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To cite this article: N R Miller and J C Bamber 2004 J. Phys.: Conf. Ser. 1 128

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Ultrasonic measurement of the temperature distribution due to absorption of diagnostic ultrasound: potential and limitations

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Abstract. Diagnostic ultrasound can cause hazardous temperature rises under certain imaging conditions. Thermocouples have been used to measure temperature, but they are invasive and it is difficult to localise the peak value. This paper examines the potential and limitations of using ultrasonic thermography to measure the temperature distribution due to a diagnostic ultrasound field. The technique exploits the principles that the sound speed in a medium is a function of temperature and that an ultrasound scanner interprets heat-induced sound speed changes as displacements. These apparent displacements can be measured by echo tracking and used to produce an image of the temperature rise. We have demonstrated in vitro that this is a sensitive technique that can portray temperature rises of <1°C above room temperature. The temperature rise can be measured in two, and potentially three dimensions, with a spatial resolution similar to that of diagnostic ultrasound. A possible limitation to the accuracy of the method is the need to know the temperature dependence of sound speed for the medium. Furthermore, some media may incur sound speed changes that are too small to be measured. The technique will not usually be possible beyond gas or bone. Finally, there are sources of error (e.g., thermoacoustic refraction and mechanical motion) that are likely to require correction. We conclude that ultrasonic measurement of temperature distributions due to diagnostic fields is worthy of study.

1. Introduction
Since the early use of diagnostic ultrasound, there has been a substantial increase in acoustic output. Operators need to be aware of the hazard-potential due to tissue heating under different imaging conditions. Simple mathematical expressions have been derived to estimate tissue heating from an in-water measurement of acoustic intensity or power. These result in a value, known as the thermal index (TI), that is displayed on the scanner alongside the ultrasound image or Doppler spectrum. Although the TI has been defined to be without units, it is an approximation to the worst-case heating in °C. However, the TI can still underestimate the temperature elevation by a factor of two [1]. Furthermore, an overestimation of the in situ temperature rise could cause the operator to compromise the quality of the examination. Therefore, there is a need for a direct method of measuring in vivo temperature rises due to diagnostic ultrasound. In addition, temperature visualisation in phantoms and excised tissue would improve our understanding of the heating patterns under different imaging conditions.

Thermocouples have been used to measure in vitro and in vivo temperature rises. However, these measurements are invasive and are subject to other limitations, including difficulty in localising the
peak temperature. Magnetic resonance thermography is a promising technique [2], but it is expensive and has relatively poor temporal resolution. This paper examines whether ultrasonic thermography can be used to measure temperature distributions due to diagnostic ultrasound.

Ultrasonic thermography [3] exploits the principle that the sound speed in a medium depends on temperature. Since the ultrasound scanner assumes a constant sound speed, changes in transit time due to tissue heating are seen as axial displacements (i.e., along the axis of propagation of the imaging beam). The axial gradient of the displacement (strain) is linearly proportional to the local change in sound speed [4]. Therefore, assuming that the temperature dependence of sound speed is linear, a strain image can be thought of as an image of the temperature rise. Alternatively, if the temperature dependence of sound speed for the medium were known, then it would be possible to reconstruct an image of the temperature rise. The technique is implemented by acquiring ultrasound images before and after tissue heating, using a block-matching algorithm to measure the apparent echo displacements, and then computing the gradient of the displacements to produce an image of strain.

This paper reviews both our own experiments, and the literature on ultrasonic thermography, with a view to examining the potential and limitations of using ultrasound to measure the temperature distribution due to diagnostic ultrasound.

2. Sensitivity

A 1.7 MHz single element transducer (focal length 15 cm, diameter 8.4 cm) was used to induce low temperature rises in degassed ex vivo bovine liver. The baseline temperature was 24°C, rather than 37°C, as the latter would have resulted in substantial bubble formation due to rapid autolysis of the tissue. A linear array imaging transducer was used to acquire pre- and post-heated radiofrequency (RF) data in a plane through the centre of the heated region (an ellipsoid of length ~26 mm and maximum diameter 0.5 cm). Images of the heat-induced echo strain were calculated as described in [5].

An example of a typical strain image (figure 1) illustrates that ultrasonic thermography can portray a circular hot spot of diameter 0.5 cm in which the spatial peak temperature rise is ~1°C. Furthermore, the cooling curve in figure 2 demonstrates that the technique can measure temperature differences of less than 0.5°C. Further work is required to verify that similar results can be achieved when starting from body temperature and when imaging different tissues and phantom materials.

1.6°C rise at spatial peak (thermocouple measurement)

![Figure 1. Image of heat-induced echo strain (image area corresponds to 3.9 x 3.9 cm).](image)

![Figure 2. Cooling curve showing the temporal change in strain at a fixed position. The temperature scale was calculated based on our observation that a temperature rise of 1°C produces an echo strain of ~0.07% [5].](image)
3. Accuracy
The heat-induced echo strain is linearly proportional to the local change in sound speed. Therefore, in order to reconstruct temperature values, we need to know the temperature dependence of sound speed, \( c(T) \), and the baseline temperature, for each point in the image. The baseline temperature prior to the application of the ultrasound field will either be known \((\text{in vivo})\) or will be easy to determine \((\text{in vitro})\,\text{by means of a single thermocouple measurement, providing equilibrium has been reached}\). For phantom experiments, it will usually be possible to characterise the \( c(T) \) of the material. Empirical \( c(T) \) curves can also be produced for \( \text{ex vivo} \) tissue samples, although spatial invariance must be assumed.

The greatest uncertainty will arise \((\text{in vivo})\). Figure 3 shows published \( c(T) \) measurements in \( \text{ex vivo} \) healthy liver. For the temperature range of interest (~37 - 45°C), the slope of the curve, \( dc/dT \), is substantially different for the two bovine liver samples. Therefore, the use of an average \( c(T) \) relationship for a particular organ in a particular species may not provide accurate temperature values. However, our recent experiments showed less between-sample variability. For example, at a particular heating-beam exposure level, the spatial peak strain in 6 tissue samples ranged from 0.12 – 0.22%. Given that this included variability in the acoustic output, variability in the acoustic attenuation by the tissue, and variability in the quality of the alignment between the imaging and heating beam, it appears as though the between-sample variability in \( dc/dT \) was low. Therefore, further \( c(T) \) measurements are required to determine whether \( dc/dT \) will need to be known on a case-by-case basis. However, if this proves to be necessary, then it may be possible to derive \( c(T) \) by means of a “calibration” heating beam, whose power deposition could be increased in known increments. Non-linear propagation effects could be avoided by using a beam with constant pressure amplitude and altering the duty cycle or exposure time to achieve different power settings. The calibration would involve producing echo strain images at each power level, allowing sufficient time between measurements for tissue cooling. The strain (i.e., change in sound speed) could then be computed as a function of relative temperature.

![Figure 3](image)

**Figure 3.** The temperature dependence of sound speed determined experimentally for a number of \( \text{ex vivo} \) normal liver tissues [6, 7].

4. Temporal resolution
Regarding rapid temperature rises, the potential for real-time strain imaging has been demonstrated [8]. Measurement of slow rises will be more challenging, as the technique relies upon comparison with an earlier ultrasound image. \( \text{In vitro} \) experiments are generally performed with a clamped transducer, so it ought to be possible to calculate temperature rises that occur over a long time interval, either by measuring strain relative to a baseline image (i.e., an ultrasound image acquired before or as soon as the heating beam is applied) or by calculating the cumulative strain from a series of frame-to-frame strain measurements throughout the heating period (i.e., incremental tracking). However, in clinical examinations using freehand scanning, there will be too much relative motion to achieve reasonable correlation with a baseline image and incremental tracking will not solve the problem that the imaging plane continuously changes and so there may not be a record of the temperature history at the site of interest. Nevertheless, it is possible that 3-D ultrasound will overcome these limitations.
5. Applicability
For *in vivo* measurements, it will be important to predict the situations in which the technique will be most successful. Ultrasonic thermography is unlikely to be possible within or beyond gas and bone. In addition, there may be some tissues in which there is a negligible change in sound speed as a function of temperature, for the temperature range of interest. This would be because water, the main constituent of soft tissue, has a positive temperature coefficient of sound speed, while fat has a strongly negative temperature coefficient. These two effects could cancel, resulting in negligible temperature dependence. We investigated this possibility using a simulation based on the mixture law [9], which predicts \( c(T) \) for a medium consisting of three components:

\[
\frac{1}{c(T)} = \frac{x_w}{c_w(T)} + \frac{x_f}{c_f(T)} + \frac{x_r}{c_r(T)}
\]

where \( x_w, x_f \) and \( x_r \) represent the volume fractions of each component and the subscripts w, f and r respectively refer to water, fat and residual components (mainly proteins and carbohydrates).

Three different media were investigated, ranging from normal liver (3.6% fat) to marked fatty liver (27.4% fat). The values of \( x_w, x_f \) and \( x_r \) for these two tissue types were obtained from Sehgal et al. [9] who measured the composition of *in vitro* human liver samples in various pathological states. The third medium considered was liver with an intermediate fat content (15.5% fat) that did not correspond to a particular pathology studied in the literature. However, Bamber et al. [10] observed considerable variation in the fat content of human liver, and from their measurements it is highly likely that liver with a fat content of 15.5% could exist. The volume fractions of water and the residual components for the intermediate case were derived from the observation that there is an approximately linear (negative) relationship between fat and water content [10]. The temperature dependences of sound speed for the components of liver tissue were taken from the literature, as described in [4].

The first stage in the simulation was to model the temperature distribution due to our heating transducer using a finite element code (figure 4). We then used equation (1) to compute the resulting sound speed distributions for the three liver pathologies. A simple analytical equation enabled calculation of the apparent displacements arising from the sound speed distributions. The next step was to simulate RF ultrasound images of the tissue before and after distortion (i.e., heating). Finally, we performed echo tracking and strain estimation to derive the strain images shown in figure 5.

For the intermediate fat level, figure 5(b), the heated region cannot be visualised, as there is only a small (<0.1%) increase in sound speed due to the induced temperature rise. This implies that ultrasonic thermography will not be possible in this tissue type. An interesting corollary to this result is that it may be possible to employ ultrasonic thermography to measure the fat content of tissues *in vivo*.

**Figure 4.** Input to simulation: temperature distribution due to a focused heating transducer (modeled region 4 cm x 4 cm).

**Figure 5.** Output of simulation: images of the heat-induced echo strain for (a) normal liver, (b) moderately fatty liver and (c) marked fatty liver. The heated region in (c) consists of negative strain values to denote a reduction in sound speed.
6. Imaging configuration

The in vitro and simulation experiments described above were performed with the imaging transducer perpendicular to the heating transducer (figure 6a), whereas the most convenient approach in vivo would be to use the heating transducer itself to image the temperature distribution (the “co-axial” configuration, figure 6b). The latter approach would require only one transducer and one acoustic window to the region of interest. It would also provide automatic registration between the heating plane and the imaging plane. However, we postulated that temperature images would be noisier for the co-axial configuration because, for a highly focused heating beam, there would be sharp lateral gradients in apparent axial displacement, resulting in increased RF signal decorrelation within the beam width. We therefore investigated the co-axial configuration experimentally, but did so using a heating transducer positioned as indicated by the dotted lines in figure 6(b).

Figure 7 compares echo strain images produced using the perpendicular and co-axial approach. The images were substantially noisier for the co-axial case, and given that there are negative strain values within the heated region, the reconstructed temperature values would be incorrect. Useful next steps would be to determine the severity of the artefact for less strongly focused heating beams and for imaging angles slightly offset from the co-axial.

![Figure 6](image_url)

**Figure 6.** (a) The perpendicular and (b) the co-axial configuration. The dotted lines in (b) indicate the position of the heating transducer in our in vitro experiments. Our simulations have shown that this is equivalent to placing the heating transducer in the same position as the imaging transducer.

![Figure 7](image_url)

**Figure 7.** Images of the heat-induced strain for (a) the perpendicular and (b) the co-axial configuration (3.7 cm x 3.7 cm region). The spatial peak temperature rise was ~1.5°C.

7. Potential artefacts

Ultrasonic thermography is subject to a number of artefacts. Firstly, as can be seen in figure 7(a), there is noise below the hot spot due to refraction of the imaging beam as it passes through the region of altered sound speed. Further work is required to determine the severity of this artefact, known as the thermoacoustic refraction effect, for less strongly focused heating beams. However, if it were found to be significant, then angle compounding appears to be a promising correction method [11].

For in vivo measurements, mechanical motion (e.g., patient motion, respiration, cardiovascular activity) may require correction. Various strategies have been proposed. Simon et al. [12] suggested
that a global shift could be estimated by fitting straight lines to the displacement profiles. Subtraction of the fitted line would leave just the heat-induced displacements. They demonstrated the feasibility of this method in phantom experiments. Strategies for dealing with cyclic motion could include a gating approach (whereby displacements are calculated from pairs of images acquired at the same moment in the cycle) and a subtraction approach (in which the periodic motion is characterised before heating, so that the cyclic component can be separated from the heat-induced shifts). These correction procedures have succeeded in improving the quality of elastograms in the presence of simulated cardiac motion [13]. We are currently studying the strategies \textit{in vitro} with application to temperature imaging.

In addition to the apparent echo shifts produced by sound speed changes, tissue heating causes real displacements due to thermal expansion. It has been demonstrated that these mechanical displacements are negligible compared to the displacements due to sound speed changes [14]. Furthermore, in our experiments, displacements due to thermal expansion were below the sensitivity of echo tracking [5].

8. Summary

Ultrasonic thermography is a non-invasive technique that has a spatial and temporal resolution comparable to diagnostic ultrasound. The sensitivity \textit{in ex vivo} bovine liver is better than 1°C, starting from room temperature. Thermoacoustic refraction is a known artefact, but angle compounding appears to be a promising correction method. The remaining limitations of the technique are confined to its \textit{in vivo} use. For example, accurate temperature measurements may require knowledge of $c(T)$ for the imaged region and the sensitivity is likely to be poor in media with an intermediate fat content. Temperature imaging in the presence of motion will require correction. Finally, the use of the heating beam to image temperature (the “co-axial” approach) may result in incorrect temperature values, while the use of an imaging beam that is perpendicular to the heating beam limits usability. However, the quality of co-axial temperature imaging has yet to be assessed for diagnostic heating beams and it is possible that steering the imaging beam will provide substantial improvement. We conclude that ultrasonic measurement of temperature distributions due to diagnostic fields is worthy of study.

9. References


Acknowledgments

The authors are grateful to Gail ter Haar, Ian Rivens, Lisa Couret, and Konstantin Bograchev for assistance with the \textit{in vitro} experiments. This work was funded by the EPSRC.