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● *Original Contribution*

ULTRASONIC TEMPERATURE IMAGING FOR GUIDING FOCUSED ULTRASOUND SURGERY: EFFECT OF ANGLE BETWEEN IMAGING BEAM AND THERAPY BEAM

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Abstract—Ultrasonic temperature imaging is a promising technique for guiding focused ultrasound surgery (FUS). The FUS system is run at an initial, nonablative intensity and a diagnostic transducer images the heat-induced echo strain, which is proportional to the temperature rise. The echo strain image portrays an elliptical "hot spot" corresponding to the focal region of the therapy transducer. It is anticipated that such images will be used to predict the location of the thermal lesion that would be produced at an ablative intensity. We demonstrated *in vitro* that heat-induced echo strain images can visualize a spatial peak temperature rise of <2°C (starting at room temperature). However, the imaging beam was perpendicular to the treatment beam in these experiments, whereas the most convenient approach *in vivo* would be to mount the imaging probe within the housing of the therapy transducer such that the two beams are coaxial. A previous simulation experiment predicted that echo strain images would be noisier for the coaxial configuration because sharp lateral gradients in axial displacement cause increased RF signal decorrelation within the beam width. The aim of the current study was to verify this prediction *in vitro*. We found, that for a temperature rise of ~4°C, the mean contrast-to-noise ratio for coaxial and perpendicular echo strain images was 0.37 (±0.24) and 2.00 (±0.72) respectively. Furthermore, the decorrelation noise seen in the coaxial images obscured the posterior axial border of the hot spot. We conclude that the coaxial configuration will be useful for localizing the hot spot in the lateral direction. However, it may not be able to depict the axial extent of the hot spot or to portray a parameter that is directly related to temperature rise. (E-mail: Naomi.Miller@icr.ac.uk) © 2005 World Federation for Ultrasound in Medicine & Biology.

Key Words: **Focused ultrasound surgery, Ultrasonic temperature imaging, Liver, Treatment guidance, Strain, Sound speed, Angle, Beam, Perpendicular, Coaxial.**

INTRODUCTION

Focused ultrasound surgery (FUS) is being developed for application to the local control of cancer and the treatment of various benign pathologies (ter Haar et al. 1989, 1998; Bihrle et al. 1994; Visioli et al. 1999; Gelet et al. 2001; Chaussy and Thuroff 2001; Wu et al. 2004). Early clinical results suggest that the technique has the potential to destroy entire tumor volumes in a conformal fashion (Wu et al. 2004). At present, there are a number of obstacles to treating large tumors, including the duration of the treatment, increased likelihood of complications and potential difficulty in finding an acoustic window (Wu et al. 2004). However, it is likely that some of these limitations will be overcome as the technique is further developed. As with other local therapies, such as surgery and radiotherapy, FUS is unlikely to be curative in patients with advanced metastatic disease.

Relative to other thermal ablation therapies, such as radio-frequency (RF) and laser ablation, FUS is the least invasive technique, since the heating source is external to the patient's body. However, a resulting disadvantage is that it is more difficult to guide the treatment, that is, to predict the location of the thermal lesion before it is formed. For example, overlying fat layers could cause aberration of the heating beam and blood vessels could act as heat sinks, drawing heat away from the treatment site. A promising method for treatment guidance is ultrasonic temperature imaging (Seip et al. 1995; Miller et al. 2004), whereby the FUS system is run at an initial,

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preablative (low) intensity and a diagnostic ultrasound transducer is used to detect the resulting echo strain. This is an apparent strain caused by heat-induced changes in sound speed. If the temperature dependence of sound speed is known, the echo strain image can be used to reconstruct the temperature rise.

Alternative thermometry techniques have been proposed in the literature, usually in the context of monitoring (rather than guiding) thermal ablation therapies. Some of these have used other modalities such as magnetic resonance imaging (Hynynen et al. 1996), electrical impedance tomography (Paulsen et al. 1996) and microwave radiometry (Meaney et al. 1996). However, ultrasound would present a number of advantages over these methods, such as low-cost, the potential for real-time temperature measurement and compatibility with the FUS system. Apart from the speed of sound, other temperature-dependent ultrasonic parameters include nonlinearity (Sehgal et al. 1986), attenuation (Ueno et al. 1990) and backscatter (Straube and Arthur 1994). However, the temperature coefficient of these properties is likely to be too small for application to FUS guidance, where the induced temperature rises will be on the order of a few degrees Celsius.

The approach to ultrasonic temperature imaging studied in this paper, originally proposed by Seip et al. (1995), exploits the principle that the sound speed in tissue depends on temperature. Since the ultrasound scanner assumes a constant sound speed, changes in transit time due to tissue heating are interpreted as axial displacements (*i.e.*, displacements along the axis of propagation of the imaging beam). It can be shown that the axial gradient of the apparent displacement (*i.e.*, apparent strain) is directly proportional to the local change in sound speed (Miller et al. 2002). Therefore, assuming that the temperature dependence of sound speed is linear, which is a reasonable approximation over a sufficiently small temperature range, an image of the heat-induced echo strain can be thought of as an image of the temperature rise.

Previous investigations of ultrasonic temperature imaging, which were discussed at length in our earlier publications (Miller et al. 2002, 2004), largely focused on achieving accurate temperature estimation, as the latter would be a desirable property of treatment monitoring systems. Some authors (Maass-Moreno and Damianou 1996; Simon et al. 1998; Sun and Ying 1999) derived expressions relating the temperature rise to time delays resulting from a combination of sound speed changes and thermal expansion. Preliminary temperature images were presented in *ex vivo* tissue using a low-intensity FUS exposure as the heating source (VanBaren et al. 1996; Le Floch and Fink 1997; Simon et al. 1998). However, while these images demonstrate the principle that ultrasonic estimation of echo strain can depict temperature rises in tissue, they do not allow us to assess the feasibility of using this technique to guide FUS. This is mainly because the authors started from a baseline temperature of 20°C, rather than 37°C, and the non-linearity of the sound speed-temperature relationship, *c(T)*, is such that a given temperature rise results in a greater sound speed change at room temperature than at body temperature (Miller et al. 2002, 2004). Furthermore, *c(T)* varies within a subject, between subjects, between species and as a function of pathologic state (Bamber et al. 1981; Sehgal et al. 1986; Miller et al. 2002; Miller and Bamber 2004). Therefore, the only meaningful way to assess feasibility is firstly to establish the maximum temperature-induced sound speed change that can be generated in a given clinical situation and secondly, to determine whether such a sound speed change can be consistently visualized. For normal human liver, starting at 37°C, the maximum achievable change in sound speed is on the order of 0.5% (Miller et al. 2002, 2004). The existence of a maximum value is due to the fact that there is a peak in the *c(T)* curve at around 50°C.

In a previous *in vitro* study (Miller et al. 2004), we demonstrated that the quality of heat-induced echo strain images is sufficient to allow visualization of elliptical hot spots (length ~25 mm, maximum diameter ~5 mm) in which the spatial peak echo strain is less than 0.2%. For our bovine liver samples, this corresponded to a maximum temperature rise of 2 to 3°C. We concluded that ultrasonic temperature imaging is likely to be useful for guiding FUS, provided it will be possible either to neglect or correct for the additional sources of error (such as cardiac-induced motion) that will arise *in vivo*.

However, the experiments described in Miller et al. (2004) were carried out with the imaging beam perpendicular to the therapy beam, whereas the system design that would be most convenient for *in vivo* treatments would involve mounting the imaging probe within the housing of the therapy transducer such that the two beams are coaxial. In addition to the fact that such a design would only require one acoustic window to the tumor site, it would provide automatic registration between the therapy plane and the imaging plane. In contrast, if the two transducers were to be arranged in a perpendicular configuration, it would be necessary to design a gantry that allows coregistration. Such a system would be bulky, more expensive and less comfortable for the patient. Furthermore, it is likely that the large acoustic window that would be required would restrict the applicability of FUS to accessible sites such as the breast, limbs, and neck and to only very superficial regions of internal organs. Therefore, the choice of angle between the imaging and therapy beams is of fundamental importance to the practical usability of the technique.