Arbitrary Waveform Generation with the Verasonics Research Ultrasound Platform

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Arbitrary Waveform Generation

Overview

The Verasonics Vantage ultrasound system is a highly programmable platform, designed to enable research in array-based sensing, imaging, and high energy delivery technology in a wide range of technical fields. This article specifically addresses the capabilities of the Vantage transmitter, and how to use it with a transducer to produce complex analog waveforms. Because of its flexibility, many applications using various types of modulated waveforms can be explored with the Vantage multi-channel arbitrary waveform transmitter.

A brief introduction in the use of large time-bandwidth product waveforms in acoustics is presented in the context of medical imaging, an application which is constrained by safety limits on the energy entering the body. A description of the Vantage tristate pulser and transmitter hardware, and waveform software programming interface follows. Basic imaging waveforms can easily be produced, but more flexibility is achieved by the use of pulse-width modulation. It is important to recognize that when filtered by a transducer's transfer function, a pulse-width modulated trinary driving signal can be used to produce analog acoustic waveform over a dynamic range that is much larger than might be expected. Indeed, with knowledge of a transducer's impulse response, digital pulse trains can be designed to produce accurate arbitrary analog waveforms: for a 5 MHz transducer with 60% bandwidth, coded waveforms with residual RMS error of nearly -30 dB have been achieved. A Verasonics software package called the "Analog Waveform Design Toolkit" can be used to determine the necessary Vantage tristate waveforms given a desired acoustic waveform output and a transducer's impulse response. Two application examples are presented: increased depth of penetration is observed when using Golay pulse encoding, and axial resolution is improved using a drive waveform that equalizes the transducer's spectral response and effectively shortens its temporal impulse response.

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Introduction

Ultrasound imaging innovation using programmable ultrasound platforms

Medical ultrasound imaging has undergone many improvements since its initial development, and clinical systems now produce sophisticated 2D images in B-mode and Doppler, and moving 3-Dimensional volume renderings.

Many ideas for improving image quality and information content have been proposed over the years, but have languished in academic and industrial labs for lack of practical tools with which to implement new concepts.

With the advent of "open" software-based ultrasound systems that provide the developer with access to raw received data, easy programmable control of the transmit-receive sequence and beam parameters, and access to all of the real-time processing functions in the imaging chain, a new era of innovation is underway. The software-based ultrasound system has led to the exploration and development of fundamentally new approaches to ultrasound imaging including elastographic methods, new beamforming variants, and the use of coded pulses. Applications of coded transmission include extending the useful bandwidth of a transducer, extending the imaging depth in attenuating media, increasing the frame rate, and using time-reversal methods to achieve focusing gain that can greatly exceed that of conventional array beamforming. The Verasonics research ultrasound platform enables exploration and development of all of the techniques listed above. This document specifically addresses the use of the Verasonics multi-channel programmable arbitrary signal transmit capability to explore and advance the use of complex waveforms in ultrasound imaging.

Fundamental limits on image resolution

Ultrasound image quality depends on many factors, but one of the most important is spatial resolution. For narrow band signals, image resolution is typically limited to scales on the order of half an acoustic wavelength. For acoustic frequency f in a medium of sound velocity c, the wavelength λ is given by

$$\lambda = c / f.$$

For broadband signals with bandwidth BW, the best range resolution ΔR (in the direction of propagation) is

$$\Delta R = c / 2\ BW.$$

Acoustic attenuation limits the distance to which sound can propagate, and the relatively weak backscattering coefficient of tissues further reduces the echo magnitude received at the array. The acoustic pressure magnitude decreases exponentially with propagation distance, and for a plane wave, the intensity at distance d is given by

$$I(d) = I_0\ e^{-2\alpha f d}$$

where I_0 is the intensity on the plane $d=0$, and α is the attenuation coefficient, tabulated for various tissues but taken to be about 0.3 – 0.5 dB/(MHz•cm) on average. Thus, the maximum practical frequency (corresponding to the best achievable resolution) is the result of a trade off between image resolution and penetration depth. Given current commercial transducer and amplifier technology, and the limitations placed on acoustic levels by safety limits (discussed further below), rules of thumb can be developed for this trade off, when using short imaging pulses. For example, the ratio between maximum imaging depth and resolution is approximately

$$Dmax / \lambda \approx 400,$$

that is, typical acoustic imaging systems can "see" into the medium about 400 wavelengths. Because the sound speed of water is about 1500 m/s, the acoustic frequency used in medical imaging of soft tissues is on the order of

$$f \approx 600 / Dmax$$

with f in MHz, and Dmax in mm.

Conventional transmit waveforms

Conventional transmit waveforms for B-mode imaging are typically short bursts of one or two cycles. Given the physical constraints of propagation and scattering losses in soft tissues, the image Signal to Noise Ratio (SNR) is ultimately limited by safety considerations which restrict the acoustic field's instantaneous pressure magnitude and average intensity to prevent tissue damage.

Ultrasound echoes using short pulses are generally processed using delay and sum beamforming methods to reconstruct images, and offer good resolution (given the center frequency of the transducer's passband) because the transmitted pulses are only slightly longer than the transducer's own impulse response; such waveforms are said to have small time-bandwidth product.

Ultrasound bioeffects and safety limits

Safety limits are based on two types of bioeffects caused by ultrasound: mechanical disruption of tissues due to cavitation, and thermal damage from heating produced by absorption of ultrasound [1]. To facilitate regulation and standardization of diagnostic ultrasound systems, two metrics were developed by a team of clinical practitioners, researchers, regulatory and standards officials to provide real-time indicators of the acoustic regime in a particular imaging mode and for current settings.

The resulting Output Display Standard (ODS) for on-screen annotation with the current values of the two metrics is still in force today (it is in its third revision);[2]. A standard for measurement of relevant ultrasonic parameters has also been produced [3].

To prevent formation of bubbles by the field (acoustic cavitation), the instantaneous rarefactional pressure must be held below a prescribed limit, and this places a maximum cap on drive voltage for a given transmission (defined by the pulse waveform and beam geometry). The likelihood of cavitation in tissue is expressed using a non-dimensional metric called the Mechanical Index, MI, which is proportional to pulse voltage and inversely proportional to the square root of the pulse frequency; the MI is scaled such that the regulatory limit for the MI in many soft tissues is MI < 1.9. (Note that the MI is a single number for any given mode and it's current control setting, defined by the greatest value of the metric computed for every point in the field.) The maximum allowable MI thus limits the maximum imaging depth for a given acquisition mode using a particular transducer.

The amount of tissue heating resulting from ultrasound absorption is dependent on the average power carried by the acoustic field. A non-dimensional metric has also been defined to characterize the thermal impact (and risk) of a particular acquisition sequence: the Thermal Index, TI, is the ratio of the average power at a given location in the acoustic field to the power required to raise the temperature in that location by 1 °C, in steady state and based on specific tissue thermal models. Thus, the TI must always be calibrated to the particular acquisition mode (beam geometry, scan format, and PRF), transducer transmit efficiency, and tissue attenuation. The TI value for a particular imaging mode is displayed to provide guidance to the sonographer, but the safety limit imposed by the USFDA is expressed in terms of spatial peak acoustic intensity (acoustic power density in mW/cm^2), averaged over the entire imaging frame (spatial peak temporal average, I_{SPTA}) or averaged only over the pulse (spatial peak pulse average, I_{SPPA}). In practice, for scanned focused beams, the TI dependence on the beam details is rather weak, and thermal damage is in fact related to "thermal dose", a time-integrated quantity dependent on the temperature history. It is commonly accepted that raising tissue temperature a few degrees for a few minutes is safe; indeed, the WFUMB (World Federation of Ultrasound in Medicine and Biology) consensus is that indefinite exposure to ultrasound with a TI < 1.5 is safe [4].

Details regarding the parameters on which the MI and TI depend are beyond the scope of this paper (for an excellent exposition with many references, see [1]), but it is important to recognize that, as voltage is increased, most acquisition sequences reach one safety limit well before reaching the other, and this observation provides an opportunity to improve the SNR by modifying the acquisition parameters such that both limits are reached at about the same voltage. For B-mode imaging using short pulses and a scanned focused beam, the MI limit is often the dominant constraint, thus permitting the use of longer pulses (at similar voltages) until the TI limit is reached. The challenge is then to determine how best to use longer pulses to produce high resolution images while taking advantage of the increase in acoustic energy to increase imaging depth.

Large time-bandwidth waveforms and pulse compression

Instead of driving the transducer with a short pulse, a longer broadband modulated drive signal can be used, especially in medical B-mode applications that are voltage limited by the Mechanical Index (MI) and yet have plenty of margin with respect to the thermal limit (Thermal Index – TI). The modulation approach depends on the application, and on the complexity of the associated time-compression algorithms to undo the broadening of the echoes.

Much of the signal processing formalism dealing with modulated waveforms was developed in radar sensing, and in communications applications where the goal is to efficiently transmit information through a complex medium used coded pulses. Radar imaging uses various modulated methods to improve the SNR (for improved resolution or extended viewing range), or to reduce the effects of clutter. Imaging applications of rapidly moving media, often encountered in medical ultrasound imaging, use coded excitation to increase the frame rate, as will be discussed in another section.

Imaging with modulated pulses always requires some form of pulse compression, that is, additional signal processing of the echo data before high resolution images can be produced using conventional reconstruction techniques.

One-way and two-way impulse responses

When a voltage drive signal is passed through a transducer, the acoustic signal in the acoustic medium is given by a convolution between the drive signal and the transducer's (transmit) impulse response. Upon reception, the backscattered acoustic pressure signal arriving at the transducer elements is once again convolved with the transducer's (receive) transfer function and appears as a voltage on the element before amplification and digitization. This view leads to one-way and two-way impulse response definitions. In general, transducers operating in their linear range are reciprocal in the sense that the impulse response is assumed to be the same, regardless of which direction energy is flowing. However, the responses may be different if the impedance of the transmitter is different than the impedance of the receiver, which is true in the case of the Verasonics system. The one-way response can be measured using a hydrophone, but that estimation is challenging in water because acoustic propagation is nonlinear, and the wave becomes distorted because energy from the fundamental drive frequency is converted to harmonics. The degree to which nonlinearity becomes apparent depends on many factors, most importantly signal amplitude and propagation distance. The two-way response is often sufficient, because imaging applications use backscattered data, and therefore the relevant response can usually be measured directly using the system and transducer, and a well defined target and propagation geometry. If the waveform in the acoustic medium is important to the application, the 1-way transmit impulse response is needed.

Pulse coding and compression

The following figure illustrates the use of a large time-bandwidth pulse to deliver more energy per pulse, and begins with the definition of a transducer's impulse response in a linear system. The coded pulse is obtained by convolution of the code with the impulse response. Pulse compression can form an approximate estimate of a system impulse response, by selecting a wideband transmit pulse that produces suitably low autocorrelation sidelobes (Fig. 1 below). A linear statistical model approach

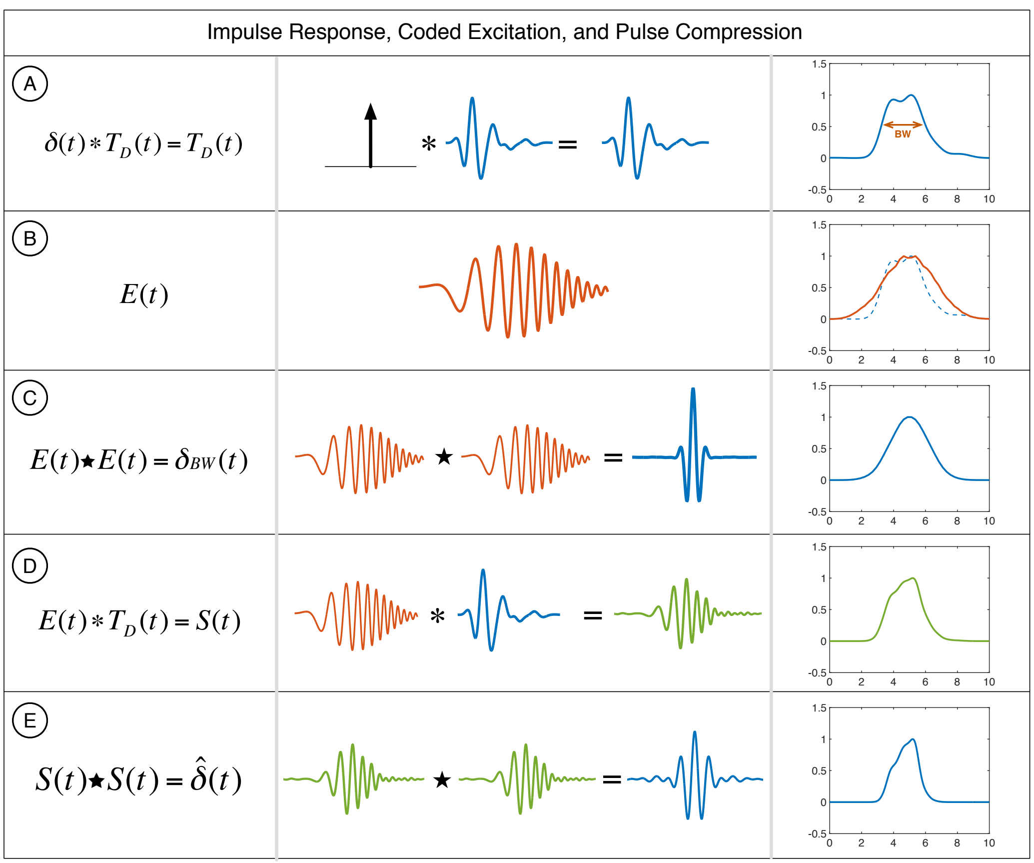


Figure 1. Impulse Response, Coded Excitation, and Pulse Compression

A. A transducer impulse response T_D is the temporal waveform resulting from an impulsive excitation; the spectrum of that response determines the transducer's bandwidth, BW. The asterisk operator $*$ represents convolution.

B. A coded excitation E is a waveform with a large time-bandwidth product, and carries much more energy in the pulse for the same signal magnitude (not illustrated because the curves are normalized). Typically, the bandwidth of the coded excitation is similar to the transducer's bandwidth, because energy outside the transducer's band will not be efficiently converted to sound.

C. The code E can be pulse-compressed to produce a short pulse by matched filtering; here we use the simplest choice of filter, and observe that autocorrelation produces a band-limited impulse. The star operator $\star$ represents cross correlation.

D. Excited by a coded pulse, the transducer's output signal S is given by the convolution between the excitation and the transducer's impulse response. The resulting waveform is longer than T_D, and its spectrum is given by the product of the spectra of E and of T_D.

E. The acoustic echo response from the medium can be modeled as the sum of delayed and scaled replicas of the output signal S. To recover the range resolution defined by the signal bandwidth, matched filtering by cross correlation with S is applied to the RF data. For a single point scatterer, the result of such pulse compression is another band-limited impulse, with significant sidelobes.

can provide far better compression and sidelobe performance [5], [6]. Further complications arise when taking into consideration nonlinear phenomena. For example, attenuation in the medium adds a depth-dependent variation to the pulse shape. Choosing suitable excitations and pulse compression methods is an active area of imaging research.

While there are many reasons to use pulse coding in Medical Ultrasound and Non-Destructive Evaluation, the following examples illustrate some important applications of the approach. Many books and articles have been written on the subject; some useful references may be obtained in the articles by Misaridis et al. [7]-[9], and in Huang and Li [10].

Application: Increasing imaging depth

At a given drive voltage, large time-bandwidth pulses contain substantially more total energy than conventional pulses. After pulse compression, echo signals appear to have been produced using short pulses at voltages far greater than the actual drive voltage, and also much greater than the maximum permissible voltage given by the MI limit. Thus, in their most basic application, coded transmits can be used to extend the maximum imaging depth when the MI limit is more restrictive than the TI limit for short pulses.

Application: Increasing frame rate

Another common application of coded transmissions is increasing the acquisition frame rate by performing normally sequential acquisitions simultaneously. For example, a conventional line mode imaging sequence performs a series of Transmit-Receive acquisitions, each of which is used to obtain one scan line in the image. This procedure is done sequentially because the transmit beams for acquisitions of adjacent lines overlap and would interfere with each other. If different coded waveforms can be transmitted simultaneously on individual array channels, the encoding can be done in such a way as to make the individual beam signals distinguishable in post-processing. Thus, many different T/R experiments may be conducted in the same acquisition, in the time it takes for one acoustic round trip.

An extreme example of overlapping transmit beams is imaging with plane waves at multiple different angles; the pulses for each angle can be encoded independently, and the coded and delayed waveforms superposed into one complex waveform for each transmit channel. These emissions ensonify the entire image field during a single transmit event, and the resulting echoes can be separated into individual data sets for each angle in post-processing, and can then be reconstructed independently as though they were acquired in sequential events. The longer the pulse code, the less "crosstalk" remains between different codes after processing, and thus the pulse code length is an important design parameter. Note that the independence properties of the coded signals after transduction depend strongly on the transducer's bandwidth and pulse length limitations. It is the user's responsibility to choose the coding scheme and post-acquisition compression algorithm that is most suitable for their application.

One interesting approach developed recently uses Hadamard encoding, where only the polarity of the pulse is used to distinguish between transmissions [11]. The goal is to improve the SNR using conventional (short) pulses, while maintaining the frame rate. By transmitting only two different directions at a time, the authors explore combinations of up to 32 plane waves that are decoded without requiring pulse compression, using very simple processing that merely performs coherent compounding. Another approach to increasing frame rate by simultaneous coded transmissions on multiple channels is disclosed by Flynn et al. in [6] and [12]. The method transmits randomly generated binary phase-shift keying codes to enable retrospective synthetic transmit aperture imaging, and demonstrates additional benefits of reducing mechanical index while maintaining sensitivity.

Application: Time-reversal methods for reverberant environments

Time-reversal applications also require use of long time-bandwidth waveforms; in a classic time-reversal experiment, the coded waveforms are measured echoes from transmission of short pulses into a highly reverberant environment, that is, with extensive multiple scattering. When these long echo waveforms are reversed and transmitted back into the medium, the propagating waves are naturally pulse compressed by the medium itself, and can achieve enormous spatial focusing and temporal compression gains [13], [14].

Amplitude Modulation using Pulse Width Modulation

In medical ultrasound, advances in electronics and transducer design have permitted increasing the number of elements in acoustic arrays, and extending array geometry from 1-D to 2-D patterns. The large channel counts have always forced a compromise between transmit signal flexibility and cost, with most designs choosing unipolar or bipolar pulsers, with two or three voltage states. Because of the significant associated cost in hardware and power consumption, only a few specialized research ultrasound systems provide digital-to-analog converters (DACs) and linear amplifiers on each channel (Open System, http://lecoeur-electronique.com [15]). Therefore most excitation coding is implemented using pulsed signals with only two or three voltage levels, and signal design flexibility is greatly increased with the use of Pulse Width Modulation (PWM). PWM can be used very effectively to produce complex analog waveforms given the inevitable band pass filtering by the transducer itself.

PWM basics

Digital switching circuits typically transition between two states. Consequently, one might assume that they would be useless for production of waveforms with many different voltage levels, that is, with high dynamic range. In fact, PWM can be used to perform amplitude modulation with a surprisingly high dynamic range, when the digital switching speed is much greater than the highest frequency in the output signal. This approach has become commonly applied in power audio circuits using "class D" amplifiers [16], [17], [18], essentially power switching circuits, to produce acceptable audio quality sound very efficiently. The high efficiency is a consequence of the rapid transition time between on and off states that characterize the devices (often FET transistors); the *off* state presents a high impedance with little current and power dissipation, and the on state has low series impedance and allows most of the total power to be dissipated in the load. Conversely, Class A amplifiers are always operated *within* the transition region, and cannot be more than 50% efficient; they dissipate nearly the same power for all signal output levels, including the "quiet" regime with very low signal output.

Class A amplifiers were widely used because the signal amplification is nearly linear when the output levels remain small with respect to the total range of the transition region. With the advent of much faster devices, Class D operation has become a practical alternative.

The drive pulses can be thought of as a series of impulses, with magnitude given by the integral over each pulse "on" time. The rectangular drive waveform is greatly smoothed through low pass filtering by the transducer which effectively integrates over the pulses. The following figure illustrates the use of PWM in a classic audio amplifiers application [18], where the bandwidth of the switch is many orders of magnitude greater than the signal frequency.

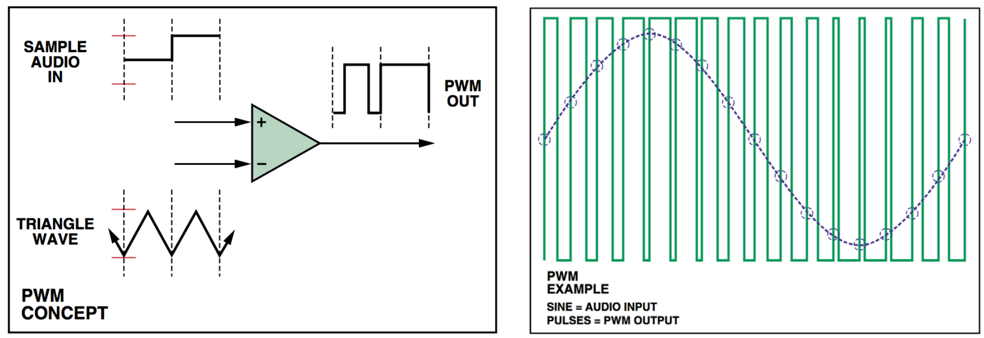


Figure 2. PWM encoding in audio signals using Class D amplifiers (Analog Devices).

When the switching transition time is only one or two orders of magnitude less than the fundamental output signal period (as is the case for Vantage), the pulses may be spaced at one rectangular pulse per half-cycle, and the pulse width is used to define the magnitude of the output signal during that half-cycle. The fundamental frequency of the output signal is determined by the spacing between pulses of the same polarity, as indicated in the diagram in the next figure below; the amplitude of the wave is related to the pulse width, defined using the PWM factor, indicated on the right.

Note that the amplitude of the fundamental frequency component of the wave is not exactly proportional to the ratio of pulse on- to off-time (PWM factor, or pulse duty cycle). This is simply a consequence of the Fourier decomposition of a rectangular pulse, graphically illustrated in figure 4.

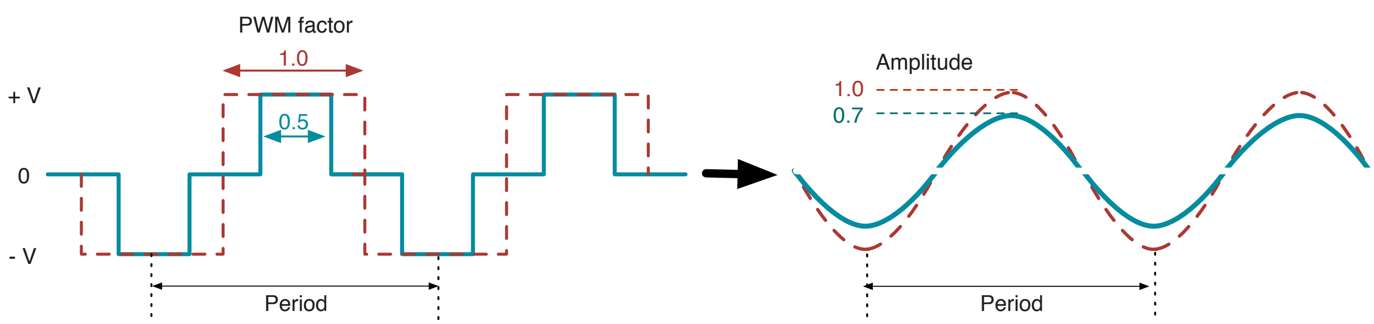


Figure 3. PWM encoding in high frequency signals using a single pulse per half cycle.

and expressed as the integral of the cosine function over the interval defined by the on-time.

$$A = 2\int_0^a cos(2\pi f t)\,dt$$

$$\text{with} \quad = \frac{sin(2\pi f a)}{\pi f}$$

$$a = \frac{PWM \cdot T}{4}$$

In the figure to the right, the magnitude of each harmonic is plotted versus PWM factor, for the first, third, and fifth harmonics. For clarity in observing the nulls, normalized magnitudes of 3rd and 5th harmonics are presented in the second figure. An example transmit frequency of 5 MHz is used to identify the limited number of possible PWM values given the 4 ns increment size. Note that the PWM value of 0.5 is not available at this particular frequency; in general, the exact value of PWM factor used in the hardware is derived from the ratio between the integral number of 4 ns counts for the on-time, divided by the number of counts defining a half cycle. Though the integral above assumes symmetry (even numbers only), in the Vantage transmitter these counts are integer values.

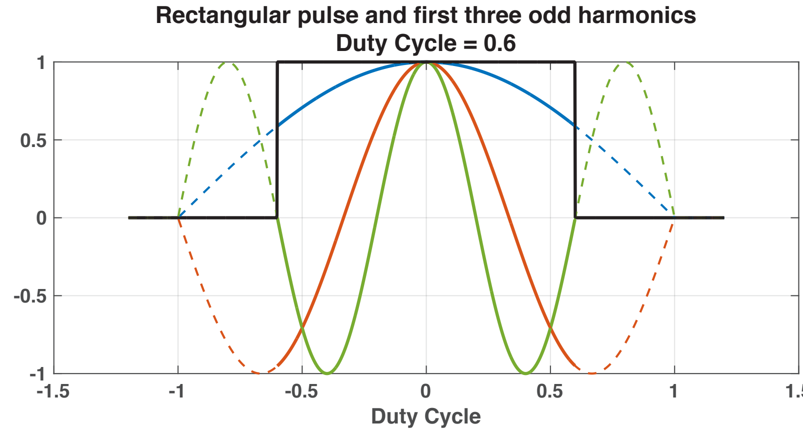


Figure 4. PWM can be used to set the amplitude of a harmonic wave, but the mapping between PWM value and magnitude is sinusoidal.

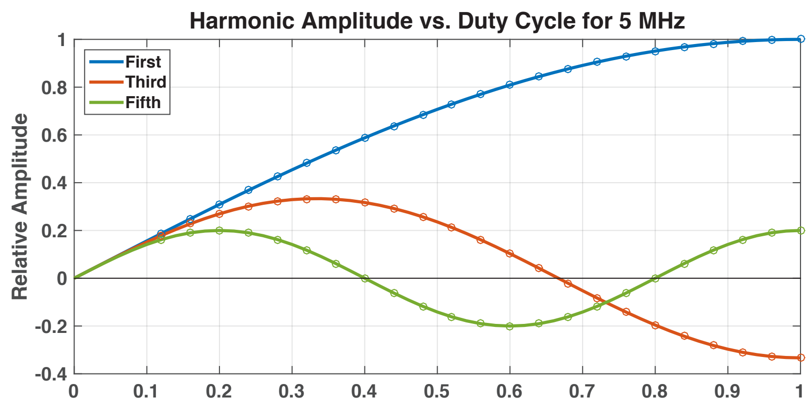


Figure 5. Harmonic amplitudes for a 5 MHz signal. Amplitude of the 1st, 3rd, and 5th harmonics, as a function of PWM factor normalized to the fundamental. The lines represent theoretical functions, and the circles represent the discrete value of PWM factor available for the 5 MHz wave, given the 250 MHz system clock.

Converting analog waveforms to PWM waveforms

The sections above have described the use of various types of waveforms for improving ultrasound imaging while satisfying hardware system and clinical safety constraints. Nevertheless, the challenge of creating desired analog waveforms using binary or trinary pulse sequences still remains.

Conversion of analog waveforms into PWM binary or tristate signals can be achieved in numerous ways. The conventional approach used in encoding audio signals is illustrated in Fig. 2; that method performs a binary comparison between a sawtooth waveform and the input analog signal to select the state of the switching circuit. The audio speaker, a band-limited load, effectively integrates the binary waveform and filters out the sharp transitions.

Another simple approach can be used for the case in which MHz ultrasound signals are encoded using one pulse per half cycle, the analog signal can be integrated over each half cycle between zero-crossings to find the relative magnitude for that half-cycle, and this result can be quantized by the clock interval (4 ns, for the Vantage) to set the pulse width. This rectangular pulse is then aligned with the location of the peak in the original signal by adding ground states before and after the pulse, as needed. Given a tristate pulser, positive and negative pulses can be treated independently, though it is helpful to maintain symmetry between adjacent positive and negative pulses to limit saturation, as discussed below, and generally to limit low frequency content that may be detrimental to the circuitry. This simple approach may be adequate for a particular application, but it neglects the role the transducer plays in filtering the tristate wave, and is surely less accurate than a method that takes the transducer's impulse response into consideration. The Verasonics Arbitrary Waveform Design toolkit includes the transducer's impulse response in an optimization algorithm that produces the tristate waveform that is needed to create the desired analog signal. Furthermore, it does so while satisfying hardware constraints described next.

Vantage pulser constraints

There are three important constraints for the Vantage pulser. First, the 250 MHz master clock sets the width of the smallest interval available to divide a pulse: the time grid is discretized by the 4 ns master clock period. Second, the transmit circuit requires at most three clock periods to complete a transition to any on-state; this "dwell' time requirement could be relaxed to two cycles for voltage swings that are substantially less than the maximum 96 V. Third, the integrated pulse (cumulative sum) must not exceed a limit defined by a transformer saturation threshold. For the standard frequency Vantage system, the pulse must maintain a time integral magnitude below 25 V μs, and the maximum length of any pulse is 175 clock periods (0.7 microseconds). The saturation limit is rarely a concern for conventional pulses which have a high degree of symmetry (except at very low frequency), but for arbitrary waveforms it is possible to exceed the saturation limit, especially when operating at low frequencies (e.g., below 2 MHz) and a series of pulses dominated by one polarity is encountered.

To increase the effective number of amplitude levels that can be encoded, positive and negative pulses can be combined: integrating over them can help overcome the 3 clock minimum dwell time limit. Nevertheless, any PWM approach is rather limited at higher frequencies (e.g., above 15 MHz), because the 4 ns clock interval becomes a significant fraction of the signal half cycle.

Programming the Vantage Transmitter

Vantage transmitter hardware characteristics

The Verasonics Vantage ultrasound transmitter uses a tri-state pulser that allows specification of arbitrary sequences of three voltage levels [+V,0,-V] at 4 ns clock intervals (250 MHz master clock rate). The voltage V is nominally constant for the duration of the pulse and is the same for all channels, but the tri-state sequences can be programmed independently for each channel, providing the flexibility required for simultaneous independent transmissions. For example, aperture apodization can be achieved by programming different PWM factors to produce different signal magnitudes across an array of elements. Furthermore, each acquisition event (including transmit and receive operations) may be programmed to emit a unique set of pulse sequences. The power supply output voltage may also be changed between events, but the supply is slew-rate limited and some transition time must be provided for this to occur.

The trinary sequences may be of arbitrary length, up to a maximum dependent upon the waveform complexity and transmit memory limitations. An internal storage format uses a compression algorithm to efficiently represent the trinary sequence and economize transmitter memory usage. This compression algorithm complicates specification of an absolute maximum pulse length for a complex waveform, because it is hard to predict how efficiently a particular waveform can be compressed. Nevertheless, one can assert that a perfectly incompressible pulse sequence, that allows for the 3-clock state transitions and uses all of the transmitter memory, will last more than 150 cycles, or about 30 μs at 5 MHz. The longest duration achievable in a single pulse is 10 Mcycles long, or 2 seconds at 5 MHz, for a waveform description that fits within the transmitter memory after compression.

It is useful to observe the result of practical circuit implementation on the transmitter output waveform given a tristate drive signal. As mentioned earlier, the transmitter output transition time between any two states is about 3 clock cycles due to output bandwidth limitations of the transmit pulser. In addition, the transmit circuitry includes internal impedance effectively modeled as an R-L series network, with Rs = 8 Ω, and Ls = 1.4 μH. Thus the waveform at the transducer connector combines pulser transitions and inductive response, as illustrated in Fig. 6 for a reasonable load impedance, here assumed to be 50 Ω.

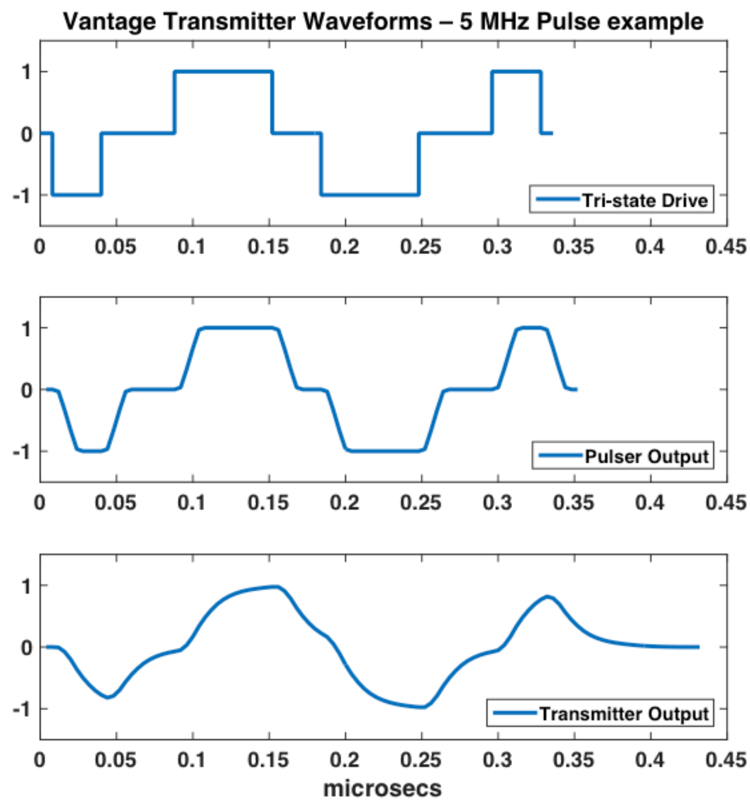


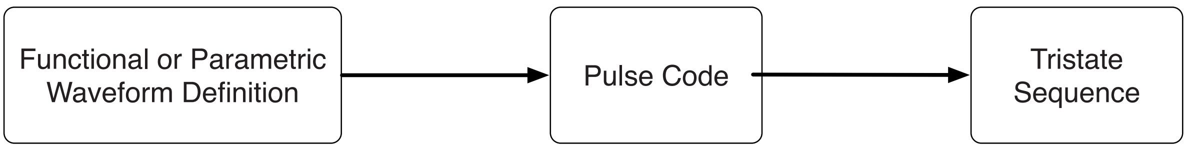
Figure 6. The tristate drive signal (upper panel) applied to the pulser is smoothed by the passband of the output FETs (middle panel). The drive signal into a non-reactive 50 Ω load is further modified by the series impedance of the transmitter which has an inductive component (lower panel).

Programming the pulser with built-in functions

The Vantage software interface provides several different methods to specify a trinary transmitted waveform. The simplest is the “Parametric” specification for a burst and includes the pulse frequency, pulse duty cycle or fractional pulse width, burst length (number of half cycles), and polarity, to generate a basic monochromatic pulse. This specification format is the standard offering, and the “Arbitrary Waveform” option is required to use additional methods.

More complex waveforms can be defined using the “Envelope” type, which produces an amplitude and frequency modulated pulse; and a “Pulse Code” specification that represents a direct definition of an arbitrary waveform. All waveform specifications are ultimately converted to Pulse Code, the Vantage standard waveform format. The Pulse Code is then passed to the compression algorithm, and that compressed representation is stored in the hardware transmit memory.

The Pulse Code specification contains a sequence of lines, each of which has five entries [Z1, P1, Z2, P2, R] representing the number of 4 ns clock periods for which the transmitter will be zero (Z1 and Z2), +V or -V (P1 and P2), and the number of times the line is to be repeated, R. The format for the Pulse Code is simply a Matlab (n,5) array, where n is the number of lines, and is easily converted to the explicit 250 MHz trinary sequence used to drive the transmit pulsers. The process is diagrammed in figure 7 below.




```
TW.type = 'parametric';
TW.Parameters = [Trans.frequency,.67,2,1];
```

Z1	P1	Z2	P2	R
2	-8	12	16	1
8	-16	12	8	1

{ 0 0 -1 -1 -1 -1 -1 -1 -1 -1 0 ... 0 1 1 1 1 1 1 1 1}

Figure 7. The process of specifying a Vantage transmitter pulse begins with a high level parametric specification of a pulse. This specification is converted to "Pulse Code", the intermediate level description of the tristate waveform as a series of cycles, one per line. Z1 and Z2 are clock counts representing the length of time the waveform is zero, and P1 and P2 are counts for the (+) and (-) states, respectively. The last entry R is an integer representing the number of times the pulse should be repeated. Finally, the tristate sequence can be easily constructed from the Pulse Code definition. The tristate waveform is not stored directly in hardware memory: instead, the sequence is compressed to reduce storage requirements, and decompressed on the fly in the hardware.

The Arbitrary Waveform Generator software toolkit

The pulse programming methods described above allow the user to specify the tristate waveform that drives the transmit pulser. A user will commonly specify an imaging pulse using the "Parametric" description, and accept that the acoustic waveform in water (1-way waveform) will be substantially different because it will be filtered by the transducer's 1-way impulse response. Similarly, the backscattered result at the receiver will be given by a convolution between the 2-way transducer impulse response, the medium scattering transfer function, and the transmitter signal driving the transducer. Depending on the transducer's transfer function, the received pulse may look very different from the programmed pulse. Indeed, because a resonant transducer impulse response is band limited and usually has some "ringdown" or internal reverberation, the result is to lengthen echo responses from ideal point scatterers. This artifact may be acceptable, but a procedure to design an excitation pulse that shortens the acoustic pulse is desirable; such a procedure solves an inverse problem that compensates for the transducer's impulse response.

The AWG toolkit (Arbitrary Waveform Generator toolkit) has been developed to examine and design the waveforms associated with a particular transmission. In particular, it is possible to have the tool design a tristate drive waveform to produce (an approximation of) a desired acoustic waveform, given the transducer's impulse response. Note that impulse responses provided by Verasonics for particular transducer models, and included with the AWG toolkit, also include the system transmitter's impulse response which results in the distortions illustrated in Fig. 6. As of this writing, only a few transducer responses are included with the toolkit, but more will be added over time (the user may always provide their own).

In addition to transducer impulse response compensation, the tool is helpful in producing any desired waveform within the transducer's passband, including time reversal recordings, AM and FM encoded signals, waveforms for coded excitation research, and others.

The algorithms used in the toolkit are described in [5], in the patent application [19] and also in the presentation slides (titled: Arbitrary Waveforms using a Tri-state Transmit Pulser) available on the Verasonics website. This approach uses knowledge of the transducer impulse response to develop an optimal least-squares solution to the inverse problem of finding a trinary pulse sequence that reproduces a given analog waveform. The residual error for the toolkit algorithm for a 5 MHz transducer is typically between -20 and -30 dB spread over the length of the pulse, depending on the transducer characteristics (lower frequencies have more dynamic range in PWM specification, and hence can produce waveforms with greater fidelity).

The essential idea behind the toolkit's algorithm is to use PWM to encode amplitude variations in the output signal, as template waveforms ("symbols"). These define a set of magnitude levels that can be used to approximate a particular analog waveform (see Fig. 8). Then an optimization process is used to produce a close approximation to the desired acoustic waveform by combining the symbols together, while adhering to pulser constraints described earlier.

The problem of estimating a transducer's impulse response from measurements is not a trivial one. Conventional methods measure the impulse response using a broadband pulser and a hydrophone or flat plate reflector. However, many potential sources of error can lead to inaccurate results; the accuracy of the toolkit-designed waveforms depends on the accuracy of the impulse response. The paper [5] describes a robust and accurate approach to determining the needed impulse response from a number of transmit / receive experiments, using a set of broadband

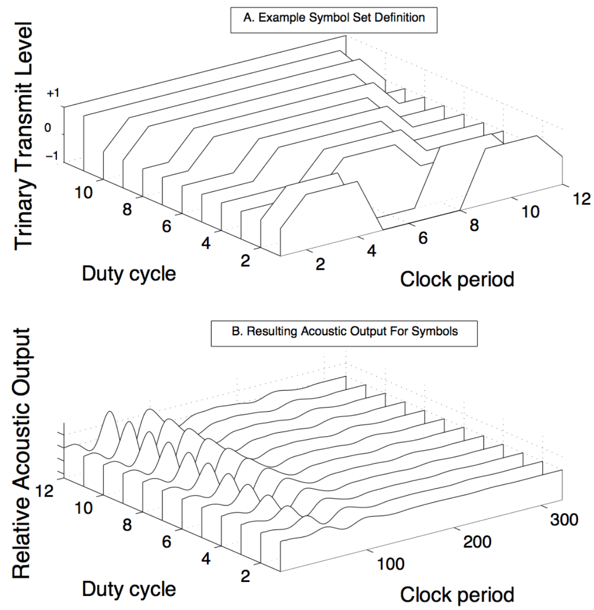


Figure 8. Pulse Width Modulated waveforms used to encode impulses of varying amplitude. See the proceedings paper [5] and a set of presentation slides on the website for details.

random waveforms. This estimation algorithm and procedure is not yet included in the toolkit. The toolkit is a GUI-based program, and its operation is fully described in a separate manual distributed with the software.

Application Examples

Golay encoding for improved SNR and extended imaging depth

As discussed in an earlier section, large time-bandwidth pulses are often used to improve imaging penetration depth when transmit signals are voltage-limited due to hardware constraints or ultrasound exposure safety limits. As in Fig.1-D, a long excitation is transmitted into the medium, and the backscattered data must be pulse compressed, Fig.1-E. Cross-correlation can be used to compress the received data using a matched filter derived from the original excitation pulse. Typically, the resulting (compressed) data suffers from large correlation sidelobes, which add to image clutter.

Complementary pairs of Golay sequences are constructed in such a way as to produce correlation sidelobes that are equal and opposite in polarity, and thus two transmit-receive experiments using the two Golay waveforms can be used to cancel the sidelobes in the pulse compression step by simple summation of the results. See the work by Nowicki et al. for background information on Golay sequences [20], [21]. Here, the arbitrary waveform synthesis software tool was not used to design the pulses. The goal of this exercise was to demonstrate the use of the Arbitrary Waveform Generator by directly programming Golay coded pulse trains using the Pulse Code format to improve penetration depth.

In this example, complementary Golay binary sequences of length 16 were convolved with a short wavelet to produce the tristate drive waveforms. This was done by hand: define the pulse code for an elementary wavelet at 5 MHz., for both positive and negative polarities, and then concatenate the pulse code lines following the Golay sequences given by:

A = {1 1 1 -1 1 1 -1 1 -1 -1 -1 1 1 1 -1 1}, and
B = {1 1 1 -1 1 1 -1 1 1 1 1 -1 -1 -1 1 -1}.

The resulting pulse code produces the transmitter drive signals illustrated below in Fig. 9, followed by a plot of the simulated acoustic waveforms obtained through convolution of the drive signals with the ATL L7-4 transducer impulse response.

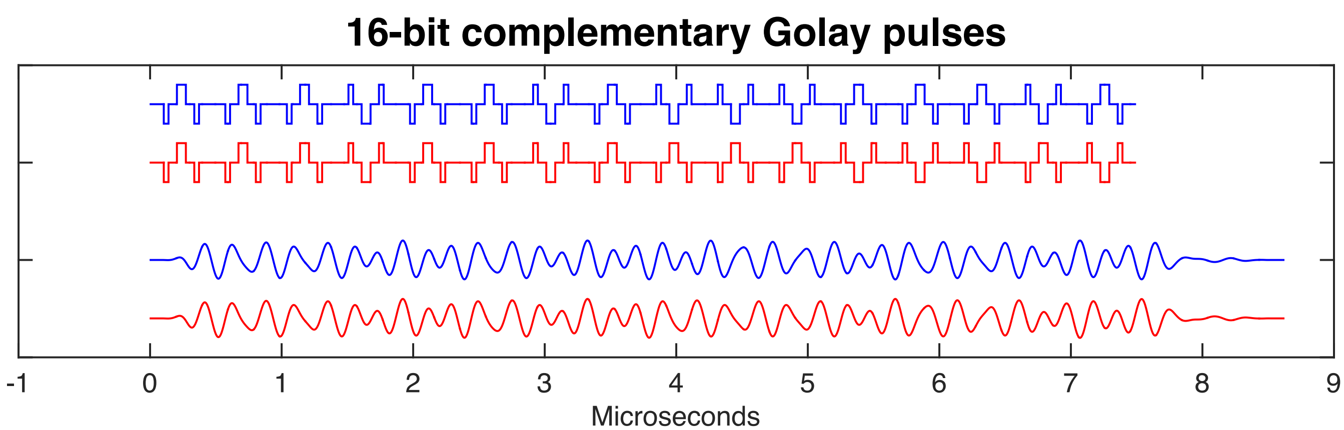


Figure 9. Tristate drive signals for 16-bit complementary Golay codes, using a simple wavelet to carry each bit (upper panel), and resulting acoustic waveforms for a L7-4 transducer (5MHz, 60% bandwidth) (lower panel). The panels are registered (the horizontal axes are correctly scaled and time-aligned) and the lag between electrical drive and acoustic output is clearly observed, as is the lengthening of the drive pulse due to the characteristics of the transducer's impulse response.

To show how the complementary Golay pulses work to reduce correlation sidelobes, the simulated pulses are pulse-compressed by autocorrelation to obtain the transducer's impulse response alone (simulating the situation in which there is only one point scatterer in the acoustic medium). This process produces the following waveforms, in which it is clear that the main lobes are very similar, and the sidelobes are equal and opposite; when summed, the result indeed recovers the L7-4 impulse response.

Two T-R events are required to collect the RF Data for a Golay pair. Each RF line of echo data is cross-correlated with the corresponding Golay

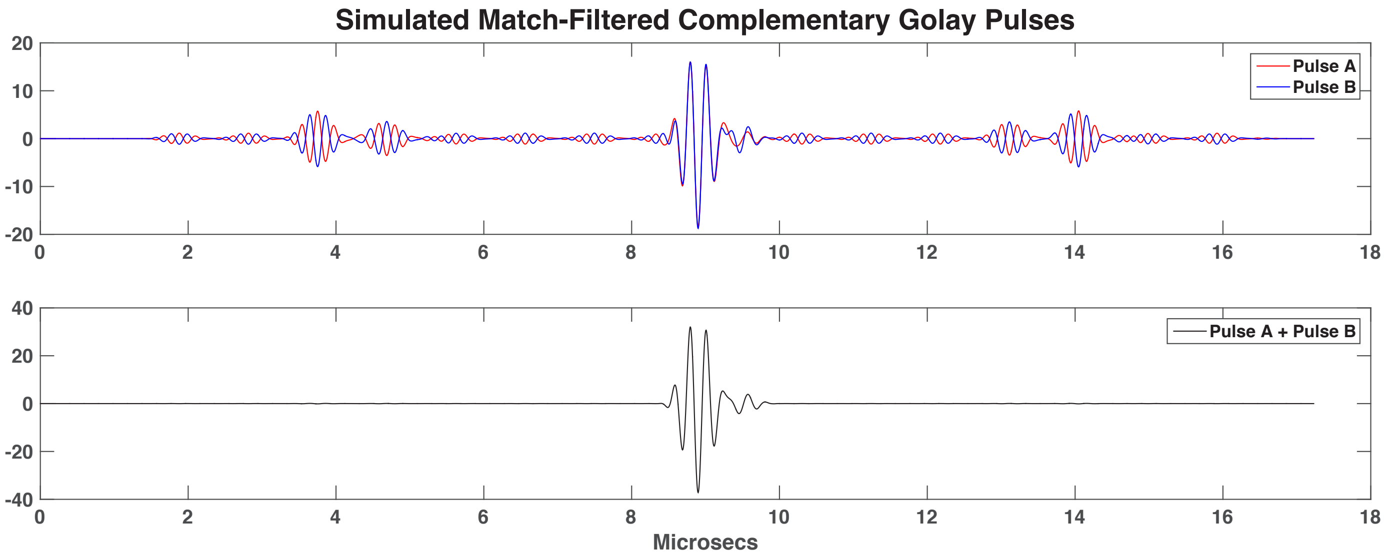
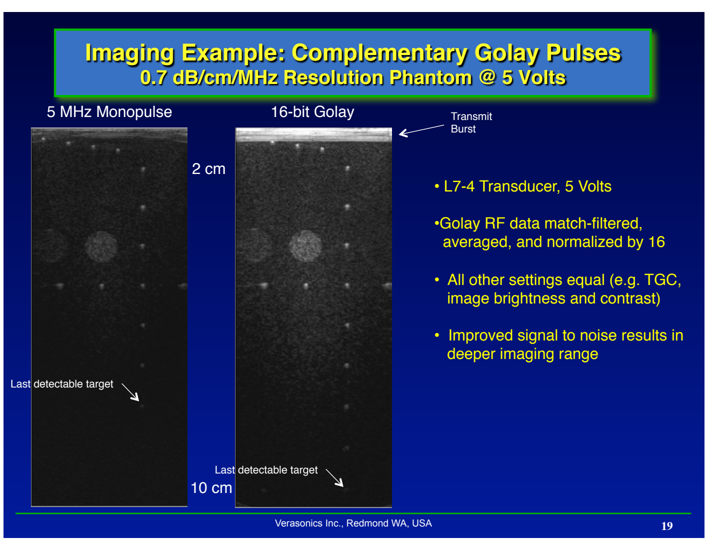


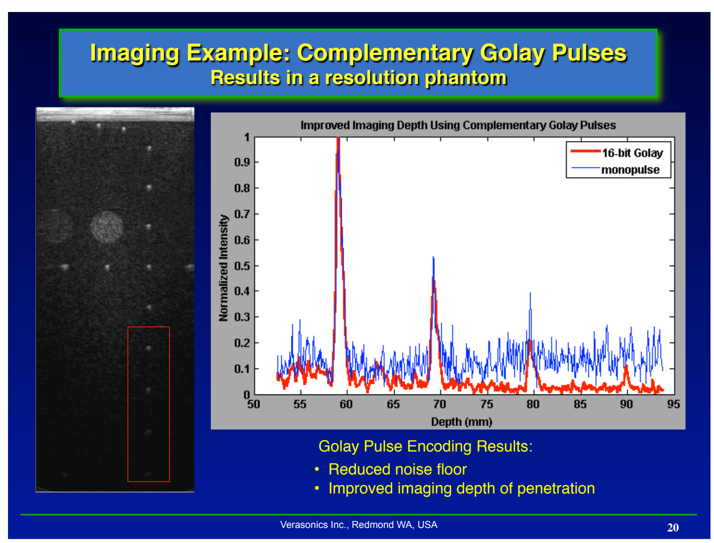
Figure 10. Autocorrelated signals for each of the acoustic waveforms in Figure 9 (upper panel). The summed result is presented in the lower panel.

acoustic transmit waveform, and then the two results are summed to produce the desired RF data. This pulse-compressed RF data is then processed using the usual data processing chain to form an image.

An example set of images comparing imaging performance using the signals described above and a tissue mimicking string phantom is presented next.

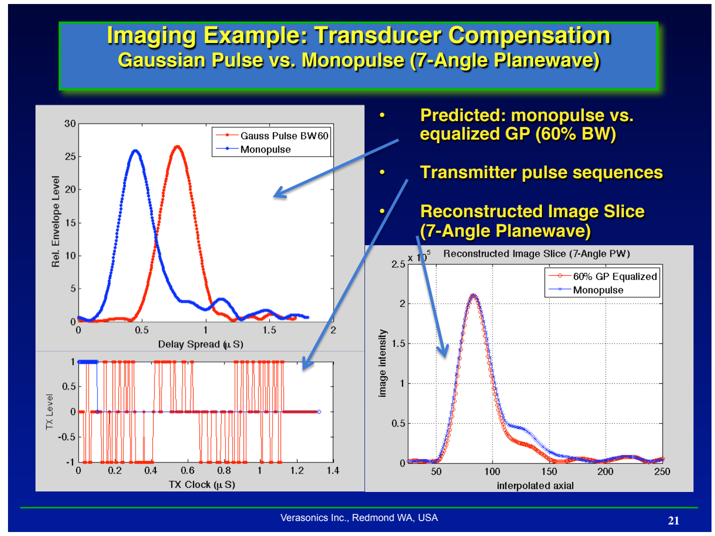
Finally, an examination of the image line passing through the point targets illustrates how the SNR improves when using the Golay excitations, at the same transmission voltage. In effect, after normalizing the intensity of the first scatterer, the noise floor appears reduced when using the Golay approach.

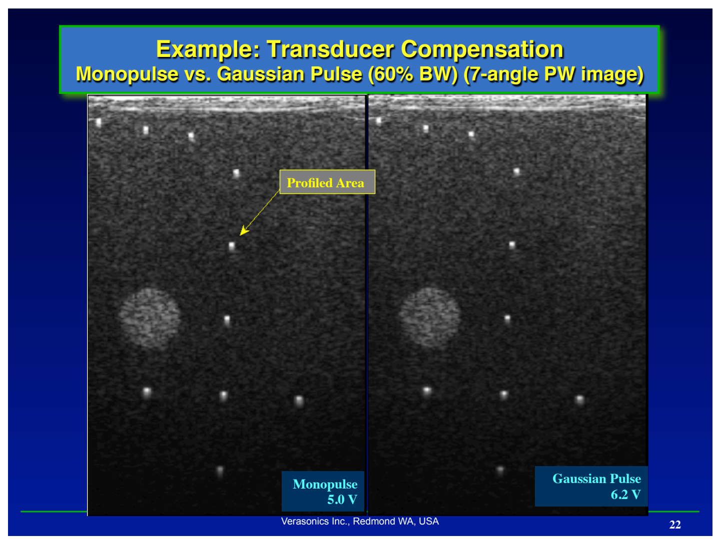




Transducer compensation

This example uses the Analog Waveform Design toolkit to improve the temporal response of the transducer, by using a drive signal that partially compensates for the long impulse response. The drive signal tends to “equalize” the spectrum of the transducer, by adding energy to the outer parts of the passband, and doing so with the proper phase so that the resulting output signal is shorter than the original impulse response. Note how complex the new drive signal is compared to a simple impulse; the resulting output is indeed more compact, but is emitted later than the impulse response. When the pulse intensities are overlapped, the improvement is easily seen.





APPENDIX A. Frequently Asked Questions

What is the Arbitrary Waveform Generator Toolkit?

The Verasonics Vantage platform has an independently programmable transmitter on each channel that can produce arbitrary tristate waveforms. The hardware can be programmed directly using high level software to define a tristate drive waveform. The software toolkit provides additional capability to visualize (analog) acoustic signals resulting from the tristate pulse sequence driving a particular transducer. The kit also includes a tristate waveform design tool which produces a sequence of pulse-width modulated pulses that will result in a desired output acoustic waveform, given the transducer's impulse response. This is very useful for creating amplitude modulated waveforms, coded excitations, and for encoding time-reversal experiment recordings. The Arbitrary Waveform Toolkit is an add-on option that enables the hardware (Arbitrary Waveform Generators) and includes the additional GUI-based design software.

Why not use a DAC and linear amplifier on every channel?

Conventional arbitrary waveform generators uses Digital-to-Analog Converters (DAC) and linear amplifiers, and these are straightforward to use and provide high dynamic range drive signals, but very expensive to implement in large channel count systems, particularly when these require high power, and are far less electrically efficient as well. Unless extremely accurate reproduction of specific waveforms is required, using a tristate pulser with PWM control on each channel is a much more cost effective hardware solution. Furthermore, Verasonics provides the software to design the tristate pulse train given a desired analog waveform as part of the Arbitrary Waveform Generator Toolkit.

What is the maximum length of an arbitrary waveform that can be transmitted by the Vantage signal generators?

The maximum transmit waveform length depends on the complexity of the desired waveform and the output frequency, because a compression scheme (based on decomposition of the waveform into nested loops of repeated states) is applied prior to storing the waveform in transmit memory. The simplest answer is to give an example: at 5 MHz, a completely incompressible waveform that fills transmit memory will last just over 30 microseconds (equivalent of 150 cycles). However, because the transmitter bandwidth is limited to about 20 MHz, some repetition of 4 ns states will be required and the minimum length of a "fully complex" 5 MHz waveform will likely be greater than 90 microseconds (>450 cycles @ 5 MHz, equivalent). A more accurate estimate requires use of the waveform compressor applied to the pulse code representation for a particular waveform of interest. The pulse length scales inversely with frequency. Note that the receivers are active during transmit, but are saturated if the transmitter on that channel is active, thus "blanking" the receive during transmission of the pulse.

APPENDIX B. References

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