Analysis of tissue changes, measurement system effects, and motion artifacts in echo decorrelation imaging

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Echo decorrelation imaging, a method for mapping ablation-induced ultrasound echo changes, is analyzed. Local echo decorrelation is shown to approximate the decoherence spectrum of tissue reflectivity. Effects of the ultrasound measurement system, echo signal windowing, electronic noise, and tissue motion on echo decorrelation images are determined theoretically, leading to a method for reduction of motion and noise artifacts. Theoretical analysis is validated by simulations and experiments. Simulated decoherence of the scattering medium was recovered with root-mean-square error less than 10% with accuracy dependent on the correlation window size. Motion-induced decorrelation measured in an *ex vivo* pubovisceral muscle model showed similar trends to theoretical motion-induced decorrelation for a 2.1 MHz curvilinear array with decorrelation approaching unity for 3–4 mm elevational displacement or 1–1.6 mm range displacement. For *in vivo* imaging of porcine liver by a 7 MHz linear array, theoretical decorrelation computed using image-based motion estimates correlated significantly with measured decorrelation ($r = 0.931$, $N = 10$). Echo decorrelation artifacts incurred during *in vivo* radiofrequency ablation in the same porcine liver were effectively compensated based on the theoretical echo decorrelation model and measured pre-treatment decorrelation. These results demonstrate the potential of echo decorrelation imaging for quantification of heat-induced changes to the scattering tissue medium during thermal ablation. © 2015 Acoustical Society of America. [<http://dx.doi.org/10.1121/1.4906580>]

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I. INTRODUCTION

Echo decorrelation imaging¹ is a recently developed pulse-echo imaging method that quantifies and maps ultrasound echo changes on millisecond time scales. The basis for this approach is the observation that during thermal ablation, tissue heating coincides with substantial, rapid fluctuations in echo signals. This observation has led to development of echo decorrelation imaging as a potential method for ablation monitoring.^{1,2} Echo decorrelation images are formed by spatially mapping millisecond-scale decorrelation of echo signals throughout the ultrasound image plane, quantifying echo changes as a function of position.

Echo decorrelation imaging has been proposed as a method for monitoring of radiofrequency ablation (RFA), a currently important clinical treatment modality for unresectable liver cancer^{3–5} and other tumors of soft tissue and bone.^{6,7} Echo decorrelation imaging may also be applicable to monitoring of other thermal ablation methods,^{8–10} including high-intensity focused ultrasound ablation.^{11–13} Preliminary research has confirmed the ability of echo decorrelation imaging to predict heat-induced tissue coagulation for RFA and ultrasound ablation, both *in vitro*^{1,14} and *in vivo*.² Most recently, echo decorrelation imaging has been shown by receiver operating characteristic (ROC) curve analysis to predict local thermal lesioning effectively for both *in vivo* RFA of swine liver and for ultrasound ablation

of rabbit liver with VX2 tumor (areas under the ROC curve 0.832 and 0.776, respectively).²

Like other ultrasound-based monitoring methods, echo decorrelation imaging potentially offers lower cost and complexity than temperature mapping by magnetic resonance imaging.^{13,15,16} The simple image formation process for echo decorrelation imaging, computationally comparable to processing already performed by modern ultrasound scanners, is also conducive to real-time implementation.¹ Thus echo decorrelation imaging is potentially useful for real-time ablation monitoring in many clinical ablation procedures where magnetic resonance imaging is too costly or cumbersome.

Echo decorrelation imaging also has potential advantages over other current ultrasound methods for real-time monitoring of thermal ablation. Several ultrasound ablation monitoring methods are based on estimation of echo signal delay, displacement, or strain by cross-correlation or similar analysis. These include echo strain imaging for monitoring temperature elevations,^{17–20} elastography,²¹ acoustic radiation force impulse imaging,²² and harmonic motion imaging.²³ However, echo signal decorrelation (or in the frequency domain, signal decoherence) fundamentally limits the accuracy of such correlation-based methods in estimating time delay, displacement, or strain.^{24–28} Thus echo decorrelation caused by tissue heating¹ or by heat-induced bubble activity¹⁴ can limit the accuracy of these correlation-based methods for real-time ablation monitoring. In contrast, echo decorrelation imaging relies on these same heat-induced signal changes as an indicator of local tissue ablation.

Other ultrasound approaches to ablation monitoring have been based on tissue echogenicity changes, including

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conventional B-mode imaging^{6,12} as well as quantitative tracking of echo energy changes (e.g., integrated backscatter)^{29–32} and statistical parameters of the backscattered ultrasound echoes.^{33,34} However, the local echogenicity of tissue is dependent on heat-induced changes in tissue attenuation, sound speed, and bubble activity, which do not consistently correspond with local tissue coagulation.^{1,34,35} Echo decorrelation imaging, which tracks stochastic changes in echo waveforms over short time scales (e.g., 10–50 ms),^{1,2} may potentially be a more robust predictor of ablation. Other ablation monitoring approaches have also tracked stochastic changes in echo signal amplitude or energy^{32–34} but over longer time scales (e.g., comparison of speckle statistics between frames temporally separated by 20 s),³³ so that these approaches may not be suitable for *in vivo* monitoring in the presence of tissue motion. Experimental studies of echo decorrelation imaging have indicated that, in comparisons using the same *ex vivo* and *in vivo* echo data, local tissue ablation was predicted with significantly greater accuracy using echo decorrelation than using integrated backscatter.^{1,2}

With further development, theoretical understanding, and validation, echo decorrelation imaging may be useful for real-time clinical ablation monitoring, with applications to image guidance and control for RFA and other thermal cancer therapies. Previous theoretical and experimental work has clarified how echo decorrelation is caused by translational tissue motion and electronic noise^{27,36–39} as well as tissue strain.^{40–42} However, for echo decorrelation to fulfill its potential as an ablation monitoring tool, greater understanding is needed on the relationship between mapped echo decorrelation and heat-induced changes to the scattering tissue medium, which may include bubble and vapor activity,^{14,35,43} microstructural changes,^{43–45} and thermal expansion,^{17,18} as well as heat-induced variations in acoustic properties such as sound speed^{17,18,46} and attenuation.^{46,47} In particular, further information is needed about how heating-induced decorrelation interacts with decorrelation caused by motion, strain, and noise. Although echo decorrelation imaging can effectively predict tissue ablation *in vivo* in the presence of significant motion, and motion effects can be mitigated by image gating,² motion- and noise-induced decorrelation may potentially mask some heat-induced decorrelation or cause false-positive prediction of local ablation. Effective compensation of motion and noise effects, made possible by analysis of how these decorrelation sources combine in a measured echo decorrelation image, would expand the utility of echo decorrelation imaging for clinical ablation monitoring.

In addition, methods are needed to quantify and interpret heat-induced tissue changes independent of the pulse-echo measurement system, similar to system-independent methods for tissue characterization from scattered ultrasound.^{48–50} This analysis would, after compensation of motion and noise effects, provide a quantitative map of heat-induced changes to the scattering tissue medium, which could then be used to detect and characterize heat-induced tissue damage. To interpret echo decorrelation images appropriately, effects of the measurement system on echo decorrelation image resolution and accuracy should be understood,

similar to previous characterization of measurement system effects in ultrasound scattering measurements.^{51–53}

This paper analyzes echo decorrelation imaging, elucidating effects of the pulse-echo measurement system, echo signal windowing, electronic noise, and tissue motion on echo decorrelation images. With measurement system effects accounted for, echo decorrelation is shown to map the local decoherence spectrum of the scattering medium at spatial frequencies proportional to the ultrasound pulse center frequency. Echo decorrelation induced by tissue motion is characterized in terms of transmit and receive beam patterns, leading to a method for reduction of motion and noise artifacts from echo decorrelation images. The theory is tested using simulations and analysis of *ex vivo* and *in vivo* experimental data. Simulation results confirm the theoretical relationship between echo decorrelation and scattering medium decoherence and provide insight into requirements for accurate echo decorrelation mapping. Decorrelation associated with translational tissue motion is shown to be accurately characterized by theory incorporating transmit and receive beam functions, while tissue strain introduces additional decorrelation. Compensation for artifacts due to tissue motion and electronic noise is shown to be feasible using both simulations and *in vivo* experimental data, enabling echo decorrelation images to accurately map intrinsic changes to the scattering tissue medium.

II. METHODS

A. Theory

Echo decorrelation imaging¹ is analyzed here using previously developed theory of scattering from weakly inhomogeneous tissue^{52,54} to determine the contributions of heat-induced tissue changes, tissue motion, electronic noise, and measurement system effects.

1. Definitions

An echo decorrelation image is analyzed as a representation of position-dependent changes between two ultrasound pulse-echo image frames $I_0(y, z)$ and $I_1(y, z)$ with y denoting the azimuth (array) direction and z denoting the range (depth) direction within the image plane. I_0 and I_1 are typically two sequential frames recorded at the same position during therapeutic tissue heating. Following the notation of a previous simulation study,⁵⁵ each image frame comprises a collection of complex pulse-echo signals with real and imaginary parts, either in radiofrequency (Hilbert-transformed) or baseband-demodulated (IQ) form. The starting point for echo decorrelation analysis is a position-dependent, zero-lag, windowed spatial cross-correlation between the two complex image frames, defined here as

$$R_{01}(y, z) = \iint w(y - y', z - z') I_0(y', z')^* I_1(y', z') dy' dz' \equiv \langle I_0(y, z)^* I_1(y, z) \rangle, \quad (1)$$

where the star denotes complex conjugation and the angle brackets denote convolution with a windowing function w

centered at the (azimuth, range) location (y, z) . Similarly, image autocorrelations are defined as $R_{00}(y, z) = \langle |I_0(y, z)|^2 \rangle$ and $R_{11}(y, z) = \langle |I_1(y, z)|^2 \rangle$, which can be regarded as maps of spatially integrated backscattered energy.¹

An echo decorrelation image, representing a position-dependent map of echo signal changes between the two image frames, is then given by a normalized correlation coefficient subtracted from unity,

$$\Delta(y, z) = 1 - \frac{|R_{01}(y, z)|^2}{R_{00}(y, z)R_{11}(y, z)}. \quad (2)$$

This definition was chosen for convenience of interpretation in the analysis in the following text in which the echo decorrelation image $\Delta(y, z)$ will be demonstrated to provide a map of the spatial-frequency, magnitude-squared decoherence of the scattering tissue medium. The definition of Eq. (2) relates to a previously defined echo decorrelation quantity¹ as $\sigma(y, z) \approx 1 - \sqrt{1 - \Delta(y, z)} \approx \Delta(y, z)/2$, so that published work employing this previous definition^{1,14} can also be interpreted quantitatively based on the analysis presented in the following text.

2. Contributors to echo decorrelation

To interpret the tissue changes depicted in echo decorrelation images, as well as the contribution of other factors, the echo decorrelation map defined by Eq. (2) is analyzed using an approximate representation of pulse-echo ultrasound images, derived previously.⁵⁵ This representation is based on scattering theory for a three-dimensional, linear, weakly scattering inhomogeneous medium⁵⁶ subjected to a narrow-band incident waveform of the form $a(t)e^{-i2\pi f_0 t}$, incorporating the effects of transmit and receive aperture beam patterns.⁵⁴ Under these assumptions, a complex image frame can be defined, within multiplicative constants, as

$$I_0(y, z) = \int_{V_0} \gamma(\mathbf{r}_0) e^{-2ik_0 z_0} \Lambda(\mathbf{r}_0, y, z) dV_0 + n(y, z), \quad (3)$$

where $\mathbf{r}_0$ is a coordinate within the three-dimensional (3D) scattering volume, γ is the position-dependent tissue reflectivity, k_0 is the wavenumber ω_0/c for the pulse radial center frequency ω_0 and the sound speed c , the additive term $n(y, z)$ represents uncorrelated, spatially varying electronic noise, and Λ is a slowly varying, position-dependent “system function” incorporating the influence of transmit and receive beam patterns and waveforms.^{52,54}

$$\Lambda(\mathbf{r}_0, y, z) \equiv p_E(x_0, y - y_0, z_0) p_D(x_0, y - y_0, z_0) \times a\left(\frac{2(z - z_0)}{c}\right) e^{2ik_0 z_0}. \quad (4)$$

In Eq. (4), p_E and p_D are complex beam patterns of the transmitting and receiving apertures, evaluated using Rayleigh integrals at the center frequency f_0 of the imaging pulse, and $a(t)$ is the complex envelope of the imaging pulse.⁵⁵ The system function Λ can be regarded as a generalized point-spread function that is also dependent on the pixel

location (y, z) within the pulse-echo image plane. The pulse-echo image model of Eqs. (3) and (4) thus incorporates diffraction effects more accurately than models based on convolutions of a position-independent point-spread function with the scattering medium.⁵⁵

Following previous theory for scattering from soft tissue,⁵² we take the tissue reflectivity γ to be a wide-sense stationary, delta-correlated random variable. The scattering media $\gamma_0(\mathbf{r}_0)$ and $\gamma_1(\mathbf{r}_1)$, comprising the tissue reflectivity imaged by frames I_0 and I_1 , are assumed to be identical except for any translational motion and small changes associated with tissue heating. The spatial autocorrelation of the tissue reflectivity, $R_{\gamma 01}$, is assumed to depend only on the spatial lag $\mathbf{r}' = |\mathbf{r}_1 - \mathbf{r}_0|$ and on the image coordinates (y, z) and is assumed small except near a peak at the position $\mathbf{r}' = \delta\mathbf{r} = (\delta x, \delta y, \delta z)$, corresponding to translational displacement of the tissue, relative to the imaging array, between the two frames. The other quantities considered, including the beam functions, transmit waveform shape, and electronic noise level, are assumed to vary negligibly over the inter-frame time (typically <50 ms) for the frames I_0 and I_1 . Under these assumptions, an expected value of the cross-correlation function from Eq. (1) can be written using Eq. (3) as

$$\begin{aligned} E[R_{01}(y, z)] &\approx \left\langle \int_{V_0} \int_{V'} R_{\gamma 01}(\mathbf{r}', y, z) e^{2ik_0 z'} \right. \\ &\quad \times \Lambda(\mathbf{r}_0, y, z)^* \Lambda(\mathbf{r}_0 + \mathbf{r}', y, z) dV_0 dV' \Big\rangle \\ &\approx \langle S_{\gamma 01}(2k_0 \mathbf{e}_z, y, z) \rangle R_\Lambda(\delta\mathbf{r}, y, z), \end{aligned} \quad (5)$$

where $E[\cdot]$ represents the expected value or ensemble average and R_Λ is the spatial autocorrelation of the system function Λ . The latter approximation in Eq. (5) assumes that the system function autocorrelation $R_\Lambda(\delta\mathbf{r}, y, z)$ is slowly varying with respect to the pixel location (y, z) , meaning that measurement system effects within a spatial window $w(y, z)$ are adequately described by a single pulse-echo point-spread function.

In Eq. (5), $S_{\gamma 01}(\mathbf{k}, y, z)$ is the position-dependent cross-spectrum of the two scattering medium realizations, equal by the Wiener-Khinchin theorem to the 3D spatial-frequency Fourier transform of the position-dependent cross-correlation $R_{\gamma 01}(\mathbf{r}', y, z)$ between the two scattering medium realizations γ_0 and γ_1 . Similar to previously derived Fourier relationships for tissue scattering,⁵⁷ this cross-spectrum is evaluated at the vector spatial frequency $\mathbf{k} = 2k_0 \mathbf{e}_z$, where $\mathbf{e}_z$ is the unit vector in the axial or range direction. Thus the windowed cross-correlation of the two images is equivalent to a product of the cross-spectrum of the tissue reflectivity functions γ_0 and γ_1 , evaluated at a spatial frequency determined by the imaging pulse, with a deterministic autocorrelation function of the system function Λ , which depends only on the transmit and receive beams, pulse waveform, and tissue displacement.

Similarly, the autocorrelation of the complex image $I_0(y, z)$ can be written in terms of the spatial-frequency auto-spectrum of the tissue reflectivity as

$$E[R_{00}(y, z)] \approx \langle S_{\gamma_{00}}(2k_0 \mathbf{e}_z, y, z) \rangle R_{\Lambda}(0, y, z) \times (1 + \text{SNR}(y, z)^{-1}), \text{ where}$$

$$\text{SNR}(y, z) \equiv \frac{E[R_{00}(y, z)]}{E[|n(y, z)|^2]}, \quad (6)$$

and likewise for the image $I_1(y, z)$.

For small decorrelations such that $\Delta(y, z) \ll 1$, the image autocorrelation (spatially smoothed integrated backscatter) functions R_{00} and R_{11} within a single image plane are nearly constant with time. Under this assumption, the expected value of the echo decorrelation image from Eq. (2) can be written in terms of the scattering medium's power spectra, the system function Λ , the tissue displacement $\delta \mathbf{r}$, and the signal-to-noise ratio (SNR) as

$$E[\Delta(y, z)] \approx 1 - \rho(\delta \mathbf{r}, y, z) \times \frac{\langle S_{\gamma_{01}}(2k_0 \mathbf{e}_z, y, z) \rangle^2}{\langle S_{\gamma_{00}}(2k_0 \mathbf{e}_z, y, z) \rangle \langle S_{\gamma_{11}}(2k_0 \mathbf{e}_z, y, z) \rangle}, \quad (7)$$

where

$$\rho(\delta \mathbf{r}, y, z) = \left| \frac{R_{\Lambda}(\delta \mathbf{r}, y, z)}{R_{\Lambda}(0, y, z)} \frac{\text{SNR}(y, z)}{1 + \text{SNR}(y, z)} \right|^2 \quad (8)$$

is the normalized position-dependent autocorrelation of the system function Λ , multiplied by a term comprising the influence of electronic noise.

If the tissue medium is unchanged and all decorrelation is due to translational tissue motion, Eq. (8) is equivalent to previous analyses of motion-induced echo decorrelation in which the echo signal correlation is equal to a spatial correlation of the imaging system's point-spread function,^{36–39} weighted by a correlation coefficient associated with electronic noise.²⁷ By incorporating the position-dependent system function of Eq. (4), Eq. (8) is equivalent to previous expressions derived for echo decorrelation, as a function of tissue or aperture translation, employing shift-variant point-spread functions.^{38,39}

Decorrelation associated with tissue strain or deformation can be incorporated into Eq. (8), given the assumption that the tissue reflectivity power spectrum S_{γ} varies slowly with spatial frequency, by applying the scaling theorem for Fourier transforms to the cross-spectrum term $\langle S_{\gamma_{01}}(2k_0 \mathbf{e}_z, y, z) \rangle$.² The result for small strains is equivalent to a multiplicative scale factor in the term $\rho(\delta \mathbf{r}, y, z)$, which becomes

$$\rho(\delta \mathbf{r}, y, z) = \left| \frac{R_{\Lambda}(\delta \mathbf{r}, y, z)}{R_{\Lambda}(0, y, z)} \frac{\text{SNR}(y, z)}{1 + \text{SNR}(y, z)} \frac{1}{1 + \epsilon(y, z)} \right|^2 \quad (9)$$

where $\epsilon(y, z)$ is the local strain. For small strains, this agrees with a previous theory relating echo signal decorrelation with tissue strain.^{40–42} For large strains exceeding several percent, previous simulation results have shown a highly variable relationship between decorrelation and strain.^{40,41} The influence of strain on echo decorrelation is depicted by Eq. (9), whether this strain is caused by tissue motion or by

heating (e.g., thermal expansion^{17,18} or echo strain induced by sound speed changes^{17,18,46}).

In the absence of tissue motion and electronic noise, $\rho(\delta \mathbf{r}, y, z) = 1$ and the echo decorrelation image from Eq. (7) becomes the magnitude-squared coherence between the two realizations of the scattering medium, computed using spatially windowed power spectra. If the tissue power spectrum is statistically stationary over the spatial extent of the windowing function $w(y, z)$, Eq. (7) simplifies further to become the spatial-frequency coherence spectrum⁵⁸ of the scattering medium, subtracted from unity,

$$E[\Delta(y, z)] \approx 1 - \frac{\langle S_{\gamma_{01}}(2k_0 \mathbf{e}_z, y, z) \rangle^2}{\langle S_{\gamma_{00}}(2k_0 \mathbf{e}_z, y, z) \rangle \langle S_{\gamma_{11}}(2k_0 \mathbf{e}_z, y, z) \rangle}. \quad (10)$$

Thus an echo decorrelation image provides a spatial map of decoherence in the scattering medium, caused by heating or other tissue changes. This result is consistent with previous analysis showing that the spatial-frequency coherence function between two ultrasound images is proportional to the spatial-frequency coherence between the two corresponding scattering media.²⁷ However, Eq. (10) more generally relates position-dependent, spatial-domain changes in pulse-echo images, mapped by echo decorrelation imaging, to corresponding tissue changes quantified by the position-dependent decoherence spectrum of the scattering medium.

3. Compensation for motion and noise effects

In the presence of tissue motion and electronic noise, the corrupted echo decorrelation image $\Delta(y, z)$ from Eq. (7) can be written as a linear combination of a decorrelation $\hat{\Delta}$, caused by tissue changes alone, with effects of motion and noise,

$$\Delta(y, z) \approx 1 - \rho(y, z)(1 - \hat{\Delta}(y, z)), \quad (11)$$

where $\hat{\Delta}$ is the decorrelation caused by changes to the scattering medium, equal to the right-hand side of Eq. (10).

Because the beam functions, transmit waveform shape, and SNR can readily be estimated or measured, $\rho(y, z)$ can be determined if the tissue displacement $\delta \mathbf{r}$ is known; however, this approach to compensation is limited by any temporal variations in these quantities (caused for example by heating-induced tissue attenuation changes). This analytic approach will also not correct any decorrelation caused by out-of-plane tissue motion or local tissue strain. Alternatively, this corrupting term can be measured from a control data set where decorrelation is entirely attributed to tissue motion and noise; this approach correctly incorporates out-of-plane motion and strain effects but requires that tissue motion and noise must be statistically similar for the control and experimental data sets. In either case, the echo decorrelation image associated solely with tissue changes can be recovered by solving arithmetically for $\hat{\Delta}(y, z)$ in Eq. (12),

$$\hat{\Delta}(y, z) \approx \frac{\Delta(y, z) - \Delta_{\rho}(y, z)}{1 - \Delta_{\rho}(y, z)}, \quad (12)$$

where

$$\Delta_p = 1 - \rho(y, z) \quad (13)$$

is the decorrelation associated with tissue motion and noise, with $\rho(y, z)$ specified by Eq. (8).

B. Simulations

1. Image simulation methods

Complex pulse-echo images were simulated by a method⁵⁵ that implements Eq. (3) by 3D numerical integration of model tissue reflectivity functions multiplied by the system function from Eq. (4). Three-dimensional tissue reflectivity functions were simulated by random media γ_0 and γ_1 , each modeled as a Gaussian white noise process (i.e., normally distributed random variables), with decoherence caused both by localized random perturbation (simulating stochastic tissue changes caused by heating) and translational displacement (simulating tissue motion),

$$\begin{aligned} \gamma_0(\mathbf{r}) &= n_0(\mathbf{r}) \\ \gamma_1(\mathbf{r}) &= \gamma_0(\hat{\mathbf{r}}) + \xi(\hat{\mathbf{r}})n_1(\hat{\mathbf{r}}), \end{aligned} \quad (14)$$

where $n_0(\mathbf{r})$ and $n_1(\mathbf{r})$ are different realizations of a normally distributed, zero-mean, unity-variance Gaussian white noise process, $\hat{\mathbf{r}} = \mathbf{r} + \delta\mathbf{r}$ is the displaced position of the medium realization γ_1 , and $\xi(\mathbf{r})$ is a cylindrically symmetric region of decoherence centered at the image coordinates (y_1, z_1) ,

$$\xi(\mathbf{r}) = \xi_0 e^{-[(y-y_1)^2 + (z-z_1)^2]/(2\alpha^2)}. \quad (15)$$

Echo decorrelation images were computed between a base reference image I_0 of the original medium γ_0 and images I_1 of perturbed media γ_1 . In these simulations, the pulse center frequency was $f_0 = 5$ MHz, and the pulse envelope was the Gaussian function $a(t) = e^{-t^2/(2\sigma_t^2)}$ with $\sigma_t = 0.125 \mu\text{s}$. The simulated linear array transducer had an elevation width of 5 mm and a fixed elevation focal distance of 25 mm. The scattering medium had a sound speed of 1.5 mm/ μs and an attenuation of 0.75 dB/cm/MHz, within the typical range for human soft tissues and tissue-mimicking phantoms.⁵⁹ The medium reflectivity was specified on a $76 \times 300 \times 651$ grid spanning 7.5 mm in the elevation direction ($\delta x = 0.1$ mm), 30 mm in the azimuth direction ($\delta y = 0.1$ mm), and 40 mm in the range direction ($\delta z = 0.06$ mm). Transmit and receive beams were computed by a Fresnel approximation⁶⁰ using continuous transmit and receive focusing with a constant f-number of 1.5.

2. Tests of decoherence estimation and artifact compensation

To assess the accuracy of decorrelation imaging, the perturbed media included spatial decoherence given by Eq. (15) with $\xi_0 = 0.3$ and $\alpha = 2, 3, 4$, or 5 mm. Echo decorrelation images were then obtained using Eq. (2) and ensemble averaged over ten medium realizations. The spatial windowing of Eq. (1) was performed using the Gaussian window

$$w(y, z) = e^{-(y^2 + z^2)/(2\sigma^2)}. \quad (16)$$

Although echo decorrelation maps do not depend strongly on the window shape chosen, the isotropic Gaussian of Eq. (16) has been found to provide uniform decorrelation maps suitable for comparison with ablated tissue histology, while also allowing the 2D convolution operation within Eq. (1) to be efficiently implemented in either the spatial or spatial-frequency domain.

Echo decorrelation maps were compared to the actual decoherence of the tissue spectrum, equal to $\xi(y, z)^2$. Accuracy was assessed based on the root-mean-square (RMS) error, defined as the RMS difference between the echo decorrelation and the actual decoherence, evaluated over the entire image plane. Because the size of correlation analysis windows is known to affect computed correlation coefficients, depending on the relative length scales of the analyzed signal variations,^{41,42} this quantitative comparison was performed for a range of window sizes $\sigma = 0.1$ –3 mm for each case.

Compensation of noise effects and known motion artifacts was tested by arithmetically solving for the uncorrupted echo decorrelation $\hat{\Delta}(y, z)$ using Eq. (12). A lesion size $\alpha = 3$ mm and spatial window size $\sigma = 0.5$ mm were chosen for illustration in these simulations. Simulated electronic noise was introduced by adding Gaussian white noise to the simulated complex echo signals before application of simulated time-gain compensation (TGC), resulting in an average SNR of 20 dB. Due to tissue attenuation as well as transmit and receive beam focusing, the SNR is highest at the depth corresponding to the array's elevation focal length and decreases with greater depths due to amplification of the noise by TGC. To test compensation of combined motion artifacts and SNR degradation, the simulated tissue was also translated by the vector displacement $\delta\mathbf{r} = (0.05, 0.05, 0.025)$ mm. The corresponding echo decorrelation map was then computed directly from the simulated echo data, with and without tissue motion, using Eqs. (1) and (2). Artifact compensation was implemented using Eqs. (8) and (12) with the known displacement, system functions computed from Eq. (4) using the Fresnel approximation, and local SNR computed from Eq. (6).

To separately assess the impact of electronic noise and displacement in each direction on echo decorrelation images, RMS error was computed between compensated and uncompensated echo decorrelation images and the known medium decoherence ($\xi_0 = 0.3$, $\alpha = 3$ mm, for a maximum decoherence of 0.09). Error was computed for average SNR values ranging from 1 to 40 dB in a stationary medium. Error was then computed as a function of separate displacements up to 0.4 mm in the elevation (x) direction, 0.2 mm in the array (y) direction, and 0.08 mm in the range (z) direction, in the absence of electronic noise.

To assess the effect of the correlation window size on accuracy of echo decorrelation imaging for cases without tissue motion, echo decorrelation images were computed for simulated lesion size parameters [Eq. (15)] $\alpha = 2, 3, 4$, and 5 mm and spatial window size parameters [Eq. (16)] $\sigma = 0.1$ –3.0 mm. RMS error was computed as a function of window size for each simulated lesion size.

C. Experiments

1. Motion-induced decorrelation *ex vivo*

To validate the theory of echo decorrelation caused by tissue motion, complex in-phase and quadrature (IQ) images and B-scan images of an *ex vivo* model for human pubovisceral muscle⁶¹ were obtained as a function of tissue displacement and strain. This model allowed pulse-echo image data to be recorded before and after well-controlled motion in the range and elevation directions, spanning the range of displacements of interest for *in vivo* tissue ablation.

Images of a fresh bovine skeletal muscle specimen were obtained using a Zonare (Mountain View, CA) z.one scanner with a C4-1 probe, a 64-element, 31.84 mm curvilinear array with a nominal center frequency of 2.1 MHz. The muscle tissue was immersed in a saline water-filled tank and wrapped in a sling around a gelatin indenter as shown in Fig. 1. The indenter was constructed with an *n*-propanol and formaldehyde formulation with a gelatin base to mimic the sound speed and stiffness of human soft tissue. Sound speed and elastic modulus of the indenter material were measured to be respectively 1543 ± 4 m/s and 15 ± 4 MPa.⁶² The array was held 5–6 cm below the muscle.

Tissue motion in the range direction (δz) was imposed by displacing the indenter toward the array from 0 to 1.6 mm in 0.2-mm increments using a stepping positioner system (Velmex, Bloomfield, NY) for both a no-strain case and a strain case as illustrated in Fig. 1. In the no-strain case, the muscle tissue ends were clamped to the pole attached to the indenter, ensuring rigid body displacement with negligible strain in either the indenter or the tissue. In the strain case, the tissue ends were clamped to the stationary sides of the tank, so that the muscle tissue would be displaced

comparably to human soft tissues *in vivo*. Out-of-plane motion (δx) was imposed by translation of the array 0–4 mm in 0.4-mm increments along the elevation direction, controlled by the stepping positioner system.

To determine whether the gelatin indenter incurred substantial deformation from water absorption or other causes, the apparatus was photographed from the same position, before and after the indentation series, in nine equivalent experiments. Distances between landmarks on opposite sides of the indenter were measured manually in MATLAB and used to compute the percent change in indenter dimensions.

For each indenter displacement and each elevation position, parallel, 2D, complex (in-phase and quadrature components, or IQ) image frames were acquired by the scanner. Echo decorrelation was analyzed in a region of interest (ROI) within the muscle in the IQ images, chosen as close to the muscle midline as possible. Care was taken in segmenting the ROI to avoid bright areas associated with reflections from any wires near the tissue and reverberations from the indenter mounting rod. A representative B-scan image is shown in Fig. 2 with the analyzed ROI marked by a superimposed rectangle.

Echo decorrelation images of the muscle specimen were computed using Eq. (2), spatially averaged within the ROI, and ensemble averaged. For analysis of echo decorrelation as a function of elevational (out-of-plane) displacement, averages were taken over all possible pairs of image planes matching an elevational displacement of δx from the first eight elevation positions. For analysis of echo decorrelation as a function of range (indentation direction) displacement, averages were taken over all pairs of image planes matching a range displacement δz from the first five indentation positions. The spatial window employed was given by Eq. (16) with $\sigma = 2$ mm, sufficient in size to spatially smooth stochastic fluctuations in heat-induced echo decorrelation while maintaining acceptable spatial resolution.

For the strain case, local engineering strain within the muscle tissue was measured using a Bayesian maximum likelihood estimator^{61,63,64} that computes rigid-body

FIG. 1. (Color online) Diagrams and photos of experimental setup for the no-strain configuration [(a) and (b)] and strain configuration [(c) and (d)] showing the indenter, bovine skeletal muscle, mounting apparatus, and imaging transducer position.

FIG. 2. A representative cross-sectional B-scan from the acquired volumetric image of bovine muscle tissue modeling the human pubovisceral muscle. The ROI used for echo decorrelation computations is shown by a superimposed rectangle.

displacement of small regions of interest within an image. Displacements in the B-scan data at the four corners of the ROI shown in Fig. 2 were tracked within both the first and the last elevational slices over eight indentations. Engineering strain was computed diagonally across the ROI in the two elevational slices as the normalized change in distance between co-planar diagonal points, resulting in four strain measurements used to compute a RMS value of strain at each indentation step.

Experimental decorrelation values were compared to theoretical values caused by tissue motion, computed using Eqs. (8) and (9), with the measurement system function Λ computed using the Fresnel approximation.⁶⁰ Transmit and receive beams were computed with a continuous f -number of 2 for a transducer elevation width of 15 mm and elevation focal depth of 80 mm, approximated as a linear aperture for simplicity. Based on a fit to measured pulse-echo signal spectra, the pulse center frequency was 2.19 MHz and the temporal width of the Gaussian transmit pulse was $\sigma_t = 0.283 \mu\text{s}$. For the no-strain case, the system function autocorrelation from Eq. (8) was computed for the translational range and elevation displacements imposed by the positioner system. For the strain case, the measured RMS engineering strain for each indentation step was used in Eq. (9) to compute the corresponding theoretical decorrelation. Displacements in the elevation direction, because they were accomplished by translation of the transducer array, caused no tissue strain.

Decorrelation values for the strain and no-strain cases were statistically compared using a paired t test performed in MATLAB. To account for repeated measurements from the overlapping ultrasound beams, the computed t statistic and p value were adjusted for the estimated number of independent decorrelation measurements, using a scale factor equal to the product of the beam width divided by the elevational step displacement and the pulse length divided by the range step displacement.

2. Motion-induced decorrelation in vivo

To test the inception and compensation of motion artifacts *in vivo*, image data from a previous *in vivo* study of RFA in swine liver² were analyzed. As described previously,² RFA was performed in an open surgical procedure with simultaneous ultrasound imaging. All animal experiments were performed according to protocols approved by the University of Cincinnati Institutional Animal Care and Use Committee (IACUC). Pulse-echo liver images were obtained using the Iris 2 ultrasound imaging system with a 7 MHz, 192 element linear ultrasound array (L7, Guided Therapy Systems, Mesa, AZ). Beamformed echo signals from the Iris system were recorded by a 14-bit A/D converter (Compuscope VS 14200, Gage Applied Technologies, Montreal, Quebec, Canada) at a sampling rate of 33.3 MHz. Image frame pairs, consisting of two pulse-echo image frames recorded 20 ms apart, were acquired once every 0.85 s. For these experiments, echo decorrelation images were computed using a variation of Eq. (2),

$$\Delta(y, z) = 2 \frac{R_{00}(y, z)R_{11}(y, z) - |R_{01}(y, z)|^2}{R_{00}(y, z)R_{11}(y, z) + \overline{R_{00}R_{11}}}, \quad (17)$$

where $\overline{R_{00}R_{11}}$ denotes a spatial average of the autocorrelation product $R_{00}(y, z)R_{11}(y, z)$ over the entire image plane. This additional normalization factor helps to avoid anomalously large echo decorrelation values within hypoechoic image regions.¹ Further details, including comparison of echo decorrelation images with ablated tissue histology and statistical testing of lesion prediction over five RFA trials, are provided in a previous publication.²

Pulse-echo image data recorded during one of these experiments were analyzed to verify the present theory of motion-induced decorrelation. Ten image frame pairs were analyzed from a sham data set, recorded before radiofrequency ablation was performed at the same location. Each frame pair was manually chosen to represent approximately the same tissue position within two sequential respiratory motion cycles so that the interval between the paired frames was 3.5–4.5 s. Echo decorrelation was then computed between these paired images using Eq. (2). To allow precise comparison with translational tissue motion estimated for a small ROI, a Gaussian spatial window [Eq. (16)] with $\sigma = 1 \text{ mm}$ was employed.

Computed echo decorrelation values were then compared with theoretical values estimated from measured in-plane tissue motion and theoretical beam functions for the imaging system employed. For each image pair used in the echo decorrelation computations, the in-plane tissue displacement between the two frames was determined using the Bayesian maximum likelihood estimator^{63,64} described previously. B-mode (logarithmically scaled amplitude envelope) images derived from the acquired echo data were used for this displacement estimation, while the full radiofrequency echo signals were used for computation of echo decorrelation images. Two-dimensional in-plane displacement was computed for a set of seven 0.49×0.90 (azimuth $\times$ range) mm^2 ROIs centered at 11.5 mm in depth (z) and at seven azimuthal (y) positions, equally spaced between $y = -16.94$ and $y = -3.74 \text{ mm}$ (20th through 80th scan lines of the 192-line pulse-echo image). Engineering strain was computed between the ROIs centered at -14.74 and -3.74 mm in azimuth as the difference in displacement between these two ROIs, normalized by the initial distance between them. Because the imaging array was aligned approximately parallel to the direction of respiratory motion, tissue displacement was primarily within the image (y - z) plane.

Theoretical echo decorrelation values based on the measured displacements were then calculated, neglecting electronic noise effects. The beam function Λ was computed using a Fresnel approximation,^{55,60} assuming an elevation width of 7 mm, a fixed elevation focus of 25 mm, an electronic transmit focus at 27.4 mm depth, and continuous receive focusing at a f -number of 2.3. The pulse center frequency $f_0 = 7.1 \text{ MHz}$ and Gaussian pulse width $\sigma_t = 0.15 \mu\text{s}$ were determined by fitting the measured pulse-echo power spectrum to a Gaussian function. Theoretical echo decorrelation values were then computed from Eqs. (8) and (13) based

on the spatial autocorrelation of the simulated beam function, evaluated at a displacement δr given by the measured displacement for each ROI. Means and standard deviations of the theoretical echo decorrelation values for the seven ROI center positions were then compared with the corresponding means and standard deviations of the measured echo decorrelation at the same positions.

To characterize liver deformation throughout the respiratory cycle, displacements and engineering strains were also computed between all adjacent frames within each respiratory cycle. Displacements relative to the first frame of each pair used in the echo decorrelation computations were then obtained by summation of individual inter-frame displacements for the remainder of that respiratory cycle. Engineering strains throughout the respiratory cycle were computed based on these displacements, and RMS values of displacements and engineering strains were then computed.

3. Artifact compensation in vivo

To test compensation of artifacts caused by tissue motion and electronic noise, echo decorrelation images recorded during RFA (Ref. 2) were compensated using measured echo decorrelation images from a sham data set recorded at the same position. The image data set for RFA comprised 115 pulse-echo image frame pairs, spanning about 20 s pre-treatment, 40 s RFA treatment, and 38 s post-treatment. Spatial windowing was employed using Eq. (16) with $\sigma = 2$ mm. Echo decorrelation maps were temporally smoothed using a running average¹ of the decorrelation map $\Delta(y, z)$,

$$\Delta(y, z, t_i) = (1 - \epsilon) \Delta(y, z, t_{i-1}) + \epsilon \Delta(y, z, t_i), \quad (18)$$

with an averaging parameter $\epsilon = 0.05$. Cumulative echo decorrelation images were then formed by mapping the current temporal maximum decorrelation for each pixel location.

Cumulative echo decorrelation images were computed from the ablation data set for frame pairs numbered 30 (after 20 s pre-treatment and 5 s treatment), 50 (after 22 s treatment), 70 (after 39 s treatment), and 110 (33 s post-treatment). These echo decorrelation images were then compensated using Eq. (12), with the artifactual decorrelation Δ_p assumed equal to the cumulative decorrelation for corresponding frame pairs of the sham data set. This approach assumes that the respiratory tissue motion pattern, and thus its contribution to the temporally averaged echo decorrelation, is approximately constant with time.

III. RESULTS

A. Simulated decoherence estimation and artifact compensation

Representative simulated echo decorrelation images of a random medium with a local region of decoherence [Eqs. (14)–(15)], computed from simulated echo data, are shown in Fig. 3. Shown are the original decorrelation map for the medium without tissue motion and with 20 dB SNR (RMS error 0.0265), a motion-corrupted decorrelation map

FIG. 3. (Color online) Simulated echo decorrelation images of a random medium with a lesion represented by a region of decoherence with average SNR of 20 dB. Top left: Logarithmically scaled decorrelation map without tissue motion and with 20 dB SNR. Top right: Decorrelation map corrupted by tissue motion with 20 dB SNR. Center left: Decorrelation map after compensation for tissue motion and noise. Center right: Actual decoherence of the medium. Bottom: Cross sections through echo decorrelation images of the simulated lesion for all four cases, along the segment at azimuth 0 mm spanning the range 10–30 mm.

after displacement of magnitude 0.074 mm and 20 dB SNR (RMS error 0.0968), and a decorrelation map compensated for motion and noise using Eq. (12) (RMS error 5.4×10^{-3}). For comparison, an image of the actual decoherence of the scattering medium is also provided, and cross sections through all four images are plotted. The echo decorrelation image without tissue motion is seen to resemble the actual decoherence of the scattering medium except for the additional decoherence caused by electronic noise. Artifact compensation using the known position-dependent SNR and the computed autocorrelation of the measurement system function is effective, substantially recovering the original echo decorrelation image due to scattering medium decoherence

alone and increasing the echo decorrelation contrast by two orders of magnitude.

Representative RMS error in reconstructed echo decorrelation images, comparing the compensated image to actual decoherence, is shown in Fig. 4 for SNR up to 40 dB. The RMS error after motion compensation is also shown in Fig. 4 as a function of displacements along each axis. For comparable displacements, motion in the range (z) direction caused the greatest reconstruction error, followed by the azimuthal (y) direction and the elevation (x) direction. These trends are consistent with transmit and receive beam dimensions for the case simulated in which the pulse-echo image resolution was greatest in the z direction and smallest in the x direction.

Accuracy of echo decorrelation imaging in mapping the actual scattering medium decoherence is illustrated in Fig. 5 for four different lesion sizes as a function of the correlation window size parameter σ . This figure illustrates that the scattering medium decoherence can be estimated with reasonable accuracy (RMS error <10%) over a wide range of correlation window sizes with accuracy dependent on the relative length scales of the correlation window employed and the region of decoherence. A tradeoff is evident between spatial resolution, quantitative precision, and spatial smoothness of the decorrelation map. For very small window dimensions σ , the normalized RMS error between mapped decorrelation and the true decoherence is large due to statistical fluctuations in the computed echo decorrelation, and for large σ , due to spatial blurring of the simulated lesions. For large window dimensions, approaching or exceeding the length scales of the region of decoherence, the normalized RMS error is large due to excessive smoothing of the decorrelation map, reducing its spatial resolution. For each lesion size investigated, the minimum RMS error occurs for a correlation window width σ on the order of one-sixth the lesion width α , with relatively small error occurring up to window

FIG. 4. (Color online) RMS error between echo decorrelation and actual scattering medium decoherence for simulations with varying displacements and SNR values with and without compensation. Top left: RMS error vs SNR for a motionless medium. Remaining panels: RMS error vs displacement in the elevation (x), azimuth (y), and range (z) directions for noiseless echo data.

FIG. 5. (Color online) Normalized RMS error between simulated echo decorrelation maps and actual decoherence for four lesion size parameters α as a function of the correlation window size parameter σ .

widths of about one-half the lesion width. For comparison, the spatial resolution limit of the echo decorrelation image can be estimated from the -3 dB ($r^2=0.5$) autocorrelation width of a 2D Gaussian window, equal to 1.67σ .

B. Motion-induced decorrelation *ex vivo*

For the *ex vivo* experiment testing echo decorrelation as a function of tissue displacement in a pubovisceral muscle model, measured echo decorrelation values, averaged within the ROI shown in Fig. 2, are plotted in Fig. 6 as a function of tissue displacement in the range and elevation directions. Also plotted for comparison are theoretical echo decorrelation values based on decorrelation of the measurement system Λ after translational motion of the scattering medium, with and without additional strain. Measured and theoretical decorrelation values show similar trends with echo

FIG. 6. (Color online) Measured and theoretical echo decorrelation caused by tissue motion in an *ex vivo* model of the human pubovisceral muscle. The theoretical decorrelation is shown with a solid line with open circle markers for the no strain case in both panels and is shown with a dashed line with star markers for the strain case in the right panel. Also in both panels is the measured decorrelation (mean $\pm$ standard deviation) shown by a dashed-dotted line with triangle markers for the no-strain case and by a dashed-dotted line with inverted triangle markers for the strain case. Left: Echo decorrelation caused by motion in the elevation (x) direction. Right: Echo decorrelation caused by tissue indentation in the range (z) direction.

decorrelation increasing rapidly between 0–2 mm of elevation displacement or 0–0.5 mm of range displacement and approaching 1 (complete decorrelation) for 3–4 mm of elevation displacement or 1–1.6 mm of range displacement. Discrepancies are likely due in part to errors in the assumed system (beam) function Λ .

In the region of interest, RMS engineering strain ranged from 0 to 5.1% over eight indentation steps in which the tissue was displaced 0–1.6 mm in the range direction. The high variance of echo decorrelation for tissue undergoing strain, evident in the second panel of Fig. 6, is consistent with previous results showing that tissue strain introduces decorrelation with large variability.⁴⁰ Measured decorrelation values for the strain and no-strain cases were significantly different both for displacements in the elevation direction ($t = 2.648$, $p = 0.0064$) and in the range direction ($t = -2.112$, $p = 0.0295$). However, the theoretical decorrelation for displacements in the range direction was very similar with or without strain (decorrelation difference < 0.01).

Based on the analysis of photographs before and after the indentation sequence from nine equivalent indentation experiments, the indenter dimensions were found to change by an average of 0.12%. Thus any deformation of the indenter was small compared to deformation of the tissue.

C. Motion-induced echo decorrelation *in vivo*

For the *in vivo* experiment testing echo decorrelation imaging of porcine liver in the presence of tissue motion, the means and standard deviations of measured and theoretical motion-induced decorrelation for 10 image pairs are plotted in Fig. 7. The plot of measured vs theoretical decorrelation values shows reasonable agreement with a highly significant correlation coefficient ($r = 0.931$). The RMS tissue displacement computed by texture correlation between frame pairs used for the echo decorrelation computation was 0.23 mm in azimuth (δy) and 0.10 mm in range (δz). The RMS engineering strain of the tissue for these sequential frame pairs was

FIG. 7. (Color online) Measured vs theoretical motion-induced echo decorrelation for paired pulse-echo images of swine liver *in vivo*. The mean and standard deviation of the decorrelation values from each measured and theoretical echo decorrelation image are shown.

0.6%. For comparison, the overall RMS displacement, computed between the initial frame in the respiratory cycle and all subsequent frames of the cycle, was 0.70 mm in azimuth (δy) and 0.73 mm in range (δz). The corresponding overall RMS engineering strain was 7.6%.

D. Artifact compensation *in vivo*

Results from a test of motion and noise compensation for echo decorrelation imaging of *in vivo* thermal ablation are shown in Fig. 8. The uncompensated image shows larger echo decorrelation near the tip of the RFA probe, where tissue heating was occurring. This region of elevated echo decorrelation also corresponds to the region of coagulative necrosis confirmed by histologic examination.² However, the uncompensated image also shows substantial artifactual echo decorrelation, comparable in magnitude to the

FIG. 8. (Color online) Echo decorrelation images of swine liver undergoing RFA *in vivo*. Each panel shows a maximum-amplitude projection of the log-scaled echo decorrelation during the ablation treatment, superimposed on the corresponding B-scan image. Left: Uncompensated images for frames 30 (early in treatment), 50 (mid-treatment), 70 (late in treatment), and 110 (post-treatment). Right: Images compensated based on motion- and noise-induced decorrelation estimated from a sham data set taken before the ablation treatment. The RFA probe tip is indicated by the dashed circle in the upper left panel.

decorrelation caused by tissue heating. In the compensated image, computed using Eq. (12) with the motion- and noise-induced decorrelation estimated from a sham data set without thermal ablation, a large portion of these artifacts are suppressed so that the compensated image more accurately depicts the tissue changes caused by thermal ablation.

IV. DISCUSSION

A. Estimation of heat-induced tissue changes

Echo decorrelation imaging has potential application in monitoring of tissue ablation.^{1,2,14} Mapped echo decorrelation can effectively predict local tissue ablation as confirmed by comparison with ablated tissue histology after both *ex vivo*¹ and *in vivo*² treatments. Mapping of scattering medium changes by echo decorrelation imaging is thus a potential means for real-time ablation monitoring to ensure adequate ablation while limiting overexposure.

In this paper, analysis of echo decorrelation imaging has elucidated relationships among measured echo decorrelation, local changes in intrinsic tissue properties, tissue motion, and measurement system effects. The local echo decorrelation was found to approximate a decoherence spectrum of the local tissue reflectivity. Similar to previous studies of tissue characterization from ultrasonic scattering measurements,^{51,52} effects of the measurement system were analyzed in terms of a system function Λ that depends on the transmit and receive beam patterns as well as the interrogating ultrasound pulse. These measurement system effects are dependent on the spatial autocorrelation of the system function. In the absence of tissue motion, effects of the measurement system cancel out to good approximation. Thus unlike quantitative measurements of tissue scattering cross section,^{48–50} no additional normalization of measurement system effects is needed for quantitative mapping of the tissue decoherence spectrum through echo decorrelation imaging.

Because the local echo decorrelation has been shown here to approximate the decoherence spectrum of the tissue reflectivity, echo decorrelation imaging also provides a tool to quantify transient, heat-induced changes in the scattering tissue medium over short time scales. These changes likely include transient gas activity.^{14,43} In addition, temperature-dependent structural changes occur in tissue during thermal ablation, including cellular swelling and microvascular changes⁴⁵ as well as denaturation of collagen and other proteins^{65,66} and microstructural tissue damage due to vaporization.^{43,44} These physical changes are associated with temperature-dependent variations in tissue bulk properties, including its attenuation and sound speed,^{46,47} as well as its elasticity.⁶⁷ Characterization of these tissue changes may provide insight into the physical processes of thermal ablation and facilitate development of improved methods for ablation monitoring and control.

B. Effect of correlation window size

When echo decorrelation imaging is used to measure the local tissue decoherence spectrum, accuracy of this measurement depends on the spatial window $w(y, z)$ that

effectively smooths or spatially averages the computed cross-correlation and autocorrelation functions from Eq. (2). Appropriately selected windows can potentially provide more accurate estimates of the tissue decoherence spectrum with reduced statistical variability and higher spatial-frequency resolution.^{53,57} However, when the spatial extent of the window $w(y, z)$ approaches or exceeds characteristic length scales of the tissue changes, accuracy for mapping these changes is reduced.

Because the correlation window size is equivalent to the size of the smallest resolvable features in an echo decorrelation map, the optimal window size for a given application depends on the length scales of expected tissue changes. For example, in RFA (Refs. 1 and 12) or bulk ultrasound ablation,^{68,69} which thermal lesioning occurs over a relatively broad region, determined by the geometry and other properties of the ablation probes employed. Appropriate probes and exposure conditions are chosen based on the targeted size of the ablation zone, which typically ranges between 1 and 5 cm in diameter.⁵ In these cases, relatively large correlation windows (on the order of 1/6–1/2 of the targeted ablation zone diameter, as illustrated by Fig. 4) are appropriate for most accurate estimation of the scattering medium decoherence. Figure 4 also indicates that relatively high accuracy in decoherence estimates (RMS error <10%) can be obtained for correlation window diameters comparable to the lesion diameter. This range of correlation window sizes is consistent with window diameters previously chosen for testing of echo decorrelation imaging, which corresponded to approximately one-half¹ or one-third² the characteristic RFA or bulk ultrasound ablation lesion diameter of 20 mm in those experiments.

For monitoring of typical high-intensity focused ultrasound^{11–13} in which single thermal lesions may be <1 mm in width, smaller correlation windows may be more appropriate and accurate mapping of tissue decoherence may be more challenging. For such applications, a significant tradeoff exists between high spatial resolution (causing increased statistical fluctuations in image autocorrelation and cross-correlation functions and thus noisier echo decorrelation images) and low-variance estimation of local tissue changes (requiring larger correlation windows and thus lower spatial resolution). In these cases, use of higher-frequency, higher-resolution pulse-echo imaging would interrogate the tissue on smaller length scales, allowing use of proportionately smaller correlation windows and thus higher spatial resolution for echo decorrelation images.

C. Compensation of motion and noise artifacts

The approach presented here allows compensation of motion and noise artifacts from echo decorrelation images. In general, this compensation requires estimation or measurement of a baseline decorrelation map, which can then be removed using Eq. (12). Because tissue displacement can contribute non-trivially to the imaged echo decorrelation,² compensation for motion artifacts using any of these methods here should improve the efficacy of echo decorrelation

imaging for practical monitoring of clinical ablation procedures.

If decorrelation is primarily caused by known or measurable translational tissue motion, the theory developed here allows accurate estimation of this motion-induced decorrelation based on the imaging system's beam functions, which can be readily simulated or measured. Because in-plane tissue motion can be measured directly from pulse-echo image data, artifacts caused by in-plane motion can be rigorously compensated. This approach may be beneficial in some cases, particularly because signal decorrelation can depend less strongly on out-of-plane than on in-plane motion, as illustrated in Fig. 6 for experiments employing a clinical curved linear array probe.

In general, artifactual echo decorrelation also results from sources that cannot straight-forwardly be simulated. These include decorrelation caused by large tissue strains, unknown out-of-plane (elevational) motion, and electronic noise. In many cases, these components can be measured or mapped at a given location before tissue ablation is performed; the measured baseline decorrelation can then be effectively compensated during the treatment. This approach is effective if the artifactual decorrelation is approximately constant with time or if it is periodic with appropriate synchronization or gating. Alternatively, a worst-case baseline decorrelation can be estimated as the temporal maximum of the measured pre-ablation decorrelation. An echo decorrelation map compensated using Eq. (12) would then depict only decorrelation exceeding the measured level of decorrelation artifacts, which may be assumed to correspond to heat-induced changes in the tissue scattering medium.

V. CONCLUSION

Echo decorrelation imaging was theoretically analyzed to clarify the effects of tissue property changes, electronic noise, tissue motion, and measurement system effects. With appropriate assumptions, echo decorrelation was found to comprise components due to intrinsic tissue-medium changes and due to motion and noise effects. This theory, including a method for reduction of the motion- and noise-induced echo decorrelation components, was validated by simulations and experiments. The results offer improvements to echo decorrelation imaging for ablation monitoring and control, including reduction of motion artifacts, understanding of tradeoffs between spatial resolution and measurement precision, and direct characterization of tissue property changes by estimation of the decoherence spectrum.

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