Feasibility of Intravascular Ultrasound Ablation and Imaging Catheter for Treatment of Atrial Fibrillation

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***Abstract*— Atrial fibrillation (AF) affects 1% of the population and is responsible for 15-20% of all strokes [1]; this results in more than 460,000 hospitalizations and a cost of more than $2.8 billion per year. Clinical studies show that circumferential pulmonary vein ablation, a surgical procedure that creates contiguous patterns of lesions that electrically isolate regions of the atria, have been effective in curing AF [2]. However, current minimally invasive catheter procedures that use radiofrequency (RF) electrodes under fluoroscopic guidance are often unable to create these contiguous patterns. Difficulties arise in visualizing the anatomy and location of previous lesions and moving the catheter tip to those locations, particularly in the dynamic environment of the heart.**

Not only can intravascular ultrasound arrays deliver dynamically, steerable therapy, they can also image the region of interest with good tissue contrast to correctly position the lesion. We propose the use of an intracardiac linear ultrasound array placed at the end of a 7 French catheter. Previously, we showed an intracardiac-sized transducer (20 mm by 2 mm) can create temperature rises of 45 degrees necessary for ablation. In this work, we illustrate the possibility of visualizing such a high intensity focused ultrasound (HIFU) lesion with conventional ultrasound imaging.

I. Introduction

An intracardiac ultrasound transducer addresses the issues of precisely positioning lesions and visualizing these lesions and anatomy during the AF procedure. Ultrasound therapy can be dynamically and electronically steered to precise locations as the transducer is roughly anchored in particular position; this dynamic steering can compensate for cardiac motion and variable anatomy. Unlike fluoroscopy, ultrasound imaging also provides real-time images of tissue and lesions with good tissue contrast. This can offer significant improvement in the ability to treat arrhythmias [4], [5], [6]. Separate ultrasound imaging probes have been used to guide high intensity focused ultrasound (HIFU) transducers by displaying the anatomy and also detecting past lesions as bright echogenic regions [7], [8], [9]. Though these echogenic regions fade 1-2 min after HIFU application, they persist for long enough periods to guide the placement of adjacent lesions for formation of contiguous patterns.

Figure 1 illustrates the operation of our proposed device. The ultrasound transducer is anchored near the ostium of a pulmonary vein and secured with a balloon placed in the pulmonary vein. First the ultrasound transducer operates in imaging mode to locate the distance to the heart wall and the location of previous burns. After imaging, the ultrasound is dynamically focused and steered, and a continuous wave signal is applied to the transducer to produce a lesion. The catheter is then rotated and the process of imaging and ablation repeats. The ability to examine the region of interest and dynamically adjust the focus makes ultrasound ideal for intracardiac surgery.

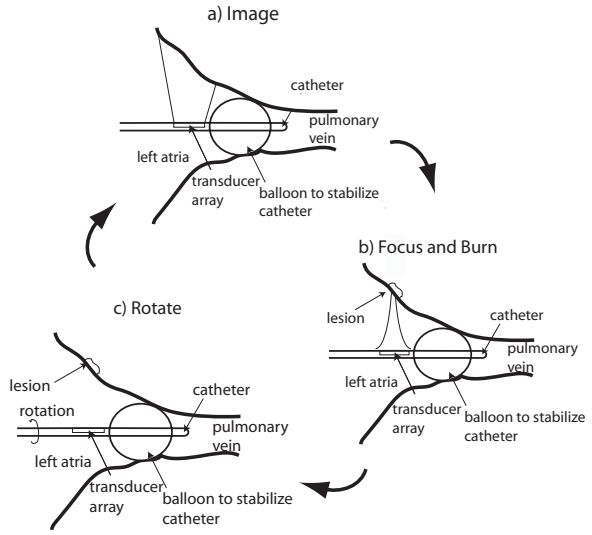

Fig. 1. An illustration of the catheter function. (a) The array will image the region of interest. (b) The anatomy and location of previously formed lesions will be identified and used to dynamically focus the therapy. (c) The catheter will be rotated and the process will repeat until a contiguous circumferential lesion is formed around the pulmonary vein.

We previously demonstrated that an intracardiac-sized transducer can produce the focal intensities necessary for ablation using a fixed focus system [3]. We will show that lesions formed from this system can be visualized using conventional ultrasound.

II. Methods and Setup

To reduce the complexity of building a full-fledged array system, we used a fixed focus system to create the lesion and a conventional ultrasound system to image the lesion. The fixed focus system consisted of a (20 mm by 2 mm) single element transducer (APC 880, APC International, Ltd.) and a focusing cylindrical reflector (10 mm focal length). We positioned the tissue at the reflector's focal point and angled the transducer to

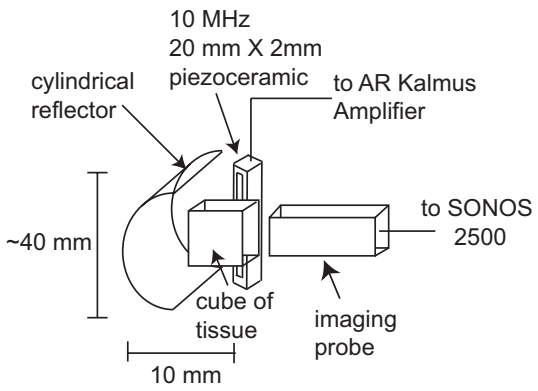

Fig. 2. A 10 MHz, 20 mm by 2 mm transducer was used for fixed focus ablation, with a focusing cylindrical mirror (focal length of 10 mm). The transducer was angled to produce an off-axis lesion in a piece of tissue. Running water was pumped over the tissue to simulate convection from blood. A 0.25 mm diameter micro-thermocouple was placed inside the tissue to measure the temperature rise at the focal point. The lesion was imaged using a 5MHz SONOS 2500 ultrasound probe.

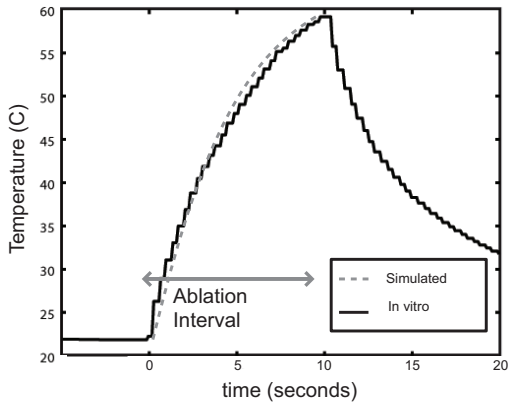

Fig. 3. Temperature rise in a piece of tissue measured by a type E micro-thermocouple. A 10W, 10MHz CW, 10 second signal was applied to the transducer in our fixed focus setup (Fig.2). A temperature rise of 45^{o}C was observed.
We simulated a lesion in CosmosWorks to determine that 275 W/cm^2 was deposited at the focus The simulated temperature rise (dotted line) matches well with the observed (solid line).

produce an off-axis lesion. We submerged the setup in running 21^{o}C water and applied a 10 MHz, 10 W continuous wave signal to the transducer (Fig. 2).

To characterize the focal intensity, we measured temperatures at the focal point with a Type E, 0.25 mm diameter micro-thermocouple (Omega, Stamford, CT). To characterize the expected lesion shape, we used a temperature sensitive HIFU gel [11], which changes from clear to opaque at 55^{o}C. We compared the size of the gel lesion with that imaged by the the HP SONOS 2500, 5 MHz imaging probe, which was placed along the direction of propagation to image the lesion in real-time.

III. Results

A. Characterization of Lesion

We measured a 45^{o}C temperature rise over 10 seconds at the focal point [3]. Using thermal simulations in CosmosWorks with a lesion size determined by the - 6 dB contour of our simulated beamprofile, we estimate that 275 W/cm^2 was deposited at the focus [3].

We characterized the lesion shape and size using a HIFU gel [11]; the 55^{o}C contour of the lesion is 1 mm by 3 mm in the focal plane, and the lesion depth is 4 mm (Fig. 4).

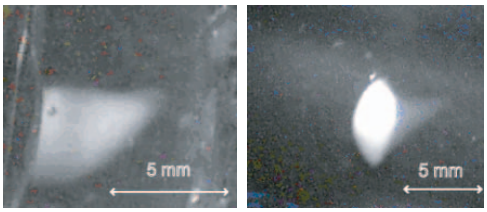

Fig. 4. Pictures of a HIFU lesion formed by our fixed focus system (Fig. 2) in the focal plane (right) and along the direction of propagation (left) formed in a HIFU gel that changes from clear to opaque at 55^{o}C.

B. Ultrasound Images of Lesions

Using a 5MHz, SONOS 2500 ultrasound probe, we imaged the lesion in real-time over a 1-2 minute period. Though the images were slightly obscured during HIFU application due to interference from the HIFU RF energy, we observed the formation of a growing echogenic spot during HIFU application. After HIFU application, the brightness of the HIFU lesion faded with time; the final brightness was comparable to the pre-HIFU image. In some samples of tissue, we observed what appeared to be popping or mechanical motion in the region where the lesion was formed. Often in these cases, the increase in echogenicity of the lesion remained more permanently. Fig. 5 shows a time progression of ultrasound images of one lesion.

We found the location of the actual lesion corresponded well with the location shown in the ultrasound image. Using our lesion detection algorithm, we tried to outline the boundary of the lesion. The size of the detected lesion in the ultrasound images was often slightly smaller than the actual lesion. The size and boundary of the lesion in the ultrasound image was difficult to determine because of speckle. Though the thresholding limit could be lowered to accommodate the lower intensity regions of the lesion, this leads to false detection of lesions elsewhere in the image from speckle. We are currently working on better image processing methods; however, underestimating the lesion area will only lead to multiple treatments of the same region, which will ensure tissue death and contiguity of the lesion pattern.

We determined the contrast of a lesion by averaging the grayscale brightness over the maximum lesion area divided by the average brightness of the same region before HIFU application. In several trials, we found the contrast in the treated region was 1.3-3. This contrast faded 15-45 seconds with a time constant of 5-10 seconds after lesion formation

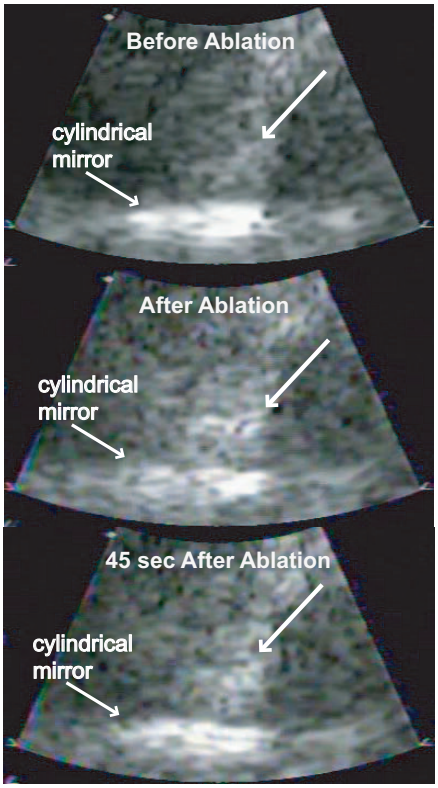

Fig. 5. Series of real-time images taken by a SONOS 2500, 5MHz probe of a our fixed focus HIFU lesion. The brightness of the lesion increases 33% from before (top) to after (middle) HIFU application. The brightness then decays over a 45 second interval (bottom).

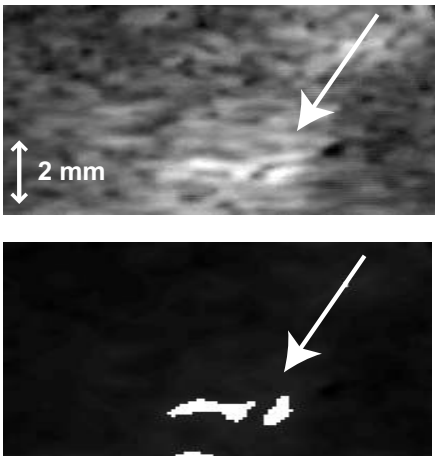

Fig. 6. Detection of lesions in a SONOS 2500 image (top) by using our image-processing technique including homomorphic filtering and prewitt edge detection.

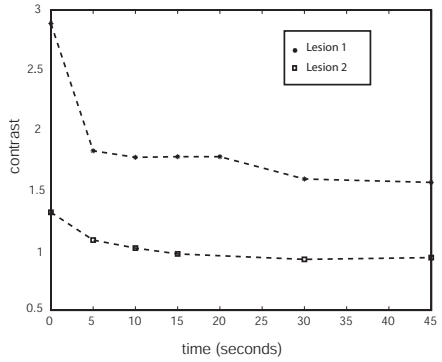

Fig. 7. The contrast of the lesions over time as measured by averaging the lesion brightness and then dividing by the brightness of the same area in the pre-HIFU image. The decay seems to be exponential with time constant between 5-10 seconds. One lesions seemed to have a contrast twice as large as the other lesions; during the formation of lesion, we observed what appeared to be mechanical motion in the lesion area indicating possible mechanical effects.

(Fig. 7). While the persistence of the lesions is rather short, we expect to image with a frequency of 15 Hz (typical of echocardiograms), so a short time constant is tolerable. In one of the lesions, the brightness seemed to increase twice as much as the other lesions. During formation of this lesion, there appeared to be mechanical motion seen in the lesion area in the real-time video. This may suggest mechanical effects, including drawing gases from solution, which may have caused a large impedance mismatch that resulted in the high echogenicity. Also, that lesion remained about 1.5 times brighter than the original image 45 seconds after ablation, while the other lesions returned to the original intensity of the pre-HIFU image. These affects will have to be studied more in detail.

Since heating and cavitation produces very noticeable non-linear affects, we tried to visualize these effects using second harmonic imaging with the SONOS 5000 4MHz probe (second harmonic at 8MHz).

In general, the contrast of the fundamental and second harmonic images was comparable. We determined signal to noise (SNR) as the average intensity of the lesion area divided by the average intensity of the other areas in the tissue. In general, the SNR of the second harmonic images was about 1.5 times greater than the SNR of the fundamental frequency.

IV. Discussion

We found it was possible to visualize contrast in tissue during the formation of a HIFU lesion. This information is necessary for an intracardiac ultrasound array to locate and steer ultrasound therapy to form contiguous lesion patterns in real-time.

The lesions in the real-time ultrasound images did not have very smooth or well-defined boundaries because of ultrasound speckle and because of the mechanical effects and non-uniformity of the tissue.

The lesions seemed to fade in 15-30 seconds on average, but we hope to achieve longer persisting lesions in the future by

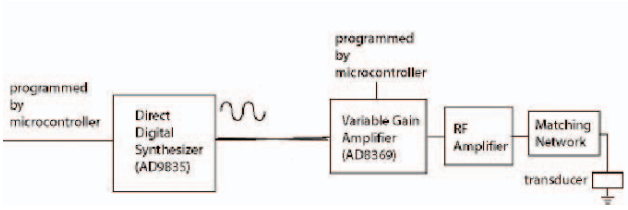

Fig. 8. Diagram of one channel of the ablation electronics system. A microcontroller programs frequency and phase of the DDS and gain stage. The output signals are then amplified and applied to the transducer elements.

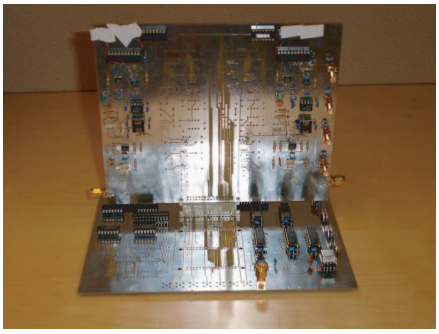

Fig. 9. Picture of the prototype ablation electronics.

increasing the focal intensity. While we only deposited 275 $\frac{W}{cm^2}$ at the focus, Vaezy's experiments showed that lesions formed from power densities of 990 to 1660 W/cm^2 produced echogenic centers that persisted for 1-2 min [8]. Since we would like to produce lesions in under a second to minimize the effects of cardiac motion on lesion formation, we will need to increase our power density to 1000-1550 W/cm^2.

Our current transducer is only 15% efficient due to losses through the backing and lack of acoustic and electrical matching. If we improve this efficiency, we could conceivably produce higher power densities. Then we should be able to visualize these burns for longer periods. We are currently improving our transducer performance by developing the proper matching and backing layers. Several preliminary 32 element arrays have been developed that deliver an output pressure of 1MPa when driven by our prototype electronics and software system. This ablation system consists of direct digital sythesizers (DDS) with 10 bit phase and 12 bit frequency resolution controlled by a PC-programmed microcontroller. The DDS signals are amplified and then applied to the individual transducer elements (A single channel is shown in Fig. 8). The prototype system (Fig. 9) can deliver 1W into an 50 ohm load with 30 dB suppression of harmonics. This system is fully scaleable and will be expanded to a 128 channel system.

V. Conclusion and Future

We have shown that imaging a lesion formed by an intracardiac-sized HIFU transducer is possible and useful for guidance of HIFU. At present, we are using a fixed focus reflector system to focus the ultrasound and observe the lesion formation. Contrast and persistence of lesions seen in the ultrasound images can be improved by increasing the transducer intensity. We are currently improving our transducer design and completing the ablation and imaging electronics necessary for operating the intracardiac array.

Acknowledgments

The authors would like to thank John Pauly for his thoughtful ideas, time, and support.

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