments

In order to evaluate the performance of the different SRAC approaches used for each of the scanning methods, we evaluated the main-lobe-to-side-lobe ratio (MLSLR) as well as the contrast ratio (CR). The MLSLR was tested using a phantom that consisted of one string at a depth of 120 λ (i.e. ~60 mm at an assumed 1500 m s⁻¹ speed of sound in water). To test the on-axis MLSLR of the X-Z plane, the string was first aligned with the center of the probe parallel to the y-axis. The setup for this experiment is shown in figure 3(A). The phantom was scanned using the four scripts (i.e. SDW, Spl, Cross, WB) in combination with one, two, three, and four SRAC, as well as with the multisystem using the four scripts. This resulted in a total of 20 image reconstructions (i.e. 16 for the Vantage-256 system, and 4 for the multisystem).

The string phantom was then rotated 90° to evaluate the on-axis MLSLR of the Y-Z plane. In this orientation the string was placed parallel to the x axis and again aligned with the center of the probe. Twenty additional image reconstructions were conducted in this configuration.

To test the off-axis MLSLR, the procedure was repeated by aligning the phantom with the y or x axis (i.e. for the X-Z and Y-Z plane, respectively) and translating the string 20 λ away for the center of probe. A total of 80 image reconstructions were taken using the string phantom for the on-axis and off-axis MLSLR in the X-Z and Y-Z planes.

To evaluate the CR of the different scripts and the impact of using multiple SRAC on the image contrast, a CIRS-054GS (CIRS Inc. Norfolk, VA, USA) phantom with a speed of sound of 1540 m s⁻¹ was used. The probe was placed over an anechoic inclusion at 95 λ from the probe surface (i.e. ~49 mm assuming a speed of sound of 1540 m s⁻¹) as shown in figure 3(B). The phantom was oriented in a way that the B-mode image of X-Z plane showed the circular cross-section of the anechoic inclusion, while the Y-Z plane showed the rectangular cross-section of the anechoic cylindrical inclusion. Similar to the string phantom, 20 image reconstructions were performed for the four scanning methods together with the one to four random apertures as well as the multisystem.

2.3. Data analysis

To analyze the results of the string phantom and to calculate the MLSLR, we first assessed the B-mode image at the center of the volume in the X-Z and Y-Z planes to determine the exact location of the string (figures
Figure 4. String phantom acquisitions for the MLSLR analysis. (A) and (D) are the B-mode images of the X-Z and Y-Z planes for the on-axis MLSLR. (G) and (J), the corresponding B-mode images for the off-axis MLSLR. The red box in (A), (D), (G) and (J), represent the depth range that is averaged over the volume. (B), (E), (H) and (K) show the corresponding X-Y plane after the averaging. (C), (F), (I) and (L) are the line profiles of the main-lobe on the centerline of the probe for the X-Z, Y-Z on the on-axis and off-axis setups, respectively. The black boxes on this last panels show the region used to average the background noise for the MLSLR calculation.

4(A) and (D), respectively). Second, the volume was averaged in depth (i.e. z-axis) around the echoes from the string (i.e. 10λ above and 5λ below) in order to include the sidelobes. The red box in figures 4(A) and (D) shows the region of interest (ROI) used for averaging. The average signal was then displayed as an X-Y image (figures 4(B) and (E)). Lastly, the profile of the main lobe was analyzed at the five closest lines next to the centerline of the probe, which are depicted as the blue lines in figures 4(B) and (E). The amplitude profiles were normalized and plotted in dB scale to determine the on-axis MLSLR for the X-Z and Y-Z planes, respectively. Figures 4(C) and (F) shows the line profile at the centerline of the probe (the other four lines are not shown to avoid cluttering the display). The MLSLR was calculated as the difference between the amplitude of the main lobe (in dB) and the average background value, which was calculated using the regions showed in figures 4(C) and (F) (i.e. the black box around the amplitude trace). The values for MLSLR reported in this study, are the mean values and standard deviation from the two planes (X-Y and Y-Z) and from the different line profiles for each plane.

The same analysis was repeated to calculate the off-axis MLSLR with the string phantom placed at an offset from the probe center (figures 4(G)–(L)). The results from this analysis are shown in tables 1 and 2 for the on-axis and off-axis, respectively.

The CR analysis was done as shown in figure 5 for the X-Z and Y-Z planes at the center of the probe as depicted in figures 5(A) and (B), respectively. Four ROIs were selected for the X-Z plane with one inside the anechoic inclusion and three in the background of the phantom (figure 5(A)). For the Y-Z plane, three
Table 1. String phantom results of the MLSLR for the on-axis setup for the different scanning methods (SRAC and multisystem). Values represent the mean and standard deviation.

| Scan method | 1 RndApe | 2 RndApe | 3 RndApe | 4 RndApe | Multisys |
|-------------|----------|----------|----------|----------|----------|
| SDW         | −12.3 ± 0.2 | −14.6 ± 0.5 | −15.9 ± 1.0 | −17.4 ± 1.2 | −19.4 ± 1.4 |
| Cross       | −16.8 ± 0.5 | −19.9 ± 0.7 | −21.3 ± 0.9 | −22.6 ± 1.3 | −23.2 ± 1.3 |
| Spl         | −17.1 ± 0.2 | −20.1 ± 0.4 | −22.2 ± 0.4 | −23.9 ± 0.6 | −26.0 ± 0.8 |
| WB          | −21.8 ± 0.3 | −24.7 ± 0.4 | −26.0 ± 0.5 | −27.1 ± 0.7 | −28.5 ± 1.1 |

Table 2. String phantom results of the MLSLR for the off-axis setup for the different scanning methods (SRAC and multisystem). Values represent the mean and standard deviation.

| Scan method | 1 RndApe | 2 RndApe | 3 RndApe | 4 RndApe | Multisys |
|-------------|----------|----------|----------|----------|----------|
| SDW         | −9.4 ± 0.2 | −12.1 ± 0.4 | −14.0 ± 0.7 | −16.2 ± 0.9 | −18.1 ± 0.9 |
| Cross       | −15.3 ± 0.3 | −18.4 ± 0.6 | −20.3 ± 0.8 | −21.2 ± 0.8 | −21.9 ± 0.9 |
| Spl         | −16.7 ± 0.2 | −19.9 ± 0.4 | −21.7 ± 0.5 | −23.4 ± 0.6 | −23.4 ± 0.7 |
| WB          | −22.2 ± 0.6 | −24.4 ± 0.7 | −26.1 ± 0.8 | −27.1 ± 0.8 | −28.0 ± 1.1 |

Figure 5. CIRS 054 GS phantom imaging for the CR analysis. (A) shows the X-Z plane; four ROIs are marked, one on the anechoic inclusion and three on the background scattering media. (B) is the image on the Y-Z plane. Three ROIs are marked, one on the inclusion and two in the scattering media. The inclusion of the CIRS phantom is an anechoic cylinder.

rectangular ROIs were placed in the B-mode image, one inside the anechoic inclusion, and two in the background with one above and one below the inclusion as shown in figure 5(B). The CR was calculated using equation (1), as it was described by Roux et al (2018).

\[
CR = 20\log_{10}\left(\frac{\mu_{in}}{\mu_{out}}\right)
\]  

\(\mu_{in}\) and \(\mu_{out}\) in equation (1) correspond to the mean signal amplitude of the ROIs inside and outside of the anechoic inclusion, respectively. The values of \(\mu_{in}\) and \(\mu_{out}\) were taken before any compression was applied to the signal.

Similar to the MLSLR analysis, five planes closest to the centerline of the probe were used to calculate the CR at each orientation. Therefore, in the case of the X-Z direction, 15 values of CR were calculated, three for each of the five planes from the three circular inclusions. In the Y-Z direction, 10 CR values were obtained, two from each of the five planes from the rectangular ROIs. The values for CR from the two orientations and the different ROIs were then averaged in order to report a mean value of CR and the standard deviation.

Table 3 shows the results for the different scanning methods using SRAC with the Vantage 256 as well as the multisystem.
Table 3. CIRS phantom results for the CR for the different scanning methods (SRAC and multisystem). Values represent the mean and standard deviation.

| Scan method | 1 RndApe | 2 RndApe | 3 RndApe | 4 RndApe | Multisys |
|-------------|---------|---------|---------|---------|---------|
| SDW         | −3.3 ± 0.5 | −4.0 ± 0.4 | −5.9 ± 0.7 | −7.3 ± 0.7 | −5.7 ± 0.5 |
| Cross       | −5.1 ± 0.4 | −6.9 ± 0.8 | −8.2 ± 0.5 | −9.2 ± 0.5 | −9.0 ± 0.4 |
| Spl         | −5.3 ± 0.6 | −7.9 ± 0.4 | −8.2 ± 0.3 | −9.7 ± 0.4 | −9.9 ± 0.4 |
| WB          | −10.8 ± 0.5 | −13.2 ± 0.5 | −14.4 ± 0.4 | −15.7 ± 0.7 | −14.8 ± 0.5 |

3. Results

Figure 4 shows the images from the string phantom with the Spl script using 4 SRAC. Panels (A) and (D) display the on-axis configuration with the string parallel to the y axis for both imaging planes (X-Z and Y-Z planes, respectively). Therefore, in Panel (A) (i.e. X-Z plane) the string is visualized as a dot in the center of the probe, while in Panel (D) (i.e. Y-Z plane), the string can be appreciated as being parallel to the y axis. Panels (B) and (E) show the X-Y plane images after averaging the volume over a depth of 15 λ around the string. These panels show the two orientations of the string parallel to the y and x axis at the center of the probe, respectively. Panels (C) and (F) show the corresponding amplitude profiles (in dB) for the center line in both directions. For simplicity only the trace in the center of the probe is displayed. Similarly, Panels (G)–(L) show the corresponding data for the off-axis setup. Panels (H) and (K) show the amplitude profiles for the off-axis acquisitions, where the 20 λ offset of the string from the probe center can be appreciated.

Figure 6 depicts the amplitude profiles at centerline for the Spl method in the on-axis setup. This figure shows the differences in the profiles when compounding different numbers of random apertures with increasing MLSLR from one to four SRAC and with the multisystem. Panel (A) shows the results for the centerline trace in the X-Z direction (data of the other 4 profiles are not shown) and Panel (B) shows the results in the Y-Z plane.

The values of MLSLR for all scanning methods and SRAC are shown in figure 7. In Panel (A) are the results for the MLSLR of the on-axis setup, while in Panel (B) are the results for the off-axis setup. The error bars in this bar graph represent the standard deviation from the five centerlines used for the calculation of the MLSLR as well as the difference from the X-Z and Y-Z planes. The data are also summarized in tables 1 and 2.

An acquisition performed on the CIRS phantom is shown in figure 5. This image was generated using the Spl script and four SRAC. Panels (A) and (B) show the images for the X-Z and Y-Z, respectively. An anechoic inclusion can be observed in panel (A) in the center of the field at a depth of 95 λ. Since the inclusion is an anechoic cylinder, the Y-Z plane displays the inclusion as a rectangle. The results of the CR are shown in figure 8 and table 3 for the different scanning methods. Similar to the results for the MLSLR, the values represent the mean and standard deviation from the different ROIs, centerlines (i.e. consecutive planes in the X-Z and Y-Z directions), and imaging planes (i.e. X-Z and Y-Z planes).
Figure 7. MLSLR results for all the scanning methods with one-four SRAC and multisystem. The results from the on-axis setup are shown in (A), while the off-axis results are shown in (B). The values presented are the mean and standard deviation from the $X$-$Z$ and $Y$-$Z$ measurement in each case and from amplitude profiles from the five lines.

Figure 8. CR results for all the scanning methods using 1 to 4 SRAC and multisystem. The values represent the mean and standard deviation from the ROIs analyzed in the $X$-$Z$ and $Y$-$Z$ plane as well from the five planes in the center of the probe in each direction.

Figure 9 presents the relationship between the measured MLSLR and CR along with the achievable frame rate given the number of transmits and receives per scanning method. The values of frame rate are theoretical and calculated taking into account a 1540 m s$^{-1}$ speed of sound and a depth of 10 cm. In this figure, each of the scanning methods is represented by a symbol (i.e. ’+’ for SDW, ’x’ for Cross, ’□’ for Spl and ’*’ for WB). The different colors represent the number of SRAC used in each case. The ‘black’ symbols represent the results for each particular scanning method using the multisystem. Each scanning method is connected by a solid ‘black’ line, showing the progression from one to four SRAC. Lastly, it is important to note that the multisystem acquisition and one SRAC have the same frame rate, because only one transmit is required per frame for both methods.
4. Discussion

The goal of this study was to introduce a new volumetric ultrasound imaging approach that uses sparse random aperture compounding to lower system cost while maintaining image quality and frame rate. Although random apertures were used previously to address matrix probes with ultrasound systems that have a limited number of channels (Goss et al 1996, Roux et al 2018), to the best knowledge of the authors, this is the first time that compounding of data from different random apertures was performed. In this work we termed this new concept sparse-random-aperture compounding (SRAC) and demonstrated an improvement in image quality while maintaining high frame rates in an ultrasound system that possesses significantly fewer channels than probe elements.

The data presented in figure 6 demonstrates the improvement in MLSLR for the Spl method as the number of random apertures are compounded. When four apertures are compounded, the MLSLR is comparable to that of the full aperture acquisition (i.e. full aperture using a multisystem). The amplitude profiles for the main lobe for the other scanning methods were very similar (data not shown).

The average improvement in MLSLR between one and two random apertures for all the scanning methods in the on-axis setup was $2.8 \pm 0.4$ dB. The improvement between using two and three apertures was $1.5 \pm 0.3$; between three and four was $1.4 \pm 0.2$ dB; and the improvement between using 4 random apertures and the full aperture with the multisystem was $1.5 \pm 0.7$ dB. For the off-axis MLSLR the values were $2.8 \pm 0.5$ dB, $1.8 \pm 0.1$ dB, $1.4 \pm 0.6$ dB and $0.91 \pm 0.8$ dB, respectively.

Important to note is the difference of the MLSLR between the $X-Z$ and $Y-Z$ plane as observed in figure 6. The $Y-Z$ plane consistently showed a lower MLSLR relative to the $X-Z$ plane for all scanning methods. This is presumed to be a result of the missing rows of the probe in the $y$ axis. These missing elements increase the size of the side lobes, therefore, decreasing the performance of the probe in this direction. The average difference for the on-axis setup between the $X-Z$ and $Y-Z$ planes for all scanning methods was $4.3 \pm 1.1$ dB. In the case of the off-axis setup the average difference between the two planes was $5.0 \pm 1.4$ dB. Therefore, using an orientation that favors the $X-Z$ plane when imaging with a probe that has missing rows is preferred.

The CR as well as the MLSLR increases as a function of the number of random apertures that are used during reconstruction. However, figure 8 provides evidence that CR performance of four SRAC is better than that obtained using the full aperture with the multisystem for most of the scanning methods (except for the Spl). This can be explained by the different clutter noise in the acquisition of each random aperture, which is reduced because of averaging the multiple random apertures. Conversely, the multisystem performs less averaging of the clutter noise because the acquisition is done using only one transmit and receive per angle. Analysis of the numbers of SRAC apertures and their impact on the CR showed that when going from one to two SRAC, the improvement was $1.9 \pm 0.9$ dB. The increase in CR was $1.2 \pm 0.6$ dB from two to three SRAC; and $1.3 \pm 0.2$ dB from three to four. As mentioned earlier, the CR decreased slightly from four SRAC to the multisystem full aperture with an average decrease for all scanning methods of $-0.6 \pm 0.8$ dB.
A comparison of the scanning methods demonstrated that the WB was the best in terms of MLSLR as well as CR. This was expected because the WB approach combines focused beams and a high number of transmit and receive events. However, due to the increased number of acquisitions, the WB is also the scanning method with the lowest frame rate (i.e. 75 volumes per second for one SRAC and full aperture, and 18 Hz for four SRAC) as can be observed in figure 9. The Cross and Spl scanning methods demonstrated a good compromise between frame rate and image quality. However, Spl showed a slight advantage over the Cross scanning in terms of MLSLR and CR, even though they both had the same number of transmits (i.e. 10 angles) and the scanning volume was comparable with a maximum steering of 20° in both cases. This can be appreciated in figure 9, where the Spl is represented by ‘□’ and the Cross by ‘x’. For both measurements of MLSLR and CR we observed that the Spl is shifted to the right (i.e. higher MLSLR and CR) for the same frame rates in each instance of SRAC and multisystem. This advantage of the Spl over the Cross scanning is most significant in the MLSLR measurement when using four SRAC and the multisystem. In the case of the on-axis setup (table 1 and figure 9(A)) these differences were 1.3 ± 1.4 dB for the four SRAC and 2.8 ± 1.5 dB for the multisystem. For the same conditions, in the case of the off-axis setup, the advantage of the Spl was of 2.2 ± 1.0 and 1.5 ± 1.1 dB, respectively. These comparisons suggest that higher heterogeneity of multitone transmits has an advantage in terms of MLSLR for the same number of transmitted planes.

As expected, the SDW scanning method presented the lowest MLSLR and CR because this approach uses only 1 transmit without any steering. However, the strength of this method is the high frame rate. Figure 9 shows that for one SRAC and fully filled aperture (i.e. multisystem) the method allows acquisition frame rates of up to 7500 Hz at a depth of 10 cm. In the case of four SRAC apertures, which has equivalent MLSLR to the Cross and Spl with one SRAC, the SDW scanning method still provides frame rates up to 1800 Hz. Therefore, in spite of its lower MLSLR and CR, this scanning mechanism is useful in applications where very high frame rates are required, such as shear wave elastography (Bercoff et al 2004, Provost et al 2014), or functional imaging (Macé et al 2013, Rabut et al 2019).

The MLSLR from our study are in good agreement with those presented by Choe et al (2012). In their study, the authors used a ring array to image a wire phantom and measured the difference between the main lobe coming from the wire and the average signal of the side lobes that are due to background noise. Choe et al used different acquisition sequences such as flash imaging, classic phased array imaging and synthetic phased array imaging among other techniques. The values for the MLSLR in their study were between −11.7 to −27.32 dB for the optimized sequences. In our example, the values for the on-axis setup were between −12.3 dB for the SDW with one SRAC to −27.1 dB for the WB with four SRAC. Similar results were also published by Santos et al (2016) in a study where they investigated the feasibility of HFR-VI using diverging waves and sub apertures in a clinical system. They concluded that a sparse transmit of moderate number of diverging waves significantly reduces grating lobes. Their values of MLSLR varied from about −3 to −20 dB, depending on the number of waves used and the scanning method.

Comparison of our results with those published by Roux et al (2018) showed a good correlation for the measurements of CR. In their study they used a scanning method similar to the Cross method as proposed in our study. The values of CR reported by Roux et al for the full aperture, a 256-element random aperture, and an optimized 256-random aperture were −10.1 dB, −4.0 dB, and −4.8 dB, respectively. These values are in good agreement with those presented in table 3. The values for CR for the Cross-scanning method were −9.0 ± 0.4 dB for the full aperture and −5.1 ± 0.4 dB for one SRAC (i.e. a 256-random aperture). However, in their study, Roux et al found that scanning with diverging waves performed better than scanning with focused beams. This contradicts our results since we observed consistently better MLSLR and CR in the WB (i.e. beams focused below the imaging region) compared to the other scanning methods. This discrepancy could be attributed to the difference in the use of conventional focused beams vs. weakly focused beams, and the scanning parameters.

In the study by Provost et al the authors observed that contrast was related to the number of virtual sources (i.e. transmitted beams). They reported a slight increase in contrast (<10 dB) when using up to five sources. The contrast increased to 20 dB when using 16 diverging waves and they observed a plateau around 30 dB with 35 or more virtual sources (Provost et al 2014). However, these values are not directly comparable to those presented by Roux et al or in our study since the contrast measurements were done differently. Provost et al quantified contrast as the difference in the B-mode signal (i.e. compressed image) in a cavity (i.e. anechoic inclusion) and surrounding tissue in a cardiac phantom. Nevertheless, the increase in MLSLR and CR as function of the virtual sources agrees with our observations where the WB (i.e. 100 virtual sources) outperformed the other three scanning methods (i.e. one virtual source for SDW and ten sources for Cross and Spl, respectively).

Our study had a few limitations. While this study investigated the effect of using complementary random apertures for compounding, the work presented here did not focus on examining whether simple random apertures (i.e. not complementary) present the same improvement. Furthermore, the transmits for each
angle when performing SRAC were always done with the first random aperture of each set. We conducted preliminary tests modifying the random apertures of both the transmits and the receives, and the results showed a decrease in performance when compared to using a consistent aperture for the transmits (data not presented). Moreover, to our best knowledge, we do not know of any study that investigated whether varying the random aperture for each transmit would provide an improvement in image quality. Along the same lines, the study did not test more than four SRAC apertures because four complementary apertures recover the full aperture. However, as seen in the CR, the more random apertures that are used for compounding the more improvement is observed. Furthermore, since four SRAC apertures provided better CR than the full aperture of the multisystem, we do not exclude the possibility that using more than four random apertures would provide some extra gain in MLSLR and CR.

Future work will focus on studying the possibility to optimize the random apertures to see if further improvement can be achieved. Several studies have looked into the optimization of apertures for transmit and receive (both random and fixed patterns) to maximized focalization, minimize side lobes, etc (Austeng and Holm 2002, Choe et al 2010). The work by Roux et al (2018) presented an optimized random aperture that showed better performance than uniform random distribution. However, future studies have to demonstrate whether this optimization can be implemented using the 4-to-1 multiplexer given the constraints of the multiplexer topology.

5. Conclusion

In this study we introduced the concept of sparse-random-aperture compounding (SRAC), which is a technique that allows the use of matrix probes with a high element count together with ultrasound systems that have lower channel counts. We demonstrated that the use of complementary random apertures improves image quality while preserving high frame rates. Furthermore, our results showed that SRAC and the scanning methods that were investigated in this study can provide a tradeoff between image quality and frame rate, which can be tuned depending on the desired application.

Therefore, we believe that SRAC methods can provide a lower-cost low-channel count alternative to more complex and expensive systems that drive the full aperture of the transducer. Future studies should focus on optimizing the random apertures that are used for compounding in order to further increase the image quality while maintaining or improving frame rate.

Conflict of interest

MB, BC and DR are employees of Verasonics Inc. RD is a co-founder and shareholder of Verasonics Inc. a company commercializing the ultrasound scanners used in this study.

References

Austeng A and Holm S 2002 Sparse 2-D arrays for 3-D phased array imaging - design methods IEEE Trans. Ultrason. Ferroelectr. Freq. Control 49 1073–86
Bercoff J, Tanter M and Fink M 2004 Supersonic shear imaging: a new technique for soft tissue elasticity mapping IEEE Trans. Ultrason. Ferroelectr. Freq. Control 51 396–409
Brunke S S and Lockwood G R 1997 Broad-bandwidth radiation patterns of sparse two-dimensional vernier arrays IEEE Trans. Ultrason. Ferroelectr. Freq. Control 44 1101–9
Choe J W, Oralkan O and Khuri-Yakub P T 2010 Design optimization for a 2-D sparse transducer array for 3-D ultrasound imaging Proc. - IEEE Ultrasonics Symp. pp. 1928–31
Choe J W, Oralkan O, Nikoodehadeh A, Gencel M, Stephens D N, O’Donnell M, Sahin D J and Khuri-Yakub B T 2012 Volumetric real-time imaging using a CMUT ring array IEEE Trans. Ultrason. Ferroelectr. Freq. Control 59 1201–11
Correia M, Deffieux T, Chatelin S, Provost J, Tanter M and Pernot M 2018 3D elastic tensor imaging in weakly transversely isotropic soft tissues Phys. Med. Biol. 63
Daigle R 2013 Ultrasound imaging system with pixel oriented processing J. Acoust. Soc. Am. 133 2521
Davidsen R E, Jensen J A and Smith S W 1994 Two-dimensional random arrays for real time volumetric imaging Ultrason. Imaging 16 143–63
Fenster A, Downey D B and Cardinal H N 2001 Three-dimensional ultrasound imaging Phys. Med. Biol.
Gesnik M, Blaize K, Deffieux T, Gennison J L, Sahel J A, Fink M, Picaud S and Tanter M 2017 3D functional ultrasound imaging of the cerebral visual system in rodents Neuroimage 149 267–74
Goss S A, Frizzell I A, Kouzmanoff I J, Barich J M and Yang J M 1996 Sparse random ultrasound phased array for focal surgery IEEE Trans. Ultrason. Ferroelectr. Freq. Control 43 1111–21
Huang Q and Zeng Z 2017 A review on real-time 3D ultrasound imaging technology Biomed. Res. Int.
Macé E, Montaldo G, Cohen I, Baulac M, Fink M and Tanter M 2011 Functional ultrasound imaging of the brain Nat. Methods 8 662–4
Macé E, Montaldo G, Osmanski B-F, Cohen I, Fink M and Tanter M 2013 Functional ultrasound imaging of the brain: theory and basic principles IEEE Trans. Ultrason. Ferroelectr. Freq. Control 60 492–506
Montaldo G, Tanter M, Bercoff J, Benech N and Fink M 2009 Coherent plane-wave compounding for very high frame rate ultrasonography and transient elastography IEEE Trans. Ultrason. Ferroelectr. Freq. Control 56 489–506
Papadacci C, Finel V, Villemain O, Goudot G, Provost J, Messas E, Tanter M and Pernot M 2019 4D simultaneous tissue and blood flow Doppler imaging: revisiting cardiac Doppler index with single heart beat 4D ultrafast echocardiography Phys. Med. Biol. 64

Provost J, Papadacci C, Arango J E, Imbault M, Fink M, Gennisson J L, Tanter M and Pernot M 2014 3D ultrafast ultrasound imaging in vivo Phys. Med. Biol. 59 L1–L13

Rabut C, Correia M, Finel V, Pezet S, Pernot M, Deffieux T and Tanter M 2019 4D functional ultrasound imaging of whole-brain activity in rodents Nat. Methods 16 994–7

Rau R, Kruizinga P, Mastik F, Belau M, de Jong N, Bosch J G, Scheffer W and Maret G 2018 3D functional ultrasound imaging of pigeons Neuroimage 183 469–77

Roux E, Varray F, Petrusca L, Cachard C, Tortoli P and Liebgott H 2018 Experimental 3-D ultrasound imaging with 2-D sparse arrays using focused and diverging waves Sci. Rep. 8

Sandrin L, Catheline S, Tanter M, Henniequin X and Fink M 1999 Time-resolved pulsed elastography with ultrafast ultrasonic imaging Ultrason. Imaging 21 259–72

Sandrin L, Tanter M, Catheline S and Fink M 2002 Shear modulus imaging with 2-D transient elastography IEEE Trans. Ultrason. Ferroelectr. Freq. Control 49 426–35

Santos P, Haugen G U, Lovstakken L, Samset E and D’Hooge J 2016 Diverging wave volumetric imaging using subaperture beamforming IEEE Trans. Ultrason. Ferroelectr. Freq. Control 63 2114–24

Tanter M, Bercot J, Sandrin L and Fink M 2002 Ultrafast compound imaging for 2-D motion vector estimation: application to transient elastography IEEE Trans. Ultrason. Ferroelectr. Freq. Control 49 1363–74

Tanter M and Fink M 2014 Ultrafast imaging in biomedical ultrasound IEEE Trans. Ultrason. Ferroelectr. Freq. Control 61 102–19

Yu J, Yoon H, Khalifa Y M and Emelianov S Y 2019 Design of a volumetric imaging sequence using a vantage-256 ultrasound research platform multiplexed with a 1024-element fully-sampled matrix array IEEE Trans. Ultrason. Ferroelectr. Freq. Control