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5aBAb7. Superresolution imaging in ultrasound B-scan imaging

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A number of imaging systems exhibit speckle, which is caused by the interaction of a coherent pulse reflecting off of random reflectors. The limitations of these systems are quite serious since the speckle phenomenon creates a pattern of nulls and peaks from subresolvable scatterers or targets that are difficult to interpret. Another limitation of these pulse-echo imaging systems is that their resolution is dependent on the full spatial extent of the propagating pulse, usually several wavelengths in the axial or propagating dimension and typically longer in the transverse direction. This limits the spatial resolution to many multiples of the wavelength. This paper focuses on the particular case of ultrasound B-scan imaging and develop an inverse filter solution that eliminates both the speckle phenomenon, and the poor resolution dependency on the pulse length and width, to produce SURUS (super-resolution ultrasound) images. The key to the inverse filter is the creation of pulse shapes that have stable inverses. This is derived by use of the standard Z-transform and related properties. Although the focus of this paper is on examples from ultrasound imaging systems, the results are applicable to other pulse-echo imaging systems that also can exhibit speckle statistics.

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INTRODUCTION

Most analytical models of pulse-echo imaging derive an integration on the product of propagating pulse and the reflectors or scatterers, over the location of the pulse at some point in time (Macovski, 1983; Szabo, 2004; Prince and Links, 2006). Under a number of approximations and simplifications about attenuation and diffraction, the integration can be reduced to a convolution model (Macovski, 1983).

The problems of poor resolution and speckle can be understood as a direct result of this convolution. The spatial resolution is set by the full spatial extent of the propagating pulse in 3D which is typically many multiples of a wavelength. However, in tissue, small scatterers at the cellular level and micro- structural level such as the arterioles and capillaries will have a dimensions on the order of 10 microns, much smaller than typical pulse shapes (the wavelength at 10Mhz is 150 microns, for example, and the pulse length will be multiples of this). With many subresolvable scatterers interacting with a propagating pulse, the resulting echo will exhibit the random constructive and destructive interference pattern known as speckle (Burckhardt, 1970; George and Jain, 1973; George *et al.*, 1976; Burckhardt, 1978; Wagner *et al.*, 1983; Tuthill *et al.*, 1988; Reynolds *et al.*, 1989). The problems with a visual interpretation of speckle images are profound, since the patterns of nulls and peaks may or may not correspond to actual nulls or peaks of the scatterers, but rather to their summation over the positive and negative portions of the propagating pulse. Furthermore, the lesion detection problem is made more difficult by the overlap of Rayleigh statistics from different distributions of a lesion and a background (Sperry and Parker, 1991; Cramblitt and Parker, 1999). Attempts to improve the situation include speckle reduction algorithms (Bamber, 1993) and deconvolution approaches (Jensen, 1992; Alam *et al.*, 1998; Haider *et al.*, 1998; Taxt and Frolova, 1999; Qinzhen *et al.*, 2003; Michailovich and Adam, 2004; Kerner and Porat, 2008; Shin *et al.*, 2010). Despite all these attempts, the typical medical ultrasound image still retains the two characteristic elements of: resolution limited by the pulse size and shape, and speckle statistics. There are reasons to suspect that this is an intractable problem. The spectrum of a typical pulse is a band-pass signal, so the DC, very low frequency, and very high frequency components of the reflectors are not captured. This limits the amount of "whitening" or equalization that can be accomplished. From a 2D imaging point of view, the illumination of k-space by a typical pulse is a discouragingly small area of support, constraining the strategies for improving image quality while reducing speckle (Munson and Sanz, 1984).

We recently demonstrated an approach to an inverse filter solution with stabilized ultrasound pulses (Parker, 2012). Stabilized pulses, in this context, means realizable continuous functions in the axial and transverse directions, that when sampled have their Z-transform zeroes lying within the unit circle. This corresponds to inverse filters that are stable because their poles lie within the unit circle, such that they are limited in time with bounded output. By applying an exact, stable inverse filter, the final result is a very high resolution, subwavelength solution to the distribution of scatterers that were previously below the resolution of the ultrasound pulse and the imaging system. The integration of random scatterers over the pulse length and width is essentially disaggregated by the inverse filter operation. Therefore, the two dominant and problematic system effects of pulse length and speckle statistics are eliminated, replaced by a more favorable and high resolution calculation of the distribution of scatterers within tissue. The solution is exact within the framework of the convolution model and sampled signals, yet is approximate in the sense that the sampling frequency (as low as twice the center frequency of the transmit pulse in some following examples) will result in aliasing of components above the Nyquist frequency. The solutions are also accurate with respect to the physical reality to the extent that the convolution model is accurate and the effect of noise is limited. The resulting images are termed SURUS images, as they are super-resolution ultrasound images.

THEORY

Starting with 1D theory for simplicity, wish to find a stable inverse filter for a pulse $p[n]$. For this we turn to the Z-transform of $p[n]$. The one-sided Z-transform of $p[n]$ is given by (Oppenheim and Schafer, 1975):

$$P(z) = \sum_{0}^{\infty} p[n] z^{-n} \tag{1}$$

For a pulse of length N, the Z-transform is a polynomial of order N-1, which can be factored into roots, giving zeros of the Z-transform. The inverse filter, given by the transform $1/P(z)$, will convolve with $p[n]$ to produce an impulse. However, it is clear that the reciprocal nature of $P(z)$ and its inverse filter transform implies that the zeroes of the pulse transform $P(z)$ become the poles of the inverse filter. Generally, for a casual, right-handed system to be stable the poles of the Z-transform must lie within the unit circle and the region of convergence includes the unit circle. This is analogous to poles of a stable system lying in the left half plane for Laplace Transforms. With poles on or outside the unit circle, the impulse response of these systems would be unstable and unbounded.

Unfortunately, the typical ultrasound pulses used for imaging are functions that, when sampled, have Z-transforms with many zeroes on and outside of the unit circle (see Michailovich and Adam (2004) for examples). These produce inverse filters with poles outside of the unit circle, leading to unstable filters.

One way to create stabilized pulses (meaning pulses that, when sampled, posses stable inverse filters) is to multiply $p[n]$ by the quantity β^n, where β is a real number < 1. In the discrete world if a right-sided sequence $p[n]$ with a Z-transform $P(Z)$ is multiplied by an exponential sequence β^n, then (Oppenheim and Schafer, 1975; Jackson, 1991):

$$Z\left[\beta^n p[n]\right] = P\left(z/\beta\right) \qquad (2)$$

Thus the multiplication by a geometric series creates an asymmetric pulse in the time domain with its Z-transform zeroes "retracted" into the unit circle depending on the factor beta. A similar consideration applies to samples of the transverse beam function $s[m]$, and examples are provided in the next section.

RESULTS

First, we examine a conventional pulse shape $p[t]$, which is modeled as a Gaussian envelope modulated by a cosine at the center frequency of the transducer as shown in Figure 1a. The continuous function is sampled at twice the center frequency and 15 points are taken as $p[n]$. The Z-transform of this sampled pulse has numerous zeroes on the unit circle, and a pair of conjugate zeroes outside of the unit circle, as seen in Figure 1b.

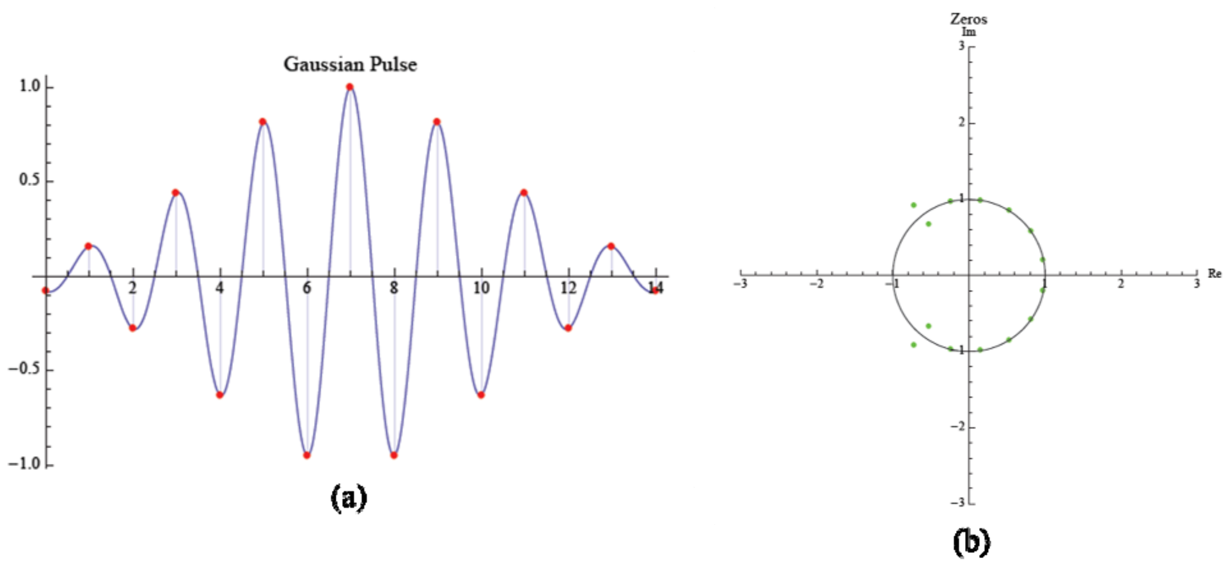

FIGURE 1. A conventional pulse with a Gaussian envelope, sampled at twice the center frequency in (a). The zeroes of the Z-transform of the sampled pulse is shown in (b). Since some zeroes lie in and around the unit circle, the inverse is unstable and unbounded.

These zeroes will become poles of the inverse filter and signify an unstable, unbounded output result. Therefore, this class of typical pulse echo shape is not conducive to inverse filters. However, by modifying the function with a geometric series, a beta term in Equation (2), the pulse can be made asymmetric and the inverse transform is stabilized. As an example, the Gaussian function in Figure 1a is multiplied by 0.7^n, and the new function is shown in Figure 2a.

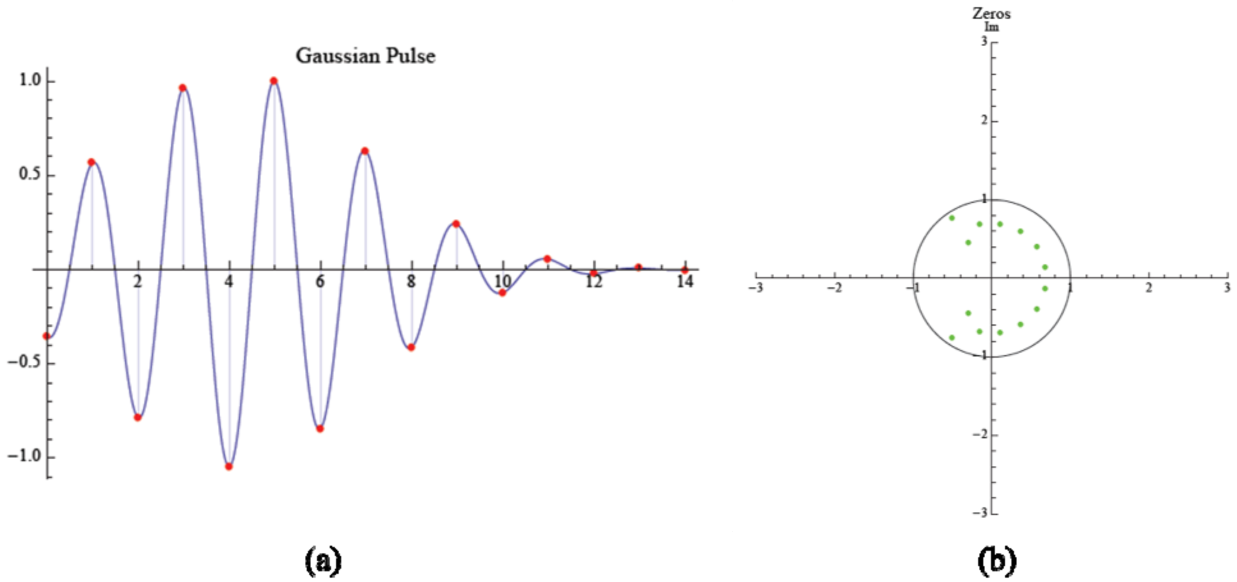

FIGURE 2. An asymmetric pulse formed by multiplying the Gaussian envelope with a geometric series is shown in (a). The zeroes of the Z-transform are retracted into the unit circle as shown in (b). This leads to a stable inverse filter.

Now all the zeroes of the transform lie within the unit circle, as seen in Figure 2b. Accordingly, the inverse filter will have poles within the unit circle and will have a bounded input/bounded output impulse response of limited duration.

In general, we have found that the formation of a stabilized pulse is not restricted to the use of a β^n type function; rather this is illustrative of envelopes that have a sharp initial rise and a more gradual fall-off from the peak. We call these "asymmetric" envelopes or pulses, and these can be characterized by a number of different analytic functions. As an example, the function $p[x] = x^2 \mathbf{e}^{-x^2/2\sigma^2}\text{UnitStep}[x]$ is selected as a stable function for the transverse beampattern. The program Field II (Jensen, 2004) was used to simulate a focused beampattern. The approximate Fourier transform of $p[x]$ is used to set the apodization function. A 5 MHz transducer with 129 active elements is simulated with half-wavelength spacing. The transverse beampattern at the focus (60 mm depth) is shown in Figure 3. There is a good match between the beampattern and the design function $p[x]$.

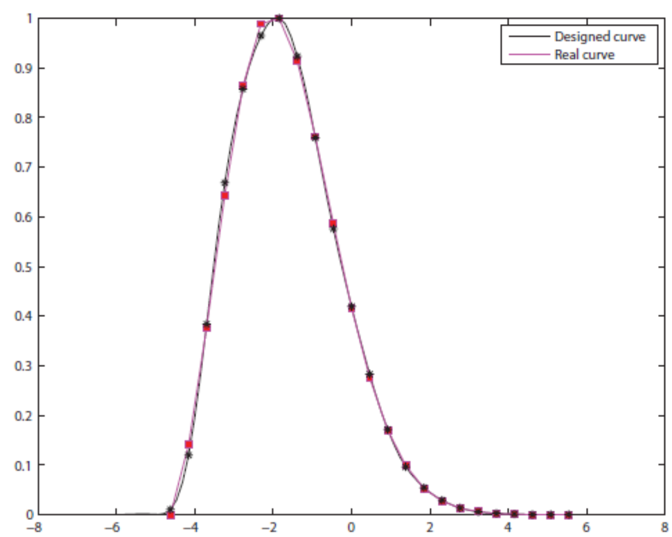

FIGURE 3. An asymmetric function (heavy line) is used as a design for a focal transverse beampattern (squares) at 5 MHz. Vertical axis: magnitude of the focal transverse beampattern (arbitrary units). Horizontal axis: transverse distance in mm. There is good agreement between the design and the realized beampattern

DISCUSSION AND CONCLUSIONS

An inverse filter approach has been derived using the Z-transform on stabilized but realizable pulses. A major issue in the use of the inverse filter is the realization of stabilized waveforms in conventional focused systems.

Can these asymmetric pulses be produced in practice? In fact it is straightforward to show that the Fourier Transform of the conventional symmetric beamshapes, and those of the asymmetric versions, are reasonably contained within a similar support or bandwidth. That means that a transducer of limited bandwidth can, with some modification of the excitation, produce either the symmetric or the asymmetric (stabilized) version of $p(t)$. For the transverse beam pattern, this means that an aperture with limited support can similarly produce either the symmetric or the asymmetric (stabilized) version of the beam pattern.

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