Adaptive Focusing Through Layered Media Using the Geophysical "Time Migration" Concept

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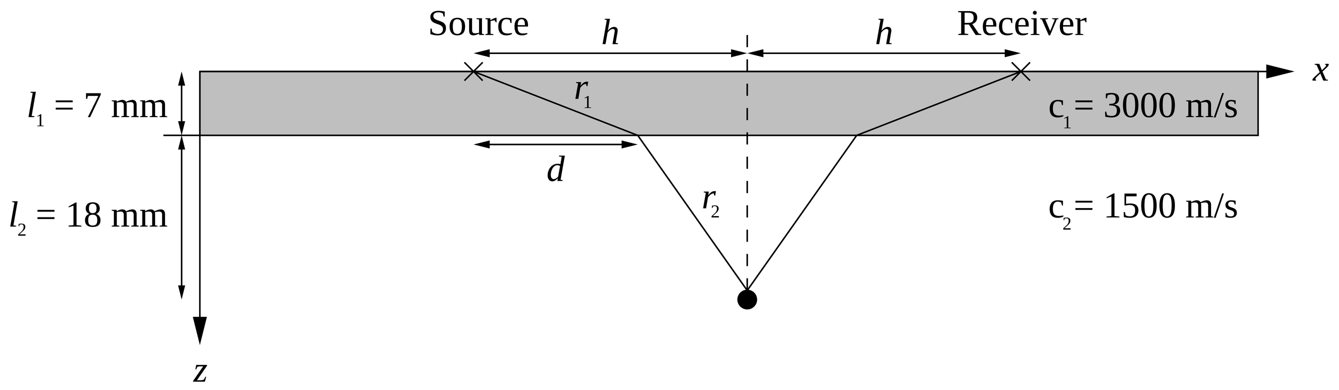
ABSTRACT

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 difficult. In many cases, an inability to estimate the screen parameters with sufficient 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. Reflection 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. (Supported by the University of Illinois Research Board)

1 Introduction

- For many scenarios in medical ultrasound imaging, aberration correction is still an open problem.
- Virtually all practical, published algorithms are based on a screen model of tissue aberration (extended via concept of *isoplanatic patches*). Hard to estimate time shifts given speckle targets and aberration.
- Our approach: Borrow concepts from geophysics (seismic imaging). Why? Much work has been done over the past century on adaptive acoustic imaging—the Earth is a highly inhomogeneous medium.
- *Time migration* describes a class of algorithms which estimate the unknown sound speed. (*Migration* is geophysics jargon for imaging.)
- Basic idea: If sound speed variation is layered ($c = c(z)$) or mostly so, each point in the image can be well-focused using some constant sound speed (assumption of straight ray-paths).
- Advantages: Only estimate one parameter per image point or [small] ROI, vs. N time shifts in screen methods; long history showing robustness to small lateral speed variations.
- Disadvantage: Does not handle multiple scattering.

2 Focusing through layered media



Echo travel time is $t = \frac{2}{c_1}\sqrt{l_1^2 + d^2} + \frac{2}{c_2}\sqrt{l_2^2 + (h-d)^2}$, where d is chosen to minimize t. Is there a general formula for t vs. h, given $\{c_i\}$ and $\{l_i\}$? Not really (even the two-layer case yields a non-trivial quartic equation).

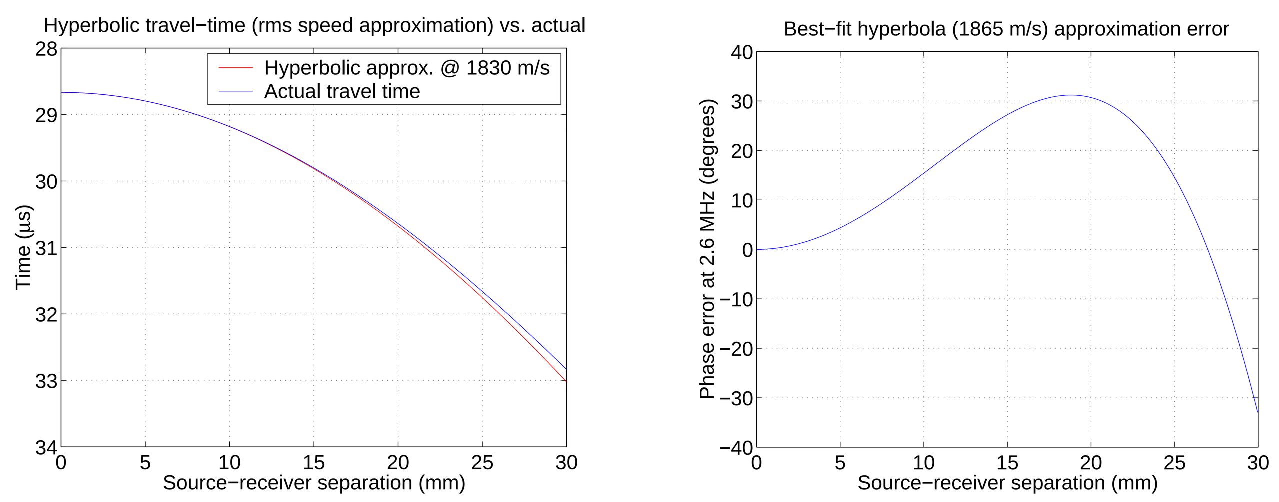
Taner/Koehler (*Geophysics* Dec. 1969): For any medium where $c = c(z)$,

$$t^2(h) = k_1 + k_2 h^2 + k_3 h^4 + k_4 h^6 + \cdots \qquad (1)$$

where $k_1 = t_0^2$ (squared vertical two-way time) and

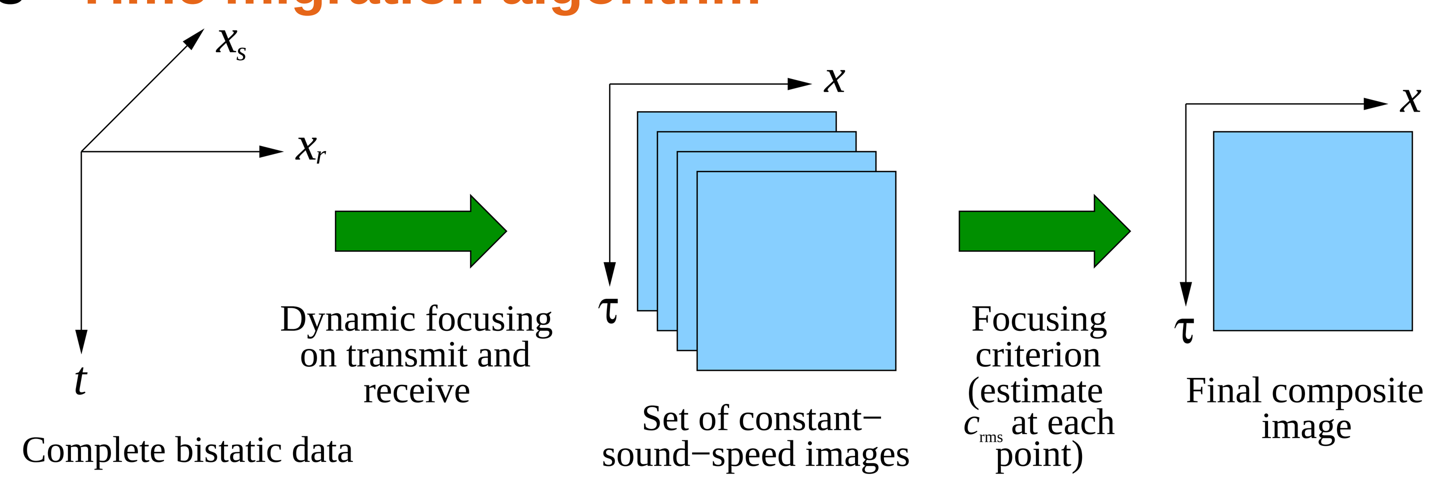
$$k_2 = \frac{4}{c_{\text{rms}}^2} \quad \text{where} \quad c_{\text{rms}}^2 = \frac{1}{t_0}\int_0^{t_0} c^2(t)\,dt. \qquad (2)$$

The second-order approximation is a *hyperbola*, identical to the travel-time hyperbola for a target in a homogeneous medium with $c = c_{\text{rms}}$.
$\Longrightarrow$ If the approximation holds, *we can focus targets in a $c(z)$ medium by assuming a constant-speed-of-sound, c_{rms}, in the imaging algorithm.* When the source-receiver separation is less than the target depth, the hyperbolic approximation is quite accurate. For the two-layer example:



We only need to estimate one parameter, c_{rms}, at each image location.

3 Time migration algorithm

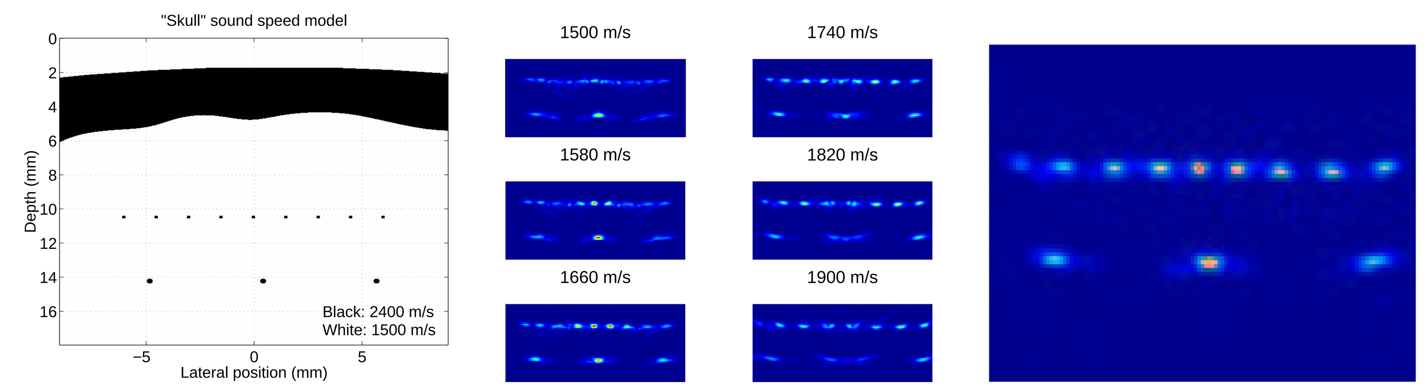


1. Form set of constant-speed, dynamically focused images.
2. Pick brightest point. Let c_{rms} = speed used to form its parent image.
3. Pick next-brightest point. If (dist. from other picked points) $> d_{\min}$, let c_{rms} = speed used to form its parent image, else skip.
4. Continue in decreasing order of brightness (subject to $d > d_{\min}$ from points already chosen). Discard if local c_{rms} estimate differs too much from c_{rms} interpolated from existing points.
5. Stop at some empirically chosen place along the brightness histogram.

4 Simulations

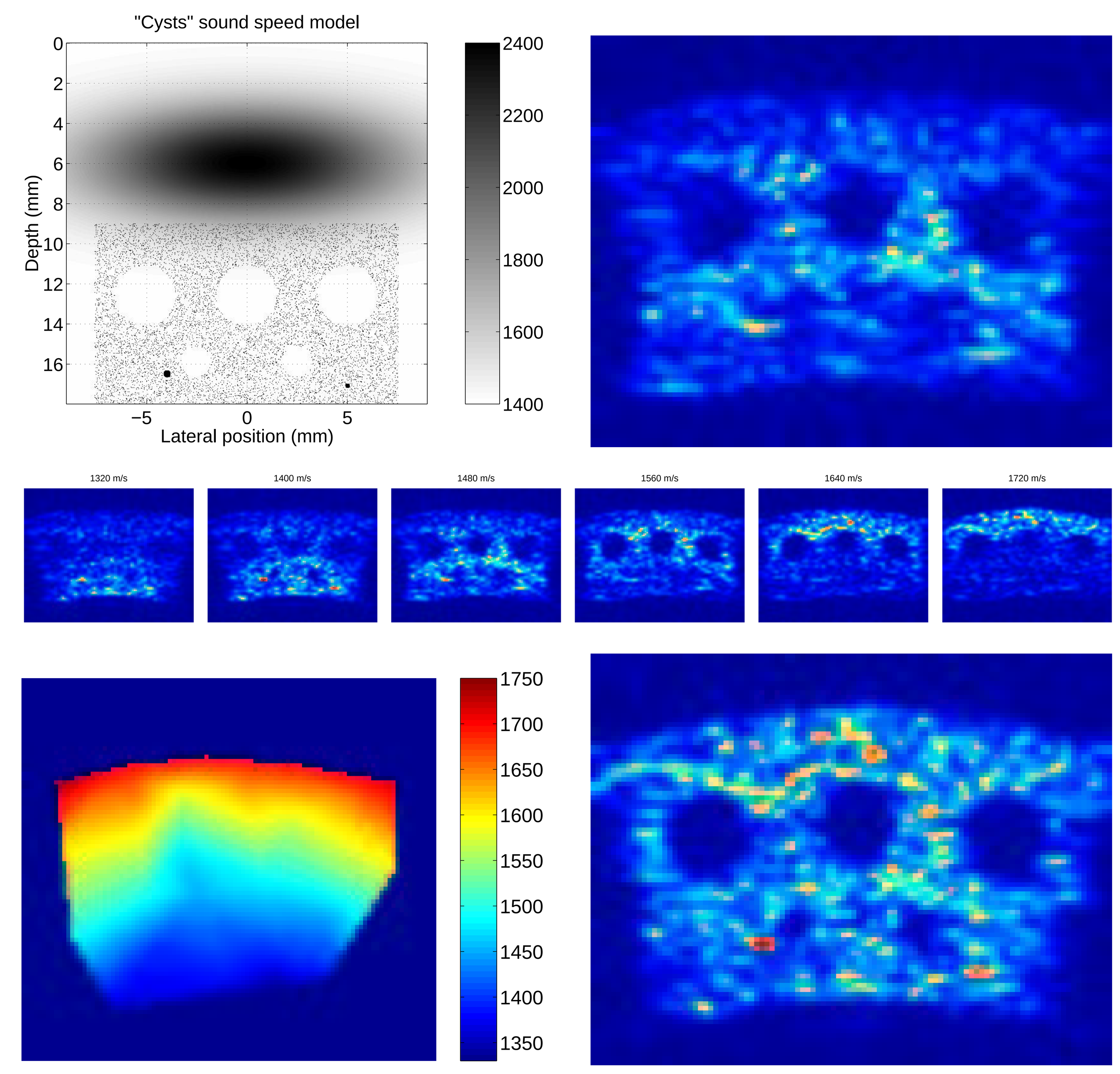
A finite-difference code was used to acquire complete, bistatic, synthetic data from a 64-element array at a center frequency of 2 MHz. After reading the data into MATLAB, images were formed at a range of assumed constant sound speeds using an efficient frequency-domain algorithm. The corrected, composite image was then formed as already described.

Simplified skull and point targets:



Left to right: Simple skull model for finite-difference simulations; conventional images formed assuming a constant speed of sound; corrected image using the algorithm described in the text. (Targets in the upper row are 2λ apart.)

Cysts and speckle:



Top row: Cysts and speckle model; conventional 1500 m/s image.
Middle row: Images formed at a range of constant sound speeds.
Bottom row: Derived rms sound speed map; corrected image.
(The small cysts are 2λ in diameter.)