# Encoded multiple simultaneous aperture acquisition for improved signal-to-noise ratio in ultrasound imaging

Hany Hussein[a], Abou-Bakr M. Youssef[a] and Yasser M. Kadah[a,b1]

[a]Biomedical Engineering Department, Cairo University, Egypt
[b]Emory University/Georgia Tech Biomedical Engineering, Atlanta 30322

## ABSTRACT

Here, we propose a new technique that allows the improvement of SNR without lowering the frame rate. The basic version of the proposed technique works by exciting multiple non-overlapping apertures at the same time. Assuming N apertures are to be acquired simultaneously, N acquisitions from such arrangement are made with different aperture coding in each acquisition. In particular, the excitation to each aperture is multiplied by a different predetermined factor for each acquisition. By properly selecting these factors, the problem of estimating the different lines become equivalent to solving a well-conditioned linear system of size N. Also, we show that such factors can be chosen to make the system matrix orthogonal, thus enabling simple solution to be obtained. We describe the implementation of such procedure and demonstrate that it is possible to achieve in practice even for systems in which the sign of excitation can only be changed. For more advanced system where coded excitations can be used, more accurate implementation taking into account element sensitivity variations within each probe for closer adherence to the desired characteristics. The proposed system results in noise reduction that is equivalent to averaging N acquisitions similar to phase encoding in magnetic resonance imaging. At the same time, the system maintains the same lateral resolution and frame rate as conventional acquisition strategies. We also discuss the use of sub-optimal overlapped apertures and describe the deterioration of SNR gain as a result of using independent rather than orthogonal apertures.

**Keywords:** ultrasound imaging, beamforming, coded excitation, Hadamard coding.

## 1. INTRODUCTION

The recent advances in ultrasound imaging technology has shown significant enhancement to image quality as well as versatility of the technique[1]. Among the exciting new techniques introduced into ultrasound is the use of coded excitation. This technique has root in radar imaging where the problem of range-Doppler ambiguity is addressed. The use of coded excitation in this problem allowed longer excitation that provide better Doppler velocity estimation accuracy while maintaining the ability to “decode” the incoming echoes to maintain a high spatial resolution. This problem bears a great deal of similarity to the problem of flow mapping in ultrasound imaging. Hence, the use of coded-excitation was extended in a similar fashion to improve the efficiency of flow mapping in ultrasound imaging.

Coded excitation works by sending a pulse or a series of excitation pulses of a particular shape that may extend in time far beyond that of a conventional pulse. Then, a “decoding” filter is utilized on the receiving end of the system to convert the impulse response of the imaging system to a desired shape. This is achieved using a matched filter or an array of matched filters depending on the application. This is usually performed on the digital side of ultrasound systems for increased flexibility. Among the applications of coded excitation in ultrasound is in basic B-mode imaging where it is used to improve the signal-to-noise ratio (SNR) of the imaging process to allow further tissue penetration depth. The signal in ultrasound imaging is a direct function of the radiofrequency (RF) excitation pulse amplitude[2]. Given the limits on the excitation amplitudes in any given ultrasound system, conventional excitation pulses see to have an upper limit for the SNR given their amplitude range and the resolution desired. However, with the use of coded excitation, it is possible to significantly enhance the SNR under the same constraints by allowing shaped pulses to be longer in duration thus increasing the energy of the pulse while maintaining the axial resolution after decoding.

---

[1] E-mail: ymk@ieee.org

Several authors have addressed the problem of spatial and temporal encoding in ultrasound imaging. Authors looked problem of temporal encoding and stressed the importance of uniform illumination to be able to reach good point spread function (PSF) with receive filters[2,3]. Coded waveforms are used and characteristics of a desirable coded excitation are summarized and the authors suggested that pseudo-random binary codes are more practical by illustrating results of PSF using different codes. The use of spatial encoding was first introduced by Chiao *et al*. where the basic idea was presented with applications to phased array imaging. The authors showed an improvement of the signal-to-noise ratio (SNR) by multiple excitations elements with spatial encoding[5,6]. Authors also investigated the impact on color flow mapping (CFM) through the use of recursive ultrasound imaging technique in CFM where a new frame is created every pulse to enhance the accuracy of CFM velocity estimation[7-9]. Multiple transmit (TX) lines are obtained at the same time using Hadamard encoding. The use of two chirp excitations for temporal coded excitation along with Hadamard spatial encoding was also proposed. Combine aperture encoding for STA and temporal encoding to achieve higher SNR that comes close to TX with all elements. (temporal improvement and spatial improvement at the same time). The proposed methods were geared towards the problem of phase array imaging and color flow mapping. No provision of how such methodologies can be implemented on less expensive platforms was given in the literature. This ignores a large class of systems that can benefit from this technology.

Here, we address the problem of enhancing the SNR for conventional ultrasound imaging using linear arrays instead of phased array imaging as in [5,6]. The difference lies in the applicability of the proposed technique in low-end ultrasound systems. The goal of this work is a scheme by which we can obtain images of higher SNR at the same frame rate. We propose to do that by transmitting and receiving from multiple lines at the same time. Similar to the line encoding method used in magnetic resonance imaging, we use an encoding technique that allows the decoding of individual lines after an equivalent number of coded spatial excitations are performed. The possible motion correction problem in the proposed scheme is addressed and a simple solution is presented. Given that the computational complexity of the new scheme is rather modest and that the hardware requirements for its simplest implementation are also not demanding, it has a large potential for improving the performance of low-end to mid-range ultrasound systems.

## 2. SPATIAL ENCODING METHDOLOGY

In conventional ultrasound imaging using linear array transducers, an aperture of a particular size is used to transmit and receive an ultrasound line. By moving this aperture in space by a whole or fraction of an element size, the line position can be changed to cover the field of view. When better SNR is desired, multiple acquisitions of the same image are obtained and averaged in time. The way this is performed is to use a moving average filter that averages over the last few acquired frames. The main disadvantage of that is the loss of temporal resolution since a change in the current frame will be weighted by a small number compared to the previous frames. This causes a “slow motion” type effect in the resultant real-time ultrasound imaging sequence, which is always undesirable. Here, we propose a technique similar to [10,11] that allows an averaging effect to be obtained at exactly the same frame rate.

Fig. 1 shows conventional ultrasound imaging using M transmit elements (transmit aperture), and N receive elements (receive aperture), the M transmit elements are fired simultaneously, with the received data at the receive aperture digitized and stored, and then the aperture (transmit, and received) are moved until the frame is completed. SNR can be gained if M transmit apertures are active simultaneously for each of the M firings as illustrated in Fig. 2. This can be achieved by encoding the M transmit apertures, which then allows the received dataset to be decoded into M ultrasound lines with a higher SNR than the conventional ultrasound imaging without encoding. The M transmit apertures are simultaneously transmitted in M consecutive firings, each time with a different M-element code vector applied to the transmit apertures.

Although any invertible matrix can be used as the encoding matrix there are significant benefits to choosing the Hadamard matrix as the encoding matrix[11]. The elements of the Hadamard matrix are either +1 or -1 which can be implemented easily as phase inversion in the transmit electronics. The inverse of the symmetric Hadamrad matrix is simply the scaled version of itself. In general, the decoding process for Hadamard encoding the decoding can be performed in M log(M) operations. The net effect of the Hadmard coding and decoding is to average together M lines of conventional transmits (each with noise independent from the others). The decoded dataset contains the same complete data as that obtained by transmitting on each of the M transmit apertures consecutively for each transmit except for the higher SNR.

It should be noted that it is possible to use non-orthogonal excitations in this application. An analysis of the SNR involved in independent coding methods was presented in [10] and the conclusion was that the SNR of such techniques will deteriorate as they tend to go away from orthogonality. Moreover, the noise in the resultant lines becomes correlated when the othogonality condition is not maintained. Even though these effects are less than the idea case of orthogonal encoding, the still represent better solution than acquisition of single lines. Given that the requirement of no spatial overlapping in orthogonal excitation apertures may not allow many such parallel excitations to be performed, we still see that independent excitations can be valuable to squeeze some more SNR out of the imaging system. For example, if we have a 128-element probe with 32-element aperture, the number of possible nonoverlapping simultaneous apertures will be 4 in only one line and 4 for all other lines. This amounts to a similar averaging factor, which can be too small in some applications. Averaging factors between 3 and 4 can be obtained by using overlapped aperture acquisitions of 4 or more apertures. This is possible by simply adding the codes from different apertures in the elements of overlap.

Since all applications that used coded excitation looked only at high-end systems, it is of interest to address the requirements of the proposed method to investigate the possibility of its implementation on lower class of ultrasound imaging systems such as low-end systems. In its simplest implementation using Hadamard coding, all what is needed is a methodology that enables the flipping of the sign of ultrasound excitation in different elements. This is usually very simple to do by selecting the baseline state of the elements to be the high voltage with a negative going excitation pulse rather than the usual zero baseline with positive going ultrasound pulse. It is required though that each element has its own high voltage pulser circuit that is independent of the others to enable the selection of different factors for different apertures and to vary that according to the line acquired. Once that is secured, the actual transmission coding can be performed in the digital domain by controlling the baseline and TTL transmit pulse direction. More general coding schemes can be obtained by adding the possibility of varying the high voltage in each of the pulser circuits independently. In basic systems, this is only possible for the whole image. This can be done by either allowing an intermediate stage between the TTL excitation pulse and the actual pulser circuit whereby the amplitude of the pulse is varied. In this case, the semiconductor operating characteristics can be utilized to move the operating point rather than fixing it at the cut-off/saturation mode of operation. Another important issue is how to properly delay the simultaneous apertures used in the proposed method given that such low-end systems allow only such delays to be generated for a number of elements equal to the aperture size. The solution we propose is simple in that it uses the natural way this is implemented in ultrasound systems. In particular, for an aperture size of M elements, the beamformer consists of element selection and delay distribution stages. In the basic form of element selection, elements that have the same element number module M are grouped together. For example, group 1 consists of elements 1, M+1, 2M+1, etc. Given that M consecutive elements are pulsed each time, only one element from each group is excited. In order to implement multiple simultaneous acquisitions, we allow more than one element from the same group (or all of them) to be excited simultaneously. This results in multiple nonoverlapped apertures with the same exact focusing characteristics, which represents the basic implementation of the new method. Finally, the decoding part of this method can be implemented rather easily during the image reconstruction stage as a simple FIR filtration stage given its simple form and its small size. Hence, to summarize, for the most basic operation of the proposed method, the implementation requirements are not generally demanding as far as hardware and/or processing. More complicated implementations such as the case for overlapped acquisitions or for more general coding methods other than binary coding schemes might well require more innovative solutions to be implemented on low-cost systems.

## 3. INTRA-SLICE MOTION COMPENSATION

Since multiple encoded acquisitions of the same structures must be done at different times, it is possible that the field of view undergoes some changes in between acquisitions. This results in motion artifacts because the image was acquired from acquisitions that do not correspond to the same field of view. Such motion artifacts if severe may limit the diagnostic value of the reconstructed images. Hence, a technique that can reconstruct artifact free images using multiple acquisitions is needed to enhance the robustness of the new method for clinical use. We note however that the number of such acquisitions under the practical imaging situations we encountered is very small (here for example we used 4-8 acquisitions), which may not cause a significant motion. We address this problem still for completeness of discussion.

In ultrasound imaging, a technique based on navigator echo motion estimation was previously proposed[12]. Even though it is not possible to obtain this projection information as in MRI, the acquired raw ultrasound lines can be used to obtain this projection information. That is, the correction will be *frame-based* and not *line-based* like in magnetic resonance imaging. Hence, the problem formulation in ultrasound imaging has several unique features as compared to the one in MRI. First, the navigator echo in ultrasound imaging is virtual unlike with MRI where real echo that may be different from the image data is collected. Second, the motion in ultrasound imaging almost exclusively happens in the direction of the probe elements. That is, simple lateral shift in linear array imaging and angular shift in convex array imaging. This reduces the complexity of the problem unlike with MRI where the simple translational motion happens in both image directions in addition to the possible rotational motion within the image plane. Therefore, even though the classical navigator echo approach is limited to simple shift in one dimension, this does not pose any limitations on its use in ultrasound imaging where the actual motion follows this one-dimensional motion model closely. Under the assumption of stationary field of view in between acquisitions, an object in the middle of the field of view shows a rotational motion in between frames. Given that this object and the field of view structures are assumed rigid between frames, this motion appears as a simple shift in the computed navigator echo. The angular and radial motion estimates are computed by finding the index of the maximum value in their respective correlation vectors. The final corrected image can subsequently be reconstructed given the relative motion between the two acquisitions using simple interpolation.

## 4. RESULTS

The proposed methods were applied to correct real data obtained from the web site of the Biomedical Ultrasound Laboratory, University of Michigan. Although the techniques proposed were applied to several data sets, the data set that was used to generate the results in this paper is the one under “Acuson14a”. The parameters for this data set are as follows: 128 channels, 13.8889 MSPS A/D sampling rate, 3.5 MHz transducer with 0.22mm element spacing, 2048 RF samples per line each represented in 2 bytes, and 8 averages. The data were acquired for a tissue phantom. We used the data to simulate a 32-channel beamformer on receive. The SNR from each acquisition is varied at will by adding white Gaussian noise to the acquired signal. The simultaneous acquisitions were formed by adding the acquired data from multiple apertures together. In our case, given the limited number of elements compared to the aperture size, we tested the technique in the case when 4 simultaneous apertures are acquired comprising lines 1, 33, 65, and 97 of the image. The SNR of individual acquisitions is compared to the new method using a 4×4 Hadamard encoding matrix and to that obtained from simple averaging of 4 independent acquisitions (i.e., with different newly generated white Gaussian noise). The SNR is calculated here as the ratio of the mean to the standard deviation of the acquired signal. The improvement of SNR observed over several experiments was always in good agreement with the factor of 2 predicted by the theory.

## 5. CONCLUSIONS

A new method for spatial encoding was proposed for ultrasound imaging based on conventional linear arrays. Instead of the usual sequential acquisition, the new method excites a number of transmit apertures corresponding to different lines at the same time and uses a simple coding scheme to enable the returned signals to be decoded into their respective lines after an equivalent number of excitations. Even though the proposed methodology does not offer any improvement as far as the speed of acquisition, the SNR is improved by a factor equivalent to an averaging with the number of simultaneous excitations. This enables higher SNR images to be obtained without sacrificing the acquisition frame rate. The effect of overlapped apertures was shown to be possible in theory at the expense of loss in SNR corresponding to the degree of overlap. This might still be beneficial given the small number of overlapped apertures that can usually be fitted within the commonly used number of elements in ultrasound probes. A motion correction method to compensate for motion between views is also proposed. The implementation requirements for the new method to be incorporated into ultrasound systems is also discussed and shown to be feasible. The

## ACKNOWLEDGEMENTS

We would like to thank Dr. Matt O'Donnell and Biomedical Ultrasonics Laboratory, University of Michigan, Ann Arbor, for making their data available to us and the ultrasound community in general. We acknowledge also the partial support received from International Biomedical Engineering (IBE Technologies), Giza, Egypt.

## REFERENCES

1. K.E. Thomenius, "Evolution of ultrasound beamformers," *Proc. 1996 IEEE Ultrasonics Symp.*, pp. 1615-1622, 1996.
2. S.R. Freeman, M.K. Quick, M.A. Morin, R.C. Anderson, C.S. Desilets, T.E. Linnenbrink, and M. O'Donnell, "Delta-sigma oversampled ultrasound beamformer with dynamic delays," *IEEE Trans. Ultrason., Ferroelect., and Freq. Contr.* **46**, no. 2, pp. 320-332, 1999.
3. M.K. Kiymik, I. Guler, O. Hasekioglu, and M. Karaman, "Ultrasound imaging based on multiple beamforming with coded excitation," *Signal Processing* **58**, pp. 107-113, 1997.
4. M.K. Jeong and E.S. Ebbini, "Optimal transmit/receive beam synthesis for coded-excitation pulse-echo imaging," *Proc. 2000 IEEE Ultrasonics Symp*., pp. 1219-1222, 2000.
5. R.Y. Chiao, L.J. Thomas, and S.D. Silverstein, "Sparse array imaging with spatially-encoded transmits," *Proc. 1997 Ultrasonics Symp.*, pp. 1679-1682, 1997.
6. R.Y. Chiao and L.J. Thomas, "Synthetic transmit aperture imaging using orthogonal Golay coded excitation," *Proc. 2000 IEEE Ultrasonics Symp*., 2000.
7. T.X. Misaridis and J.A. Jensen, "An effective coded excitation scheme based on a predistorted FM signal and an optimized digital filter," Proc. 1999 IEEE Ultrasonics Symp., pp. 1589-1593, 1999.
8. S. Nikolov, K. Gammelmark, and J. Jensen, "Velocity estimation using recursive ultrasound imaging and spatially encoded signals," *Proc. 2000 IEEE Ultrasonics Symp*., 2000.
9. T.X. Misaridis and J.A. Jensen, "Space-time encoding for high frame rate ultrasound imaging," *Ultrasonics* **40**, pp. 593-597, 2002.
10. Y.M. Kadah and X. Hu, "Pseudo-Fourier imaging: a technique for spatial encoding in MRI," *IEEE Trans. Medical Imaging* **16**, no. 6, pp. 893-902, Dec. 1997.
11. L. Bolinger and J.S. Leigh, "Hadamard spectroscopic imaging (HSI) for multivolume localization," *Journal of Magnetic Resonance* **80**, pp. 162-167, 1988.
12. A. Elsharkawy, A.M. Youssef, and Y.M. Kadah, "Navigator echo based in-slice motion artifact suppression in ultrasound images based on synthetic aperture," *Proc. SPIE Med. Imag. 2001*, San Diego, Feb. 2001.

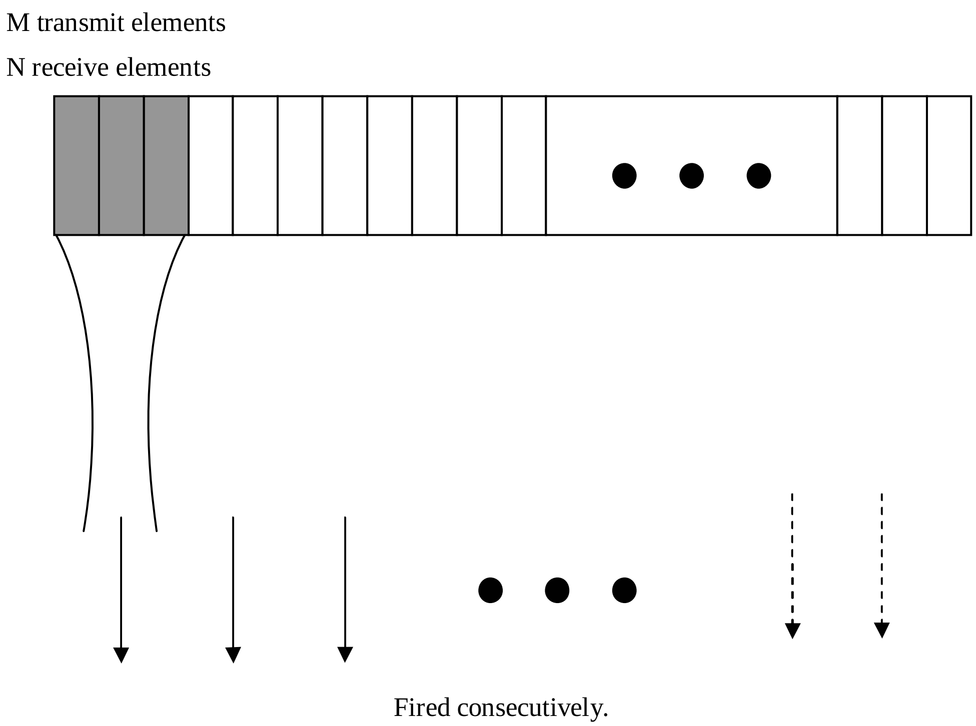


**Figure 1**. Conventional sequential acquisition scheme.

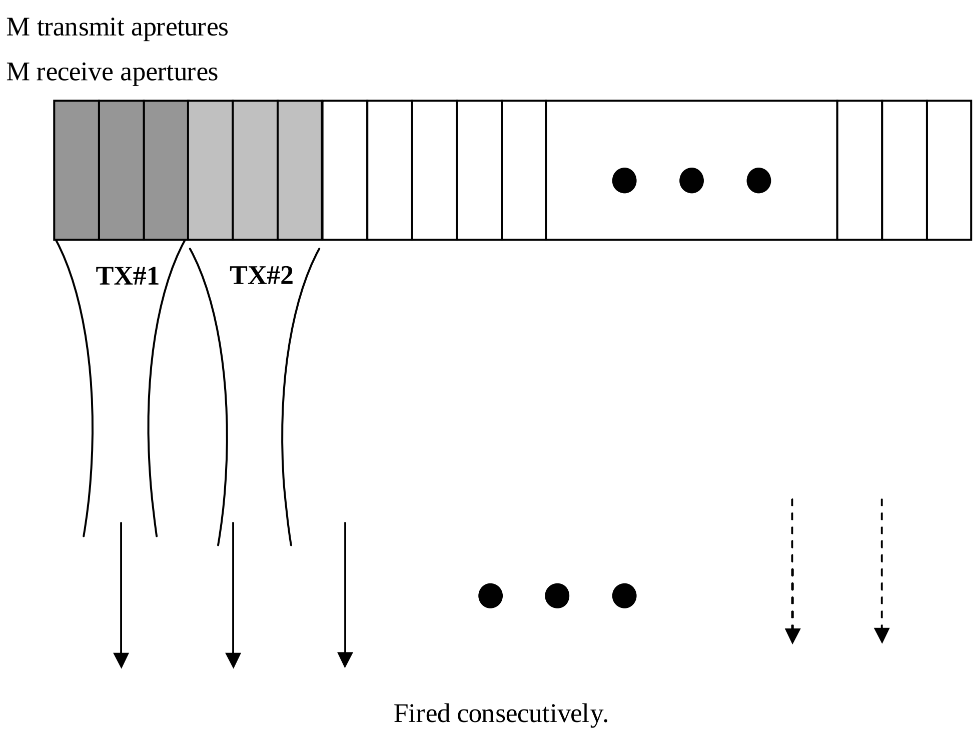


**Figure 2.** Encoded multiple simultaneous apertures acquisition.