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Attenuation Mapping for Monitoring Thermal Therapy Using Ultrasound Transmission Imaging

N. Parmar[1] and M. C. Kolios[1,2]
[1] Department of Electrical and Computer Engineering, Ryerson University, Toronto, Ontario, Canada
[2] Department of Mathematics, Physics and Computer Science, Ryerson University, Toronto, Ontario, Canada

Abstract—**The use of an ultrasound (US) transmission imaging system to monitor attenuation changes during tissue heating was investigated. This paper presents preliminary results of images obtained from an acoustic camera before, during and after heating tissue phantoms using a heated needle. Two types of tissue-mimicking phantoms were used, agar and polyacrylamide-based. Regions of interests were chosen in images obtained from the real-time imaging system, and the pixel intensity values before, during and after heating were compared. In both phantoms, a decrease in image intensities was observed during heating, indicating an increase in tissue attenuation. Additionally, an irreversible change in image intensity was observed in regions close to the heat source. The reversibility of the intensity change was shown to be a function of the distance from the heating needle to the selected region. Initial results indicate that US transmission imaging can be used to monitor thermal therapy.**

Keywords—**Ultrasound transmission imaging, Thermal therapy, Ultrasound, Attenuation.**

I. Introduction

Minimally invasive thermal therapy (MITT) is an attractive alternative to surgery and radiation therapy because of its ability to locally kill tumors while preserving surrounding normal tissues. An integral part of the success of MITT is a real-time monitoring system to monitor tissue damage in order to control the area being heated and protect normal tissue. Several studies have investigated the change in acoustic properties of heated tissue. These studies have shown that tissue ultrasound attenuation rises when exposed to temperatures above 55 ^{o}C [1]-[4]. Worthington and Sherar [2] investigated changes in ultrasound backscatter and attenuation during long exposure times (30 min) at temperatures up to 65 ^{o}C in porcine kidney. They found an increase in attenuation with time to be a factor of about 1.9 after 30 minutes. Clarke [3] and Gertner [4] measured attenuation changes at higher temperature heating (70 ^{o}C to 80 ^{o}C). These studies show that depending on tissue type, temperature and heating time, attenuation can increase by a factor up to 3.5. Therefore a method of measuring tissue attenuation can be useful for monitoring thermal therapy.

This paper investigates using ultrasound transmission imaging to monitor attenuation changes. Magnetic Resonance Imaging (MRI), US [5], and Computer Tomography (CT) [6] have all been investigated to monitor thermal therapy - however these methods are either expensive, lack accurate real-time monitoring, or have poor temporal resolution during imaging. Transmission ultrasound is a promising alternative to previously studied methods of monitoring thermal therapy, since it directly measures tissue attenuation, which is known to significantly change when tissue is heated. King [7] has investigated the same system used in this study. Using high-intensity focused US to heat tissues they were able to observe reversible time-dependant variations in transmission ultrasound images in bovine fat, rabbit liver and porcine liver.

II. Method

An ultrasound transmission camera, developed by Imperium Inc. (Rockville, MD) was used for the experiments. The system, called the Acoustocam, uses a 5 MHz unfocused planar transducer to illuminate the target tissue. The resultant pressure wave strikes the tissue and is scattered and attenuated. The transmitted ultrasound energy is focused onto a piezoelectric array by an acoustic lens system. The lens system consists of a pair of acoustic lenses. The outermost lens is in a fixed position, while the innermost lens is moveable (using a focusing knob) to achieve good focus. The piezoelectric array is 1 cm^2 with 128 x 128 pixel elements and 0.085 mm pixel spacing [8]. The ultrasound energy that strikes the array is converted to an analog voltage which is digitized and fed into a frame grabber (Matrox Meteor II), which is installed in a PC. Images are recorded via a RS-170 video stream at a rate of 30 frames/s. Real-time imaging and still image recording was done using the Acoustocam imaging software, and image processing was done using Matlab® (Mathworks, Inc.). The images recorded through the camera's software are gray-scale images, resulting in pixel values ranging from 0 to 255. The lower the pixel value, the darker it appears in the image. A low pixel value is observed when little ultrasound energy is captured by the array. Hence, objects imaged with high ultrasound attenuation properties will result in darker images.

The control unit of the system, which sends excitation pulses using a spike pulser, controls the transducer. The pulser sends a 4.3 MHz impulse with a user-adjustable peak voltage ranging from 240 V to 1000 V to the transducer. Within the Acoustocam software, the

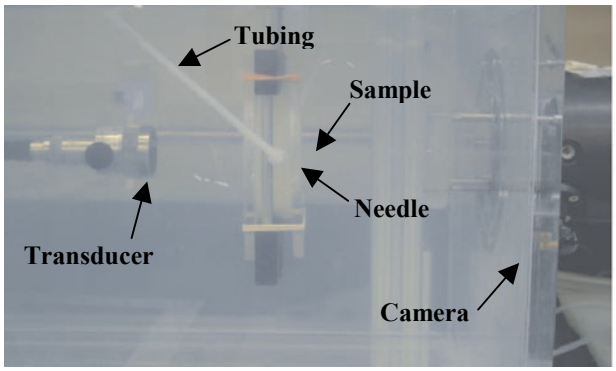

Fig.1. Experimental setup.

capture window width and transducer distance settings are set to match the physical distance between the transducer and array. The capture window width is the amount of time the camera detector is turned on. The higher the value, the more of the incident ultrasound energy will be integrated to form the image. The transducer distance is the length of water path from the transducer to the array, and is based on the assumption of an ultrasound velocity of 1500 m/s. The capture window width and the transducer signal are synchronized through the transducer distance, which compensates for the time it takes the ultrasound pulse to travel through water. This distance can be changed depending on the speed of sound of the object being imaged.

Experiments were conducted in a tank filled with degassed and deionized water. Water was degassed using Echowet® (Sonotech, Inc., Bellingham, WA) - an immersion additive that lowers the surface tension of the water, thereby releasing air bubbles, which may otherwise interfere with the ultrasound waves. The transducer was placed 25.4 cm from the camera, and the tissue was placed approximately in the middle of the two in order to obtain the best quality image. The experimental setup is shown in Fig. 1.

Two tissue-mimicking phantoms, agar and polyacrylamide-based, were used in the experiments. Samples were heated using a water bath, water pump and a 12-gauge needle (3.0 mm diameter). Hot water was pumped through the needle, which was inserted into the target phantom. The temperature of the water used was 100 ^{o} C. This resulted in an approximated surface needle temperature of 75 ^{o}C, due to the heat loss of the water being pumped through long tubing. A block diagram of the heating system is shown in Fig. 2. Phantom cooling was achieved by pumping water of temperature that matched the tank water. All phantom samples used were 1 cm to 3 cm thick. Images were recorded before, during and after heating. The phantom was placed in the tank, and pre-heated images were taken after 10 minutes to ensure that the sample had reached a stable temperature. Pre-heat images were recorded for 5 minutes, followed by 11-12 minutes of heating, and then 5 minutes of cooling.

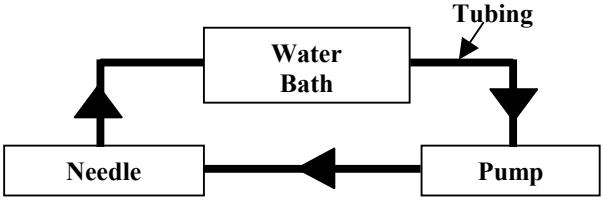

Fig. 2. Block diagram of heating system.

III. Results

For each experiment, the needle was inserted in the center of the phantom, and the phantom was positioned so that the needle appeared in the center of the image. A region of interest (ROI) was selected in the image, and the average pixel intensity within the ROI was calculated. The same ROI was then used to calculate the mean intensity value for all images/frames recorded over the duration of the experiment. Because of the large number of images stored, every 20th image was sampled and used in the analysis. This results in an analysis of 1 image/s. Various ROIs were chosen in relation to the position of the needle to determine whether areas close to the source of heating varied from those further away.

Images before, during and after heating in an agar phantom are shown in Fig. 3. The sample was heated to 75 ^{o}C for 11-12 minutes, and then cooled by pumping 20 ^{o}C water through the needle. The selected ROIs are shown in Fig. 3 (a)-(c), and Fig. 3 (d) shows the mean pixel intensity values for each ROI during the three parts of the experiment. All ROIs were 2 x 12.5 mm. For all ROIs a decrease in image intensity during the heating period was observed, indicating an increase in attenuation. In cases where the ROI was chosen adjacent to the needle (ROI #1), the increase in attenuation was irreversible during cooling. Choosing an ROI further from the needle (ROI #2 and ROI #3), resulted in a return in image intensity to its before heating values (within 5%), indicating reversible tissue changes in these areas. ROI #2 and ROI #3 were 3.5 mm and 6.5 mm from the needle, respectively. The data for the heating portion of the experiment was exponentially curve fitted. The slope of the exponential fit was calculated, and ROI #1 had an exponential rate constant 0.72 s^{-1}. ROI #2 and ROI #3 had values of 0.31 s^{-1} and 0.27s^{-1}, respectively - demonstrating a lower rate of change in attenuation in areas further from the heat source. Since no optical changes were observed in the heated agar phantoms it was not known whether irreversible temperature changes occurred. The error bars shown in the plots are the standard deviations of the calculated mean pixel intensity. As seen, the deviation for the preheat duration is very small as compared to the deviation during heating. This indicates that during heating, changes in attenuation are

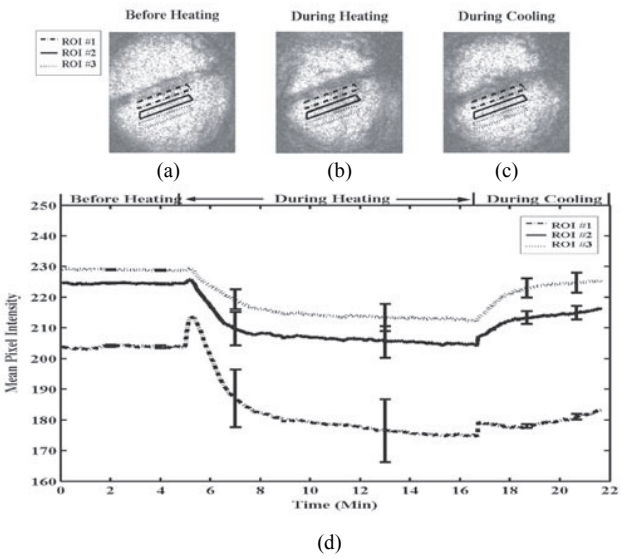

Fig. 3. Acoustocam images of agar tissue phantom (a) before, (b) during and (c) after heating. (d) Mean pixel intensity values for ROI #1-3.

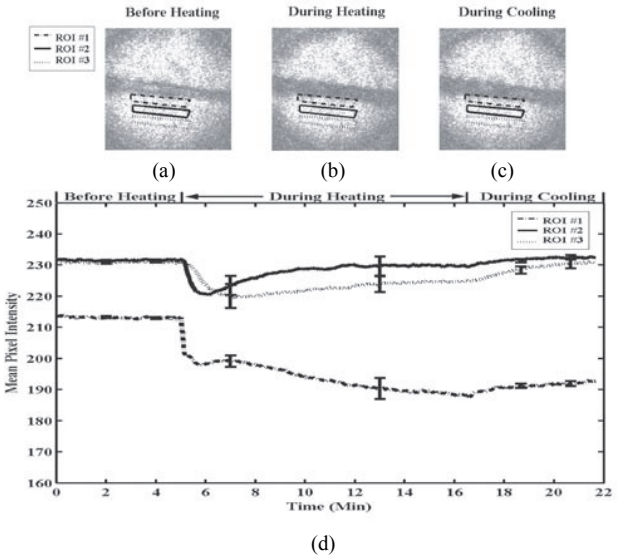

Fig. 4. Acoustocam images of polyacrylamide-BSA tissue phantom: (a) before, (b) during and (c) after heating. (d) Mean pixel intensity values for ROI #1-3.

not uniform, and that possibly the chosen ROI may have a gradient of attenuation.

Fig. 4 shows the results of heating performed on the polyacrylamide phantom. This phantom was combined with bovine serum albumin (BSA), which is a temperature sensitive protein, allowing for thermal coagulation [9]. The same heating and cooling temperatures were used as described for the agar phantom experiment. ROI #1 shows an irreversible decrease in pixel intensity, and hence an irreversible increase in attenuation. ROI #2 and ROI #3 show that the image intensity decreased during the first minute of heating, but then increased during the remaining heating period and during cooling. Thermal coagulation of this phantom did occur and the lesion size measured was 9 mm.

IV. Discussion

This study investigates the potential of using a transmission ultrasound imaging system to monitor changes in tissue attenuation during thermal therapy. The effect of heating on acoustic properties of tissue has been previously studied, and has shown increases in speed of sound and attenuation. The preliminary results in this study demonstrate that the images obtained are sensitive to attenuation changes caused by heating, and hence there is considerable potential in using such a system to monitor thermal therapy.

Changes in image intensity were observed while heating the agar phantoms. For ROI #1, the image intensity dropped from 203 to 174 during heating. ROIs further from the source of heating resulted in smaller changes in image intensity (18 pixels for ROI #2 and 14 pixels for ROI #3). This shows the temperature relation between the phantom and change in image intensity. The exponential fits of the heating duration of the agar phantoms show an expected correlation between the rate of change in attenuation with the amount of heat applied to the phantom. Areas further from the needle are exposed to less heating, and thus have smaller changes in attenuation, and so the rate at which these changes take place is more gradual. Future experiments will include temperature measurements at various points in the sample, and the attempt to correlate the temperature profile with changes in image intensity.

Regions adjacent to the needle showed a 11% – 12% decrease in image intensity during heating of the polyacrylamide-BSA phantoms. Regions further from the needle showed an initial drop in image intensity during the first minute of heating and a continuous increase for the remaining heating duration. The reason for this is not yet known. Future investigation will be done on the thermal and acoustic properties of polyacrylamide and BSA in order to determine the source of this pixel intensity increase. The polyacrylamide-BSA phantom was chosen because of its optical changes with thermal coagulation. The transparency of the phantom (which becomes white after coagulation), allows for simple lesion measurements. It is expected that boundaries of increased attenuation derived from the analysis could be compared with the actual lesion boundary. Future experiments will be done with calf liver, in which we will attempt to correlate actual lesion sizes with differences in image intensities in various regions of the image. This will

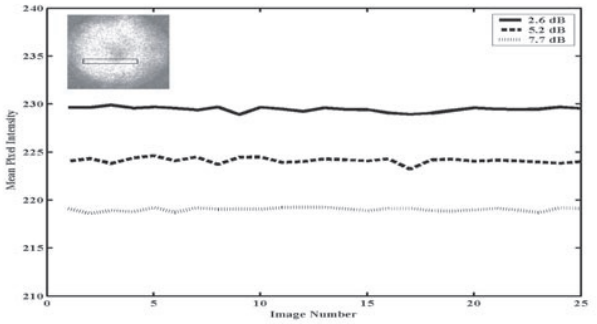

Fig. 5. Acoustocam images of polyacrylamide-BSA tissue phantoms with different attenuation properties.

confirm the camera's ability to accurately determine lesion sizes.

The relation between image intensity and attenuation coefficient has not yet been determined, however initial tests have been performed. Polyacrylamide-BSA phantoms were prepared with different Intralipid® concentrations. Phantoms with no Intralipid®, 30% concentration, and 60% concentration have attenuation coefficients of 0.26, 0.52 and 0.77 dB cm^{-1} MHz^{-1}, respectively [9]. Phantoms of 2 cm thickness were imaged, and the mean pixel intensities for a ROI are shown in Fig. 5. 500 images were taken for each phantom and every 20^{th} image was sampled. As shown, for the range examined, there is a proportional relation between the attenuation properties and the pixel intensity values. This initial test shows the reliability of measuring attenuation by observing the pixel values of images obtained from the camera. Further tests will be done under different camera settings and with various tissues types to confirm this relation.

Although no phantom temperature data was given in correlation to the changes in image intensity, the changes observed during heating and cooling lead us to believe that this is related to attenuation changes caused by heating. In both phantom tissues, irreversible structural damage was observed close to the needle, and areas further from the needle undergo less structural damage, shown by the rise in image intensity during cooling. Changes in image intensity were greatest adjacent to the needle and decreased with distance, as was expected due to less damage done to the phantom that is further from the heating source.

The results shown in this paper will be further investigated to confirm the effectiveness of using transmission ultrasound to monitor changes in tissue attenuation. Determining the relationship between image intensity values and tissue attenuation coefficients is to be determined, as well as accurate temperature mapping during experiments to obtain the time-temperature history of the ROI. The system will be further tested by heating different tissue types with known attenuation coefficients, under various heating conditions and methods.

V. Conclusion

The results shown in this paper provide a foundation to further investigate the use of an ultrasound transmission camera for monitoring thermal therapy. A correlation was found between phantom heating and image intensity changes. Irreversible tissue damage and the largest intensity changes were observed in regions closest to the needle; whereas for areas further from the needle, the damage was reversible and smaller changes in intensity were measured. This reflects on the camera's accuracy in measuring tissue attenuation. However, more work needs to be done to correlate these changes in relation to temperature measurements and attenuation coefficients.

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