Optimization of Thermal Dose Using Switching Mode Patterns of a Spherically Shaped Square Element Phased Array

D. R. Daum[*] and K. Hynynen
Brigham and Women's Hospital, Harvard Medical School
221 Longwood Avenue, Room 007
Boston, MA 02115
[*]MIT and Harvard Division of Health Sciences and Technology

Abstract—A 16 element, spherically sectioned array has been constructed for application in ultrasound surgery guided by magnetic resonance imaging. To reduce the peak temperature and cooling time interval, a treatment based on rapid switching between multiple focus fields during a ten second sonication has been investigated. First, a simulation study for the array was performed to determine an optimal treatment from a set of multiple focus fields. These fields were generated using the mode scanning technique with power levels determined numerically using a direct weighted gradient search to create an optimally uniform thermal dose over a 0.6 x 0.6 x 1.0 cm^3 volume. Second, the simulated results were experimentally tested using MRI to non-invasively monitor temperature elevations and predict lesion size in rabbit thigh muscle *in vivo*.

INTRODUCTION

Therapeutic phased array ultrasound transducers are advantageous in the treatment of large tumors since they are capable of generating larger lesions than their non-phased counterparts [1]-[3]. The creation of these lesions is possible through the use of multiple focus intensity patterns which distribute power and create temperature elevations over regions much larger than the regions from a single spot focus. For multiple focus therapy to be successful, however, care must be taken to reduce near field heating and secondary temperature elevations [1][4][5]. Therefore, a single multiple focus pattern may require extensive cooling times to avoid unwanted heating while still coagulating a continuous volume [6]. Since treatment time is a major cost in MRI guided focused ultrasound surgery, this research investigates a method to rapidly switch between multiple focus patterns such that the thermal response of the array is more uniform, thus lowering peak temperatures and decreasing total treatment times. Therefore, the thermal dose may be optimized to find the lowest possible threshold power level to cause contiguous necrosis. In addition, since a small scale array (less than 64 elements) does not have as much control as large scale arrays, switching between multiple fields can be used to approximate complex fields that otherwise could not be created.

Similar simulation and experimental studies using multiple pattern switching have been performed for hyperthermia treatment [7], but have not been experimentally demonstrated for short duration sonications due to the previous lack of high resolution non-invasive temperature mapping. Recent publications, however, demonstrate the ability of magnetic resonance imaging (MRI) to map ultrasonically induced temperature elevations *in vivo* with resolutions smaller than 1 mm [8][9]. This paper will show that switching between multiple fields can be monitored using MRI to create more uniform temperature distributions—an important step towards real-time treatment control.

ARRAY DESIGN

Phased Array Design and Simulation

A sixteen element, spherically sectioned array introduced by Ebbini, et al. [10] and described by Fan, et al. [2] was designed and constructed for application in MRI guided surgery. The array was matched to 50Ω at 1.64 MHz and yielded acoustical efficiencies ranging from 70% to 80% at 2.5 W/cm^2 as measured by a radiation force technique. The array was powered by a newly constructed phased array driving system with 8-bit phase resolution and self leveling 0-60 W/channel power control. Sample beam plots of the multiple focus patterns were found to be similar to simulated results. Throughout this research the ultrasound field was simulated using a set of geometrically superimposed point sources described by Zemanek [12]. The temperature elevations were calculated numerically using the

Pennes bioheat transfer equation [13] and the dose calculations were calculated from a numerical integration of the Sapareto and Dewey model [14].

Optimization Routine

Six patterns which covered the possible region of sonication for the array geometry were chosen from a set of driving signals to be used in a power optimization routine (Figure 1). These driving signals were manually chosen from a set of experimentally successful signals calculated using the mode scanning technique described by McGough, et al. [11]. This method helps reduce near field heating by destructive interference along the axis of the transducer, thus decreasing the required treatment time for a single sonication of a multiple focus pattern [6]. These six patterns were then used as inputs to the optimization routine.

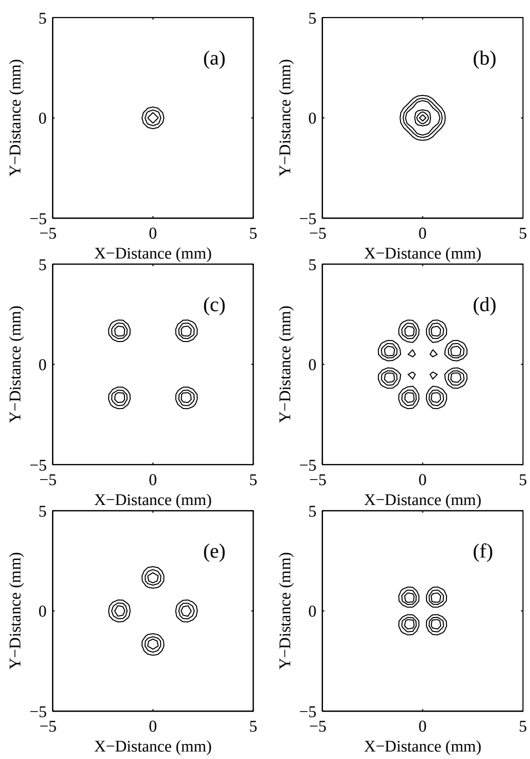

Figure 1: Field patterns generated by mode scanning used in the optimization routine.

The goal of the optimization routine was to determine the input powers for each of the pre-selected patterns. A mean-squared cost function was used to compare the simulated dose (D_{sim}) for a set of input powers with a uniform thermal dose (D_{ideal}) of 5000 minutes over a 0.6 x 0.6 x 1.0 cm^3 region (*V*) (See Equation 1).

$$C_f = \sqrt{\iiint_V (D_{sim}(x,y,z) - D_{ideal}(x,y,z))^2}$$

Eq. 1: The optimization cost function.

The power levels were then optimized using a numerically based direct-weighted gradient search with random initial conditions. During this search, it was found that the optimal driving powers were only substantial for three of the patterns {b,c,e average 76 W} therefore the other three patterns {a,d,f averaged 1 W} were discarded from the set. From a practical standpoint, reducing the number of switched fields is important due to hardware switching limitations (20 Hz). Using three fields, it was found that a switching frequency of 9 Hz was fast enough to produce an effective average field as described in [7]. The sonication length was chosen as ten seconds to allow the MRI enough time to gather several temperature images, and thus monitor the treatment during sonication. The simulated thermal dose for the optimized switching pattern and a non-switched pattern {c} of the same average power is found in Figure 2.

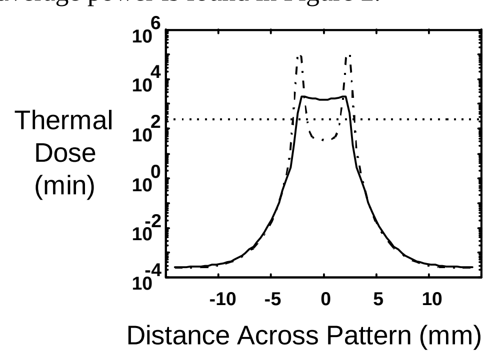

Figure 2: Simulated dose lines for optimized switching (solid) and non-switched (dotted) fields. Horizontal line corresponds to the thermal necrosis threshold of 240 min [14].

EXPERIMENTAL SET UP

The phased array was placed in a submerged 3-dimensional positioning system within a clinical 1.5 T Signa MRI (both GE Medical Systems, Milwaukee, WI) for sonication on the thigh muscle of three New Zealand white rabbits *in vivo* [8]. For each rabbit, the thigh was situated at the natural focus of the array and a series of higher power sonications both with and without pattern switching were performed while obtaining temperature sensitive images [9]. After sonication, T2 weighted images were taken to demarcate the lesion areas and evaluate treatment execution.

EXPERIMENTAL RESULTS

Figure 3 contains temperature sensitive images experimentally obtained through the focal plane of a switched sequence sonication and a single four focus sonication (pattern (b) of Table 1) of the same average power (76 W). One can see from the two images that the temperature distribution is more uniform for the switched pattern than for the simple four focus pattern. Specifically, the center of the four focus pattern was thermally "filled in" by switching between multiple focus patterns instead of relying on thermal conduction from the four outer foci. Both sonications yielded continuous lesions as determined in post-sonication images (Figure 4).

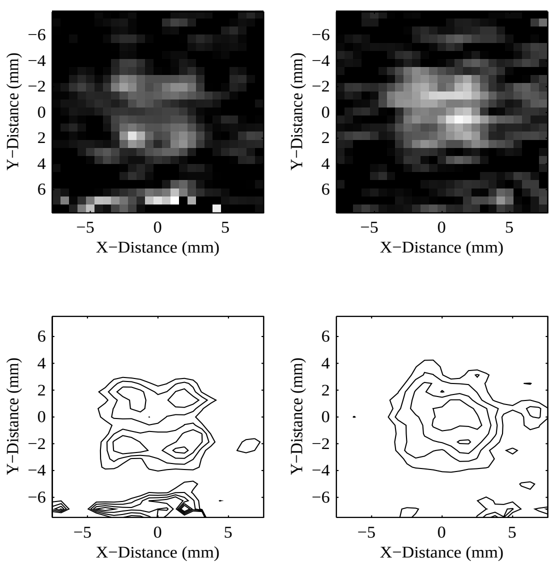

Figure 3: Thermal contour images for (left) non-switched and (right) switched sonications.

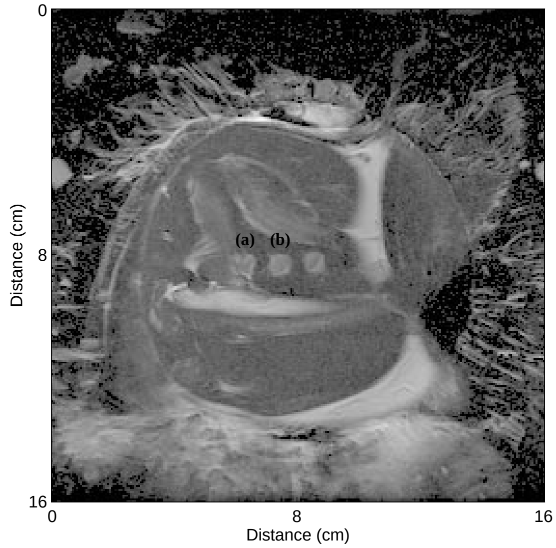

Figure 4: T2 weighted image of thermal necrosis caused by (a) single four focus pattern and (b) switched focus pattern across axis.

To show that the switched fields could produce continuous lesions at lower powers than a simple multiple focus pattern, a second set of 10 second sonications at 68 W were peformed. Figure 4 was obtained after these lower power sonications. Note that the switched pattern completely coagulated the treatment volume while the multiple focus pattern left an unaffected region in its middle. This response may be better explained by considering the temperature profiles across the sonication regions during treatment (Figure 5). The switched pattern yielded a more uniform temperature distribution and overall lower peak temperature than the single multiple focus pattern as seen in simulations.

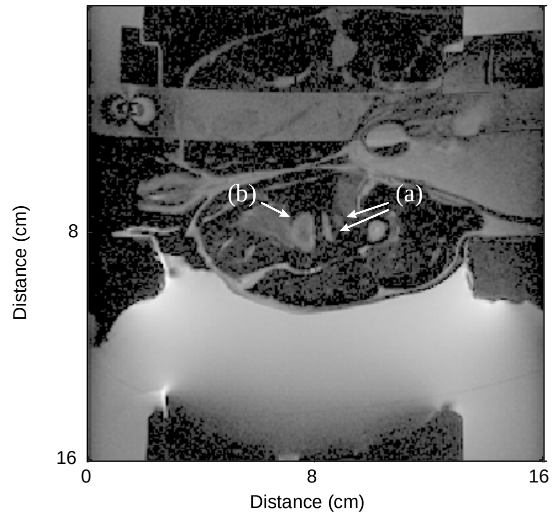

Figure 5: T2-weighted image of lesions produced using 68 W average power: (a) non-switched lesion and (b) switched pattern lesion.

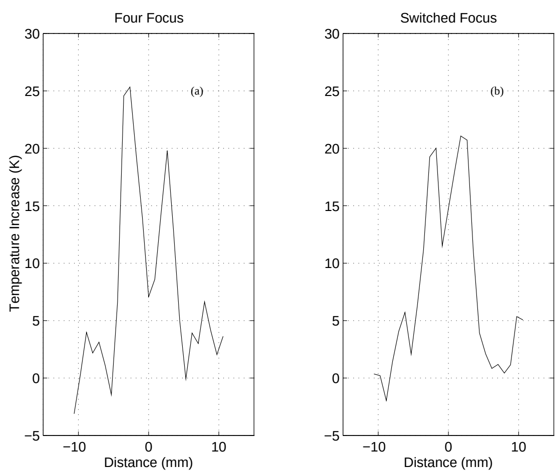

Figure 6: Temperature response across 68 W average sonications: (a) four focus pattern and (b) switched focus pattern.

DISCUSSION

The temperature images demonstrate the ability of MR imaging to detect thermal increases of switched multiple focus patterns, confirming the single multiple focus pattern results by Hynynen et al. [9]. A comparison of the temperature increases generated from a four focus pattern and the sequence of switched patterns illustrate that the temperature distribution can be improved through the rapid switching technique: the peak temperature may be decreased while still creating a complete thermal necrosis. This is particularly advantageous for small scale arrays (less than 64 elements) since they have less ability than large scale arrays to create complex and spatially continuous deposition fields.

The main advantage of optimizing power among a set of deposition patterns was to reveal those patterns which had the greatest effect on the thermal dose delivered to a given volume. The precise power levels of a set of patterns, however, is somewhat forgiving since a variation of powers using the set {b,c,e} yielded similar thermal and dose responses close to the global minimum. Therefore, when designing a treatment, the choice of deposition patterns is the most critical feature—the gradient search having the ability to highlight those patterns of greatest utility and to indicate the relative power levels for those patterns. These levels can then be proportionally scaled for *in vivo* experimentation to produce equivalent thermal gradients to the simulation.

The *in vivo* lesions demonstrate that switching suffers from tissue inhomogeneities as do the single multiple focus patterns. However, continuous lesions using the switching method can still be formed at lower average power levels than their non-switched counterparts. This is illustrated through a comparison of the 76 W and the 68 W optimized power sonications. When the higher power was used, a continuous lesion was formed for both the switched and non-switched cases. However, the lower power sonication did not yield contiguous lesions for the non-switched sonications. Thus, the switching technique can lower the threshold of power to overcome tissue inhomogeneities, although real-time monitoring and control of power deposition would be critical at these lower levels to ensure proper treatment.

SUMMARY

Short pulse duration ultrasound surgery may be improved through the use of rapid switching between multiple focus patterns. These patterns may be determined through a gradient search optimization of thermal dose to indicate the most effective set of deposition patterns for a given treatment volume. The thermal response of this treatment can then be monitored using temperature sensitive MRI images to ensure tissue coagulation. Future work to create a real time control of switching patterns and their associated power levels could further improve treatment conditions.

ACKNOWLEDGEMENTS

This research was done under NIH Grant CA46627. The authors thank GE Medical Systems for use of their temperature imaging sequences and positional equipment.

REFERENCES

[1] E.S. Ebbini and C.A. Cain, "Multiple-Focus Ultrasound Phased-Array Pattern Synthesis: Optimal Driving-Signal Distributions for Hyperthermia," *IEEE Trans. Ultra. Ferro. Freq. Contr.*, vol. 36, no. 5, pp. 540-548, 1989.

[2] X. Fan and K. Hynynen, "Control of the Necrosed Tissue Volume During Non-Invasize Ultrasound Surgery Using a 16-Element Phased Array," *Am. Assoc. Phys. Med.*, vol. 22, no. 3, pp. 297-308, March 1995.

[3] Fan, X. and Hynynen, K., "A Study of Various Parameters of Spherically Curved Phased Arrays for Noninvasive Ultrasound Surgery," *Phys. Med. Biol.*, vol. 41, pp. 591-608, 1996.

[4] C. Damianou and K. Hynynen, "Focal Spacing and Near-Field Heating During Pulsed High Temperature Ultrasound Therapy," *Ultrasound in Med. & Biol.*, vol. 19, no.9, pp. 777-787, 1993.

[5] E. Moros, R. Roemer, and K. Hynynen, "Pre-focal Plane High-Temperature Regions Induced by Scanning Focused Ultrasound Beams," *IJH*, vol. 6, pp 351-366, 1990.

[6] X. Fan and K. Hynynen, "Ultrasound Surgery Using Multiple Sonications—Treatment Time Considerations," *Ultrasound Med. Biol.*, vol. 22, no. 4, p 471-482, 1996.

[7] E.S. Ebbini and C.A. Cain, "Optimization of the Intensity Gain of Multiple-focus Phase Array Heating Patterns," *IJH*, vol. 7, no. 6, pp. 953-973, 1991.

[8] K. Hynynen, W.R. Freund, H.E. Cline, A.H. Chung, R.D. Watkins, J.P. Vetro, F.A. Jolesz, "A Clinical, Noninvasive MR Imaging-monitored Ultrasound Surgery Method," *Radiogr.*, vol. 16, pp. 185-195, 1996.

[9] K. Hynynen, A. Chung, T. Fjield, M. Buchanan, D. Daum, V. Colucci, P. Lopath, F. Jolesz, "Feasibility of Using Ultrasound Phased Arrays for MRI Monitored Noninvasive Surgery," *IEEE Trans. Ultra. Ferro. Freq. Contr.*, vol. 43, no. 6, Nov 1996.

[10] E. Ebbini and C.A. Cain, "A Spherical Section Ultrasound Phased Array Applicator for Deep Localized Hyperthermia," *IEEE Trans. Biomed. Eng.* vol. BME-38, pp. 634-643, 1991.

[11] R.J. McGough, H. Wang, E.S. Ebbini, and C.A. Cain, "Mode Scanning: A Heating Pattern with Ultrasound Phased Arrays," *IJH*, vol. 10, no. 3, pp. 433-442, 1994.

[12] J. Zemenek, "Beam Behavior within the Nearfield of a Vibrating Piston," *J Acoust. Soc. Am.*, vol. 49, 1971.

[13] H.H. Pennes, "Analysis of Tissue and Arterial Blood Temperatures in the Resting Human Forearm," *Jour. Appl. Phys*, vol. 1, pp. 93-122, 1948.

[14] S. Separato and W. Dewey, "Thermal Dose Determination in Cancer Therapy," *Int. J. Rad. Oncol. Biol. Pys*, vol. 10, pp. 787-800, 1984.