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Monitoring and Guidance of HIFU Beams with Dual-Mode Ultrasound Arrays

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Abstract— **We present experimental results illustrating the unique advantages of dual-mode array (DMUA) systems in monitoring and guidance of high intensity focused ultrasound (HIFU) lesion formation. DMUAs offer a unique paradigm in image-guided surgery; one in which images obtained using the same therapeutic transducer provide feedback for: 1) refocusing the array in the presence of strongly scattering objects, e.g. the ribs, 2) temperature change at the intended location of the HIFU focus, and 3) changes in the echogenicity of the tissue in response to therapeutic HIFU. These forms of feedback have been demonstrated *in vitro* in preparation for the design and implementation of a real-time system for imaging and therapy with DMUAs. The results clearly demonstrate that DMUA image feedback is spatially accurate and provide sufficient spatial and contrast resolution for identification of high contrast objects like the ribs and significant blood vessels in the path of the HIFU beam.**

I. INTRODUCTION

Several imaging modalities including MRI [1], CT[2], [3] and diagnostic ultrasound [4], have been proposed for image guidance and monitoring of noninvasive and minimally invasive HIFU surgery. The recent availability of piezocomposite materials have led to an advancement in the ability to design and fabricate ultrasonic arrays that are capable of providing both the sufficient power needed for HIFU therapy and reasonable image feedback [5], [6]. This has led to the advent of DMUAs for use in image guided surgery, where a single DMUA can switch intermittently between therapeutic and imaging modes before, during and after treatment to provide feedback from the treatment location.

The major advantage of this approach is the co-registration between the imaging and therapeutic coordinate systems. This allows for adaptive refocusing of the HIFU beam from images obtained with the DMUA of the treatment location. The image-based refocusing algorithm minimizes the energy across critical structures, such as the ribs when targeting liver or kidney tumors, while maximizing the energy at the target location. Image feedback from single transmit focus (STF) images with the refocused HIFU beam can be used to determine the effectiveness of the refocusing algorithm by analyzing the integrated backscatter before and during treatment.

During treatment, temperature imaging is vital for monitoring and control of HIFU. The ability of ultrasound to track shifts in time and frequency of RF data has allowed ultrasound to quantitatively monitor temperature by taking advantage of the temperature dependence of the speed of sound and thermal expansion of tissue [7], [8].

The ability to monitor lesion formation during treatment is essential. Recent advances utilizing nonlinear ultrasound contrast agents in ultrasound imaging have lead to the formation of a second order Volterra filter model to extract the quadratic component from RF data and form quadratic B-mode (QB-mode) images that produce higher contrast [9], [10]. Due to the nonlinear nature of lesion formation, these same principles can be applied to increase the contrast and visualization of lesions formed with HIFU [11].

In this paper we present experimental validation of image guidance and monitoring utilizing a 64-element, 1-MHz spherical shell DMUA. The image feedback available with this prototype clearly demonstrates the DMUAs ability to accurately monitor temperature and lesion formation as well as its ability to monitor the HIFU beam to refocus around critical obstacles that partially obstruct the HIFU beam.

II. ADAPTIVE REFOCUSING

Image-based adaptive refocusing of DMUAs provides a non-invasive technique for targeting deep seated tumors where the HIFU beam is partially obstructed by the ribcage, e.g. liver. STF images obtained with the DMUA are used to identify the target and critical regions for enhancement and avoidance, respectively. We take advantage of the DMUAs inherent self registration between the imaging and therapeutic coordinate systems in the estimation of array directivity vectors at these target and critical locations. The adaptive refocusing algorithm uses these directivity vectors in solving a constrained optimization problem which minimizes the acoustical energy at the critical structures while maximizing the energy at the target location [12], [13], [14].

The solution to this constrained optimization problem is as follows:

$$\mathbf{W}_C = [\mathbf{H}_C \mathbf{H}_C^{*t} + \gamma \mathbf{I}]^{-1} \tag{1}$$

$$\hat{\mathbf{u}} = \mathbf{W}_c \mathbf{h}_T^{*t} (\mathbf{h}_T \mathbf{W}_C \mathbf{h}_T^{*t})^{-1} p_0, \tag{2}$$

where $\mathbf{H}_C$ and $\mathbf{h}_T$ are the array directivity vectors from the DMUA to the critical points and target(s) respectively, p_0 is the specified complex pressure at the target, and γ is an appropriately chosen regularization parameter. This leads to the optimal complex array excitation vector, $\hat{\mathbf{u}}$.

Adaptive refocusing of the HIFU beam has been previously been shown to reduce the incident power across critical structures by greater than 70% while improving the power

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at the target by greater than 50% [14]. In addition, the integrated backscatter from the STF images was experimentally shown to correlate with the incident power at the critical structures and target locations. Thus, image based feedback can be used to test and monitor the HIFU beam before and during therapy to ensure proper focusing throughout the treatment session.

III. Temperature Imaging

Temperature imaging with ultrasound is possible due to the ability to track shifts in time and frequency from the speckle in beamformed RF data. This ability to detect local temperature shifts arises from two physical phenomena; the change in the speed of sound and thermal expansion of tissue. RF data is collected before, during and after HIFU treatment but due to the extended time needed to perform synthetic aperture imaging, a sparse array design in which the appropriate transmit and receive elements are chosen through Vernier interpolation, such that the directivity pattern of the sparse aperture closely resembled that of the synthetic aperture was utilized with the DMUA prototype [15]. The sparse array data collection reduces the total data collection time by ≈ 95 %, thus allowing for visualization of the effects of slow temperature changes.

A. Speckle Shift Estimation Method

A time domain method in which a speckle tracking algorithm is implemented to track the echo time shifts in real time has been studied [16], [8]. In this algorithm the time shifts are differentiated axially to obtain temperature estimates. To form 2-D temperature images, windowed axial RF lines are processed to obtain spatially-temporal maps for each axial line in the image. A separable FIR 2-D smoothing filter is then applied in both the axial and lateral directions of the image [16].

B. Spectral Analysis Technique

The mean scatterer spacing has shown to be related to peaks at the harmonic frequencies of the power spectrum of pulse-echo ultrasound data where regular scatterers exist inside the resolution cell [17]. Therefore, if the frequency shifts in the harmonics can be estimated from the spectra of the pulse-echo data, a reliable quantitative estimate of the temperature can be achieved. The first algorithm to track the spectral shifts proposed using an autoregressive (AR) model [7]. The difficulty of this technique was the proper selection of the AR model. High resolution nonlinear spectral analysis techniques have been recently studied to control the resolution at frequency bands around the harmonics [17]. These techniques utilize estimates of the second order statistics to formulate an adaptive tracking algorithm to track small shifts at the harmonic frequencies. A complete description of the adaptive algorithm can be found in [17].

IV. Lesion Visualization

Lesions formed using HIFU exhibit nonlinear behavior that can be detected using pulse-echo imaging techniques similar to those used for ultrasound contrast agents. This nonlinear behavior is due to the fact that nonlinear microbubbles are present in the lesion. Due to the high resonance of the DMUA at the second harmonic, quadratic images can be formed with post-beamforming filtering of the RF data to eliminate the fundamental component, and thus image the lesion.

A. Volterra Filter

The validity of the second order Volterra filter model has previously been shown to separate the linear and quadratic components from pulse-echo ultrasound data for increased visualization of ultrasound contrast agents and lesion visualization [18], [9], [10], [11]. The previously proposed signal separation model of the response of this quadratically nonlinear system, $y(n+1)$, relies on the past m values as follows:

$$y(n+1) = y_L(n+1) + y_Q(n+1) \sum_{i=0}^{m-1} y(n-i)h_L(i) + \sum_{j=0}^{m-1}\sum_{k=j}^{m-1} y(n-j)y(n-k)h_Q(j,k) + \varepsilon(n) \tag{3}$$

where $h_L(i)$ and $h_Q(j,k)$ represent the linear and quadratic filter coefficients respectively and $\varepsilon(n)$ is the error which is assumed to be an i.i.d random variable with zero mean. The linear and quadratic components can be decomposed from the echo signal once the filter coefficients are known. These coefficients can be obtained by forming a set of linear equations from Equation 3 and solving for the minimum-norm least squares (MNLS) solution. The details of this solution can be found in [9]. Recently, an adaptive approach utilizing the least mean squares algorithm has been proposed with a computational cost that is significantly less than the MNLS solution [11]. The resulting quadratic B-mode (QB-mode) image have shown to improve the contrast of nonlinear effects compared with that standard B-mode images [18], [9], [10], [11].

V. Results and Discussion

A. Adaptive Refocusing

An STF image taken with a geometrically focused therapeutic beam was used to identify the ribs and target in a tissue mimicking phantom with Plexiglas ribs for use with the adaptive refocusing algorithm. The 50 dB STF images in Figure 1 show the results of the geometrically focused (top) and adaptive refocusing (bottom). One can clearly see the echogenicity across the ribs is reduced while the target is increased.

B. Speckle Tracking

Sparse array images were collected before, during and after a heating experiment of a tissue mimicking phantom in which the DMUA alternated between providing therapy at low power for 15 seconds and 15 seconds of data collection. Six images were taken before any heating to establish a

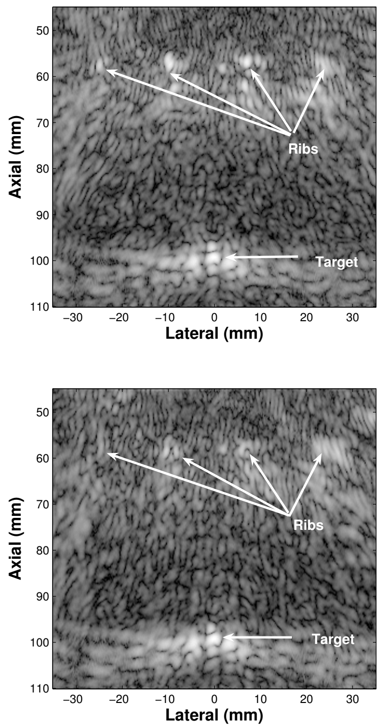

Fig. 1. Normalized grayscale images (50 dB) of the target region obtained with STF imaging using the geometric driving pattern (top) and the adaptive focusing algorithm (bottom) in a tissue mimicking phantom with Plexiglas ribs

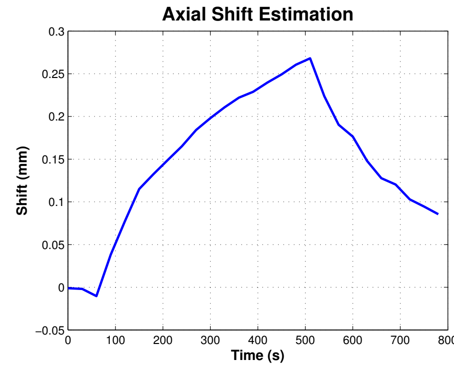

Fig. 2. Speckle tracking the axial shift of a single a-line in a tissue mimicking phantom during heating with HIFU. The shift is the result of the change in the speed of sound and thermal expansion of the tissue and is used to calculate the temperature change quantitatively.

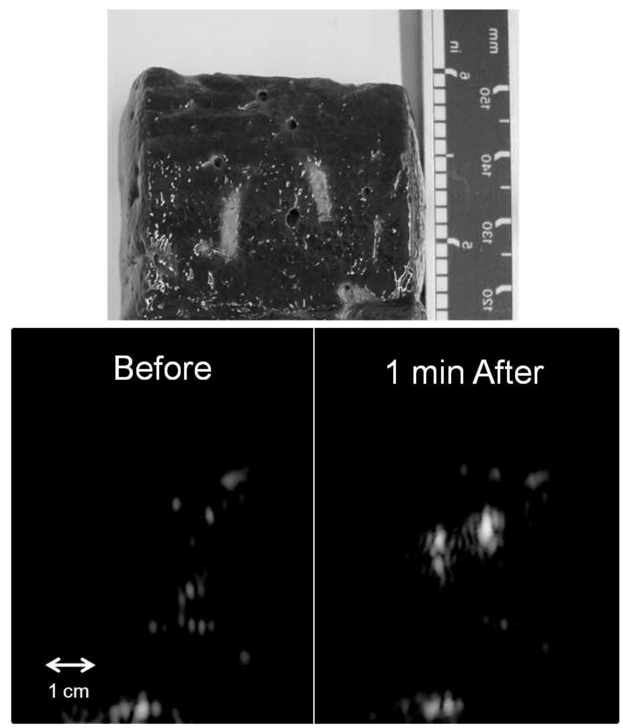

Fig. 3. Top: A cross section of a freshly excised porcine liver sample treated using a double-focus 5-second HIFU beam. Bottom: 50-dB gray scale QB-mode images of the tissue sample before and after lesion formation. The change in echogenicity at the lesion locations is quite pronounced.

baseline measurement, with 12 images taken intermittently during treatment, and 12 images taken while the phantom was cooling. The preliminary results indicate that heating can be detected by tracking the speckle shift. Figure 2 shows the shift with respect to time of a speckle location around the focus. Temperature change estimates can be calculated by differentiating the shifts along the axial direction with 2-D temperature images formed from a collection of windowed axial lines and appropriate filtering to smooth the image.

C. Lesion Visualization

An advantage of utilizing phased arrays is the ability to focus at multiple locations to reduce the total treatment time. In this experiment the DMUA was focused at two locations during a 5-second HIFU treatment in freshly excised porcine liver. QB-mode images were taken before and after lesion formation for comparison. Figure 3 shows the lesions formed in the tissue (top) and the corresponding before and after QB-mode images (bottom). The echogenicity enhancement of the lesions can clearly be seen after lesion formation.

VI. Conclusion

DMUA systems have been experimentally shown to be capable of providing the necessary feedback for image guidance and monitoring of noninvasive HIFU surgery. The current image quality available with DMUAS has accurate spatial registration, allowing for refocusing of the HIFU beam in the presence of large scattering obstacles, such as the ribs, that partially obstruct the focus and thus distort the HIFU beam. Sparse imaging and speckle tracking techniques have been shown to be able to track shifts in the RF data, which relate to temperature changes in the tissue. While the temperature change shown here was relatively slow, the design of a real-time imaging system for DMUAs will allow for temperature imaging to track faster changes such as those observed during HIFU surgery. QB-mode images showed that the visualization of lesions and the ability to track lesion formation during and after HIFU were feasible. A complete real-time DMUA system capable of imaging and therapy has been shown to be feasible and provide the ability to monitor HIFU before, during and after treatment.

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