



● Original Contribution

CALIBRATION AND EVALUATION OF ULTRASOUND THERMOGRAPHY USING INFRARED IMAGING

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Abstract—Real-time monitoring of the spatiotemporal evolution of tissue temperature is important to ensure safe and effective treatment in thermal therapies including hyperthermia and thermal ablation. Ultrasound thermography has been proposed as a non-invasive technique for temperature measurement, and accurate calibration of the temperature-dependent ultrasound signal changes against temperature is required. Here we report a method that uses infrared thermography for calibration and validation of ultrasound thermography. Using phantoms and cardiac tissue specimens subjected to high-intensity focused ultrasound heating, we simultaneously acquired ultrasound and infrared imaging data from the same surface plane of a sample. The commonly used echo time shift-based method was chosen to compute ultrasound thermometry. We first correlated the ultrasound echo time shifts with infrared-measured temperatures for material-dependent calibration and found that the calibration coefficient was positive for fat-mimicking phantom (1.49 ± 0.27) but negative for tissue-mimicking phantom (-0.59 ± 0.08) and cardiac tissue ($-0.69 \pm 0.18^\circ\text{C-mm/ns}$). We then obtained the estimation error of the ultrasound thermometry by comparing against the infrared-measured temperature and revealed that the error increased with decreased size of the heated region. Consistent with previous findings, the echo time shifts were no longer linearly dependent on temperature beyond 45°C – 50°C in cardiac tissues. Unlike previous studies in which thermocouples or water bath techniques were used to evaluate the performance of ultrasound thermography, our results indicate that high-resolution infrared thermography is a useful tool that can be applied to evaluate and understand the limitations of ultrasound thermography methods. (E-mail: cxheng@umich.edu) © 2015 World Federation for Ultrasound in Medicine & Biology.

Key Words: Ultrasound imaging, Ultrasound thermography, Temperature, Infrared thermography, High-intensity focused ultrasound.

INTRODUCTION

Thermal therapies have been used to treat benign and malignant diseases. For example, hyperthermia, in which tissue temperature is elevated to 41°C – 45°C , has been used for cancer treatment in conjunction with radiation therapy and chemotherapy (Dewhirst et al. 1997; Franckena et al. 2009; Ryu et al. 2004; Zagar et al. 2010). Elevated temperatures are also exploited for controlled drug release (Staruch et al. 2011). High-temperature thermal ablations are employed to destroy diseased tissue through induction of coagulative necrosis (Goldberg et al. 2000). Among the common heat sources for thermal therapies, including high-intensity focused ultrasound (HIFU), radiofrequency (RF) wave, microwave and laser, HIFU uses ultrasound energy, is

non-invasive and is capable of inducing localized heating in deep tissue regions without affecting intervening tissue (Crouzet et al. 2010; Kennedy 2005).

Real-time temperature information in tissue is important in guiding thermal therapies. Traditional temperature measurements using thermocouples placed at discrete locations may not be feasible for clinical implementation because of the requirement for thermocouple insertion. The most promising non-invasive techniques for *in vivo* temperature measurement are magnetic resonance imaging (MRI) and ultrasound imaging. MR thermometry is based on temperature-sensitive MR parameters such as T_1 and T_2 relaxation times, proton resonance frequency and the proton diffusion coefficient (Rieke and Butts Pauly 2008; Rivens et al. 2007). Although capable of millimeter spatial resolution and temperature sensitivity of a few degrees (Jolesz 2009; Rivens et al. 2007), the acquisition duration of MRI (a few seconds per frame) (Kopelman et al. 2006) may

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limit its use in treatments involving rapid tissue heating (Rivens et al. 2007) or result in extended treatment time (Wu 2006). In addition, MR-compatible treatment devices are also required. The high cost of MRI scanners, claustrophobia and discomfort experienced by some patients in closed-bore systems and the lack of portability of MRI systems are additional disadvantages that could restrict the widespread use of the technology for thermal therapies (Kennedy 2005).

In contrast, ultrasound imaging is non-ionizing, highly portable and lower in cost. Ultrasound imaging systems are also readily compatible to various types of thermal treatment devices, especially for HIFU procedures, for which dual-mode ultrasound transducers may be conveniently used for both imaging and thermal therapy (Owen et al. 2010). Ultrasound thermography measures tissue temperature by detecting temperature-dependent changes in ultrasound backscattered signals. A variety of signal processing techniques have been explored, including methods based on ultrasound echo time shifts (Anand et al. 2007; Liu and Ebbini 2010; Maass-Moreno and Damianou 1996; Maass-Moreno et al. 1996; Seip et al. 1996; Simon et al. 1998; Varghese et al. 2002), changes in backscattered energy (Arthur et al. 2010; Straube and Arthur 1994; Tsui et al. 2012a) and spectral parameters (Amini et al. 2005). There are several challenges in applying ultrasound thermometry. To estimate temperature accurately, a sufficient signal-to-noise (SNR) level is required to detect small changes in ultrasound backscattered signals, particularly in tissues with low fat content (Miller et al. 2002). Also, the relation between temperature and changes in ultrasound backscattered signals can vary in different ranges of temperatures. As most methods depend on a linear relation that is valid up to approximately 50°C, most studies were investigated in the hyperthermia range.

More importantly, ultrasound thermography requires calibration to determine quantitatively the relationship of ultrasound signals to temperature in the specific tissue type of interest. Such calibrations may be performed with global or uniform heating of specimens submerged in a water bath with controlled temperatures (Arthur et al. 2010; Liu et al. 2009; Simon et al. 1998; Tsui et al. 2012b). However, water bath heating requires a prolonged time to allow specimens to reach a uniform temperature. For example, for a 20°C temperature range, measurements can take several hours (Simon et al. 1998). To obtain independent temperature measurements, thermocouples are often used; they need to be inserted into the specimen at locations in the field of view of ultrasound imaging (Maass-Moreno et al. 1996). Yet thermocouples are invasive, can be disruptive to the HIFU field and may introduce error in measurements because of viscous heating and thermal

conduction by the thermocouples themselves (Clarke and ter Haar 1997; Fry and Fry 1954; Morris et al. 2008; Rivens et al. 2007). In addition, measurements at discrete and sparse locations (Liu et al. 2009; Simon et al. 1998) provide limited spatial resolution often insufficient for performance evaluation of ultrasound thermometry, particularly in HIFU thermal ablations where the temperature gradient is relatively large.

In this study, we describe the use of infrared (IR) thermography for calibrating and evaluating ultrasound thermography with spatiotemporal temperature information unavailable using thermocouples. IR thermography typically uses mid-IR (3–5 μm) and long-IR (8–12 μm) spectrum for thermal imaging (Diakides et al. 2008). Although limited to surface measurements, IR thermography is easy to implement and can be readily used to measure temperature with high spatial (<100 μm) and temporal resolution (>100 Hz) without contact.

These advantages motivated the use of IR thermography for diagnosis and treatment monitoring, as demonstrated by applications in oncology (breast cancer, skin diseases), skin burns, vascular disorders, surgery, tissue viability and mass screening (Diakides et al. 2008; Ogan et al. 2003; Song et al. 2009). IR imaging has been exploited to visualize the heat deposition from HIFU transducers (Bobkova et al. 2010; Hand et al. 2009; Patel et al. 2008) and to record temperature profiles during HIFU exposures to optimize ultrasound parameters (Qiu et al. 2009; Song et al. 2005). IR thermography has also been proposed as an alternative way to calibrate transducers by converting the spatiotemporal IR temperature measurement to spatial ultrasound beam intensities through mathematical derivations (Giridhar et al. 2012; Myers and Giridhar 2011). We have also reported its use in identifying the temperature characteristics indicative of tissue coagulation and cavity formation in HIFU applications (Hsiao et al. 2013).

The goal of this study was to develop a method and an experimental platform using IR to aid the development of ultrasound thermometry. We implemented simultaneous and correlated IR thermography and ultrasound imaging, and measured the spatiotemporal temperature changes and ultrasound backscattered signals in the same surface plane of the phantoms or tissue specimens subjected to HIFU heating. We employed an echo time shift method using cross-correlation speckle tracking for ultrasound thermography (Pernot et al. 2004; Seip et al. 1996; Simon et al. 1998; Varghese et al. 2002). We first used the echo time shifts and IR-measured temperature for material-dependent calibration, and then applied the calibration coefficients to the ultrasound speckle tracking method to estimate temperature. Finally, by comparing the estimated temperature with the IR

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