Adaptive Focusing Through Layered Media Using the Geophysical “Time Migration” Concept

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Abstract — Virtually all practical algorithms for aberration correction in medical ultrasound have thus far modeled the aberrating tissues with a thin time-delay screen. While this assumption is probably reasonable for small image regions (isoplanatic patches), practical application is still dif£cult. In many cases, an inability to estimate the screen parameters with suf£cient accuracy in the presence of aberration and speckle targets has led to disappointing performance. A new aberration correction approach is proposed, inspired by the geophysical imaging concept of time migration. This technique is motivated by considering complete, bistatic, pulse-echo data acquired from layered media, where sound speed is a function of depth only. Re¤ection travel times as a function of source-receiver offset in such a model are approximately hyperbolic, just as they would be if the sound speed in the medium were constant and equal to the rms speed of the layers. Seismic imaging practice has shown this approximation to be robust in the presence of minor lateral speed variations. By focusing each point in the image using a constant sound-speed assumption, but allowing this assumed speed to change from point to point, a well-focused image may be obtained. A focusing criterion is all that is needed to determine the optimum focusing speed at each image point, without *a priori* knowledge of the medium properties. FDTD simulations provide synthetic data acquired from a 64-element array. A simple skull model was interposed between the array and targets in one simulation; in another, a speckle-producing region with embedded cysts was imaged through a Gaussian-shaped, high-speed anomaly. In both cases, images formed using different assumed sound speeds show different parts of the image in good focus. Application of the proposed focusing criterion produces a composite image showing improvement over any single image formed assuming a constant speed of sound.

I. Aberration Correction

Much attention has been given to improving the spatial resolution of medical ultrasound by using higher frequencies and/or increasing the size of the transducer aperture. The attenuation properties of tissue impose an upper limit on frequency, yet under this constraint, diffraction theory still permits a signi£cant improvement in resolution over current systems.

The main obstacle to improved spatial resolution in medical ultrasound is the variation of sound propagation speed in tissue. The true speed of sound deviates from the assumed value by up to 10% in soft tissues and much more in bone [1]. Although ultrasound depends on the echoes generated by scattering from tissue inhomogeneities, current systems assume a bulk homogeneity—straight raypaths and a constant speed of sound at scales above a wavelength. If this assumption fails, imaging resolution suffers as the point spread function becomes broader. Aberration seriously harms the diagnostic usefulness of ultrasound [2].

Virtually all of the published algorithms for aberration correction that would be practical *in vivo* are based on a screen model of tissue aberration. In it, the effects of aberration are modeled entirely by variable time delays on the received and transmitted signals due to a thin screen at the aperture. If these delays can be estimated correctly, perfect focusing is, in theory, easily accomplished by adding compensating delays to the geometrically determined hyperbolic travel-time pro£le.

The screen model clearly fails to describe reality in most medical imaging scenarios; nevertheless, one may consider a set of delay corrections to be valid for a small region of the imaging area called an *isoplanatic patch*. By calculating a different set of delay corrections for each isoplanatic patch, the entire image may be corrected. The necessary time shifts have been estimated by cross-correlation of adjacent received signals following a focused transmit pulse [3], cross-correlation of common-midpoint signals [4, 5], and iterative optimization of an image quality metric [6].

Very few of the screen-based algorithms for aberration correction have gone on to be implemented in commercial scanners, and the results in clinical situations have not been as dramatic as expected. If the weak-scattering assumption still holds, however, the disappointing performance of screen algorithms in ultrasound is likely due to the dif£culty of estimating the time shifts with suf£cient accuracy, and not to any fundamental failure of the screen model. This

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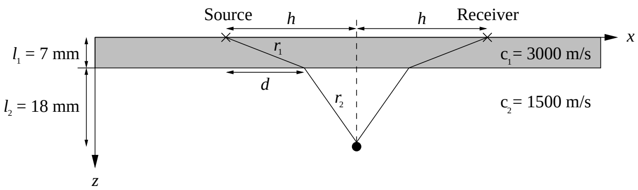

Figure 1: A simple two-layer model illustrates the $c_{\rm rms}$ approximation. The travel times for a scatterer 18 mm beneath the "bone" are modeled almost perfectly by assuming propagation in a homogeneous medium with sound speed $c_{\rm rms} = 1830$ m/s.

view is supported by a recent study of seismic imaging with simple time-shift corrections in a complex earth model [7]. It was found that a careful choice of one wave arrival at each position along the aperture (corresponding to a choice of time delays for a screen method) resulted in high-quality images, even though multiple arrivals were obvious in the data but ignored.

Our new approach to the aberration correction problem borrows concepts from geophysical (exploration seismic) imaging, where numerous techniques have been developed over the past £fty years for acoustic imaging of the Earth. The acoustic properties of the Earth vary far more than those of biological tissues; a constant propagation speed assumption has never worked well in this £eld. *Time migration* is the term applied to one class of imaging algorithms which estimate the unknown speed. (In seismic terminology, a *migration* algorithm is an image formation algorithm.)

In the following sections it is assumed that echo data is available in a *complete* format, that is, with a separate record for each pair of individual transmit and receive elements. (For the 1-D arrays considered here, this means starting with a data cube having transmit element position, receive element position, and time as its three axes.)

II. Focusing Through Layered Media Using an rms Speed of Sound

Consider the case of a horizontally layered medium in which the wave speed is a function of depth z only. Figure 1 shows one such model with a re¤ecting target beneath a layer of "bone." An expression for travel time versus offset h may be written as

$$t = \frac{2}{c_1}\sqrt{l_1^2 + d^2} + \frac{2}{c_2}\sqrt{l_2^2 + (h-d)^2}. \qquad (1)$$

The true ray-path may be determined by £nding the stationary point with respect to d, following Fermat's principle of least time which he used to derive Snell's law,

$$\frac{dt}{dd} = \frac{2d}{c_1\sqrt{l_1^2+d^2}} - \frac{2(h-d)}{c_2\sqrt{l_2^2+(h-d)^2}} = 0. \qquad (2)$$

Obtaining an expression for the travel time in terms of h, l_1, l_2, c_1, and c_2 is far more dif£cult, however. It involves solving a quartic equation which does not have a compact solution.

It was shown in [8] that for the general case of any number of ¤at layers having different thicknesses and sound speeds (thus, in the limit, for any $c(z)$ medium),

$$t^2(h) = k_1 + k_2h^2 + k_3h^4 + k_4h^6 + \cdots \qquad (3)$$

where $k_1 = t_0^2$ (the squared vertical two-way travel time) and the other $\{k_i\}$ are functions of $c(z)$. The second-order approximation is of special interest, because then the travel time as a function of source-receiver offset is a hyperbola, just like the case of constant sound speed. (For a re¤ector at depth z' in a medium with contant c,

$$t \;=\; \frac{2}{c}\sqrt{h^2 + z'^2}.) \qquad (4)$$

It turns out k_2 has a nice interpretation in terms of the *rms sound speed* $c_{\rm rms}$:

$$k_2 = \frac{4}{c_{\rm rms}^2} \;\; \text{where} \;\; c_{\rm rms}^2 = \frac{1}{t_0}\int_0^{t_0} c^2(t)dt. \qquad (5)$$

The integral of squared wave speed is taken over the round trip of the vertical ray, i.e. for a N-layer model where t_n is the vertical two-way time and c_n is the speed in the n-th layer,

$$c_{\rm rms}^2 = \frac{\sum_{n=1}^{N} c_n^2 t_n}{\sum_{n=1}^{N} t_n}. \qquad (6)$$

This hyperbolic approximation is surprisingly accurate for source-receiver separations equal to or less than the target depth. For the example in Figure 1 and a 20-mm array aperture, the worst-case phase error at 2.6 MHz is only about 20 degrees.

The implication, subject to the limitations of the hyperbolic (second-order) travel time approximation, is that *targets in a horizontally layered $c(z)$ medium may be accurately focused by assuming a constant sound speed of $c_{\rm rms}$ in the imaging algorithm.* Of course, the correct "constant speed" will vary from point to point in the image. This is the essence of *time migration*, so called because, since the true target depths are unknown, the "depth" axis of the £nal image is left in units of time. In geophysical applications, time migration has been successful in obtaining good images even though the subsurface structure is not exactly $c(z)$ in practice [9].

Smith et. al. [10] investigated ultrasound focusing through single distorting layers, including the skull, with promising experimental results, but they did not note the suitability of the hyperbolic approximation. This approximation is crucial, for it allows an imaging algorithm to determine the point-by-point $c_{\rm rms}$ blindly using some focusing metric, without the need for *a priori* knowledge of the intervening layers' wave speeds and thicknesses.

Note that this technique does not handle multiple scattering properly, for it presumes a single-valued travel time from any re¤ecting target to the array. Thus, it can be viewed as a screen method, much like the previous approaches for aberration correction mentioned in the introduction. A key difference, however, is a reduction in the number of unknown variables from N (a time shift on each array element) to one (the rms focusing speed).

III. Time Migration Algorithm

A practical time migration algorithm starts with a series of images formed using a constant-speed-of-sound assumption at a set of sound speeds spanning the expected range of c_{rms} for all the targets in the image. The £nal output is a composite image carved from this set of images, where c_{rms} at each point is chosen based on some quality measure.

The component images are formed using dynamic focusing on both transmit and receive. This highlights re¤ections consistent with the chosen sound speed and maximizes the rejection of other re¤ections. If a complete data set is being used, focusing may be envisioned as a coherent sum over the 2-D surface corresponding to the point-target response for each focus point. Alternatively, the transmit focusing could be performed "online" as in conventional ultrasound beamforming, but the requirement for many complete scans at various c with full dyanmic focusing in transmit and receive makes this a time-consuming proposition.

The general approach of estimating c_{rms} from the constant-speed images is known as *migration velocity analysis*, and many techniques have been described in the seismic imaging literature (see e.g. [11, 12]). Only a very simple procedure, used for the simulations which follow, will be described here:

1. From the ensemble of constant-speed, dynamically focused images, pick the brightest point. Set c_{rms} for this point to the speed index of its parent image.
2. Find the next-brightest point. If it is a suf£cient distance from the last point, set its c_{rms} to the speed index of its parent image, otherwise skip.
3. Continue picking points in decreasing order of brightness (subject to a minimum spacing constraint). Each time, use the established estimates to interpolate c_{rms} over the region of interest, and discard the new point if its c_{rms} estimate differs too much from the interpolated value.
4. Stop when all of the points brighter than some threshold value have been scanned.

IV. Simulation Results

Figure 2 shows simulation results designed to validate the aberration correction principles proposed here. A £nite-difference code from the *Seismic Unix* package [13] simulates acoustic propagation through arbitrary media. The simulation domain is sampled at 30 μm, corresponding to λ/25 in soft tissues at the center frequency of 2 MHz. A map of the sound speed at every point in the domain serves as a "virtual phantom" and is constructed by a MATLAB program. An impulsive spherical wave is launched from the top of the domain, which has perfectly absorbing boundary conditions on all sides. During each simulation run, the up-coming re¤ections are recorded at 64 positions along a 15-mm aperture. By running the simulation 64 times with 64 source positions across the aperture, a complete dataset is obtained, approximating the data acquired with a real 64-element phased array. After these data are read into MATLAB, images are formed at many constant sound speeds using an ef£cient Fourier-domain algorithm, then combined into a single composite image using a c_{rms} map computed using the algorithm in Section III.

The model for a simulated brain imaging problem is shown at top left. In this simple approximation, the skull bone is modeled as a homogeneous medium with 2400 m/s sound speed. The background speed is 1500 m/s. The nine points in the upper row are two wavelengths apart. In the £rst image, focused assuming 1500 m/s sound speed, the targets are not cleanly resolved—we cannot even tell how many there are. At higher speeds approaching 2000 m/s, different targets come into focus at their corresponding rms sound speeds, indicating success of the hyperbolic travel-time approximation. In the corrected composite image, all of the targets are reasonably well-focused, even though the speed variation has a minor lateral component.

The second row of Figure 2 shows simulation results for a speckle-producing target with embedded scatter-free regions (cysts). The background wave speed is 1400 m/s with a Gaussian-shaped, high-speed anomaly above the target. In the conventional image, formed assuming 1500 m/s speed, the large upper cysts are poorly de£ned and the lower small cysts are almost invisible. In the corrected composite image, all of the cysts are much improved.

V. Conclusions

Aberration is a problem that still plagues medical ultrasound imaging, harming resolution, reducing its diagnostic abilities, and even preventing its use in certain parts of the body. Existing correction algorithms in the ultrasound literature are either impractical in clinical situations or have had only limited success, falling short of achieving diffraction-limited images. Concepts borrowed from a half-century of adaptive imaging experience in geophysics have the potential to enable an improvement in ultrasound resolution and inspire new directions for aberration correction research.

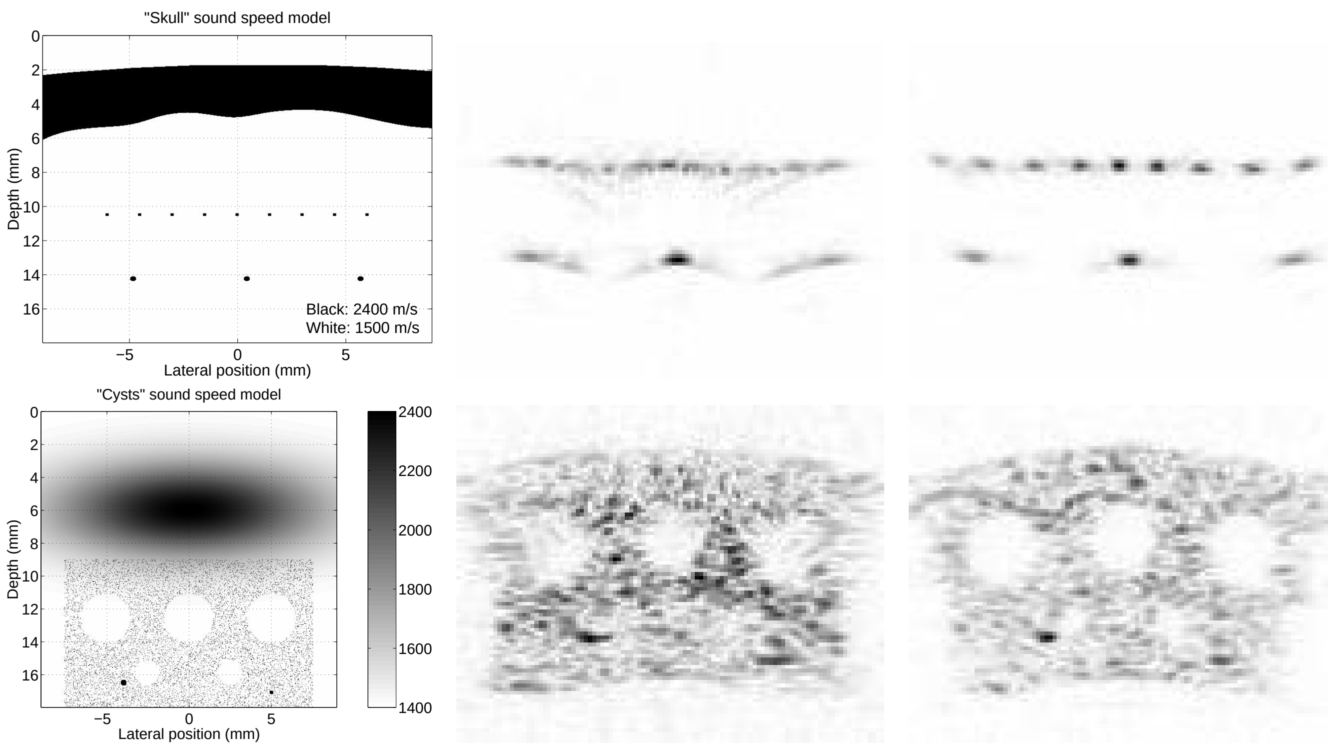

Figure 2: Top row: Simple skull model for £nite-difference simulations; conventional 1500 m/s image; corrected image using the algorithm described in the text. (Targets in the upper row are 2λ apart.) Bottom row: Cysts and speckle model; conventional 1500 m/s image; corrected image. (The small cysts are 2λ in diameter.)

VI. References

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