Spatiotemporal filtering of MR-temperature artifacts arising from bowel motion during transurethral MR-HIFU

Alain Schmitt^{a)}

Sunnybrook Research Institute, 2075 Bayview Avenue, Toronto, Ontario M4N 3M5, Canada

Charles Mougenot

Philips Healthcare, 281 Hillmount Road, Markham, Ontario L6C 2S3, Canada

Rajiv Chopra

*Sunnybrook Research Institute, 2075 Bayview Avenue, Toronto, Ontario M4N 3M5, Canada
and Department of Radiology, UT Southwestern Medical Center, 5323 Harry Hines Boulevard,
Dallas, Texas 75390-9061*

(Received 17 April 2014; revised 17 September 2014; accepted for publication 18 September 2014;
published 23 October 2014)

Purpose: Transurethral MR-HIFU is a minimally invasive image-guided treatment for localized prostate cancer that enables precise targeting of tissue within the gland. The treatment is performed within a clinical MRI to obtain real-time MR thermometry used as an active feedback to control the spatial heating pattern in the prostate and to monitor for potential damage to surrounding tissues. This requires that the MR thermometry measurements are an accurate representation of the true tissue temperature. The proton resonance frequency shift thermometry method used is sensitive to tissue motion and changes in the local magnetic susceptibility that can be caused by the motion of air bubbles in the rectum, which can impact the performance of transurethral MR-HIFU in these regions of the gland.

Methods: A method is proposed for filtering of temperature artifacts based on the temporal variance of the temperature, using empirical and dynamic positional knowledge of the ultrasonic heating beam, and an estimation of the measurement noise. A two-step correction strategy is introduced which eliminates artifact-detected temperature variations while keeping the noise level low through spatial averaging.

Results: The filter has been evaluated by postprocessing data from five human transurethral ultrasound treatments. The two-step correction process led to reduced final temperature standard deviation in the prostate and rectum areas where the artifact was located, without negatively affecting areas distal to the artifact. The performance of the filter was also found to be consistent across all six of the data sets evaluated. The evaluation of the detection criterion parameter M determined that a value of $M = 3$ achieves a conservative filter with minimal loss of spatial resolution during the process.

Conclusions: The filter was able to remove most artifacts due to the presence of moving air bubbles in the rectum during transurethral MR-HIFU. A quantitative estimation of the filter capabilities shows a systematic improvement in the standard deviation of the corrected temperature maps in the rectum zone as well as in the entire acquired slice. © 2014 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4897382>]

Key words: thermotherapy, MRI, ultrasound, prostate cancer, temperature artifact

1. INTRODUCTION

Transurethral MR-HIFU is being developed as a minimally invasive image-guided treatment for localized prostate cancer that enables selective thermal coagulation of target regions within the gland. The treatment is performed within a clinical MRI to obtain real-time temperature measurements during heating using MR thermometry. These measurements are used as an active feedback to control the spatial heating pattern enabling high treatment accuracy. However, this requires that the MR thermometry measurements are an accurate representation of the true tissue temperature. This technology has been evaluated in gel phantoms,¹ animals,^{2,3} and pilot human studies,^{4,5} and appears to be an efficient and accurate method

for generating thermal coagulation in the prostate gland. The proton resonance frequency shift (PRFS)⁶ method is used in this treatment to provide temperature measurements during heating.^{7–9} This method involves subtraction of a reference image from an image acquired during heating, making it very susceptible to tissue motion. In addition, susceptibility changes (which result in phase changes) arising from gas in the rectum can result in temperature artifacts that are not removed through the simple phase subtraction. To address these sources of error for PRFS thermometry, several methods have been proposed.

For the periodic motion of the respiratory cycle, this issue can be covered using the respiratory gating method¹¹ if temporal resolution is not critical. A more elegant alternative is

the multibaseline method,^{12,13} in which a set of phase maps acquired prior to the heating and at different times in the cardiac or respiratory cycle can be used as the phase reference. The multibaseline method has been shown to be a robust method *in vivo* for removing motion and susceptibility-induced temperature artifacts for periodic motion in liver,¹⁴ kidney,¹⁴ and breast.¹⁵ Artifacts arising from gas in the rectum are typically not periodic, making these methods less applicable for this anatomical site.

The referenceless methods^{16,17} also perform well in the presence of tissue motion in so far as no reference baseline is needed. These methods involve estimating the reference baseline image from the current image through a polynomial fit in a region of interest outside of the heating zone. Because this method is applied to a current image, it is not sensitive to tissue motion; however, susceptibility variations that affect the region outside the heated zone can still cause temperature artifacts. In the case of transurethral MR-HIFU, the presence of phase discontinuities at the prostate and fat interfaces makes this polynomial fitting approach very challenging. This problem can possibly be circumvented by identifying the water and fat regions around the prostate and performing the polynomial fit separately on both regions.¹⁸ In addition, phase discontinuities within the prostate gland due to the presence and rotation of the transurethral device can also present challenges to this method.

During transurethral MR-HIFU, the prostate itself is negligibly affected by periodic motion¹⁹ partly due to the stabilization of the organ by the rigid transurethral treatment device, and the fact that other sources of period artifacts (lungs and heart) are far enough away to have little impact. On the contrary, the neighboring rectal wall, where accurate temperature measurements are important to ensure no thermal damage,^{21,22} can be affected by random motion caused by the presence of moving gas pockets or peristalsis of the organ. PRFS method relies on accurate registration of the reference and current phase images, as well as on an unmodified background magnetic field, as observed in the following equation, showing the resulting phase variation ($\Delta\rho$) dependence to both temperature and local magnetic susceptibility:¹⁰

$$\Delta\rho = \alpha \cdot \gamma \cdot B_0 \cdot T_e \cdot \Delta T + \gamma \cdot B_0 FT^{-1} \left[\left(\frac{1}{3} - \frac{K_z^2}{K^2} \right) FT(\Delta\chi) \right] \cdot T_e \quad (1)$$

with α , the temperature coefficient, γ , the gyromagnetic ratio, B_0 , the main magnetic field, T_e , the echo time, $\Delta\chi$, the modification of the susceptibility field, FT , the Fourier transform operator, and $K = (K_x, K_y, K_z)$, the position vector in the reciprocal space. The randomness and the complexity of the air bubble motion represents a serious obstacle to the possibility of registering accurately the reference with the current phase image, and results in a nonlinear local shift in the background magnetic field, mostly outside of the ultrasonic heated beam. However, a motion in a contained region such as bubble in the rectum can propagate to a much larger region including the prostate.

In this study, we propose a method for the filtering of temperature artifacts due to this type of random motion of air based on the variance of the temperature and an approximate a priori knowledge of the heating region. This model of the temperature variance is used during the temperature processing of each voxel to detect areas of inconsistent temperature variation. A two-step correction is then applied on the affected voxels: the corresponding temperature variations are first canceled in the first step and spatial averaging using reliable voxels is applied in the second step to maintain a low noise level. The performance of the filter is evaluated by retrospectively evaluating MRI-temperature data acquired during human treatment of localized prostate cancer in five subjects with transurethral ultrasound therapy.

2. MATERIALS AND METHODS

2.A. MRI-controlled prostate therapy system

The patient temperature data analyzed in this study were obtained from a pilot clinical evaluation of MRI-controlled transurethral HIFU performed immediately before radical prostatectomy. The results presented in this paper have been obtained by retrospectively processing data acquired during the treatment of five patients being referred to patients #1, #2, #4, #5, and #6. Note that patient #3 was excluded because the transurethral applicator catheter could not be inserted in the prostate gland, therefore, no imaging or treatment was performed.

The treatment system was composed of a transurethral ultrasound heating applicator incorporating an eight element planar transducer which was rotated in the prostate gland. A directional high-intensity ultrasound beam (4 W acoustic power/element, 4.5 MHz) was used to produce a localized region of thermal coagulation in the prostate. A motorized angular positioning system was used to rotate the ultrasound applicator in order to treat a predefined angular sector in the gland. More information about the rotating ultrasound transducer can be found in previous publications.¹

MR thermometry acquired continuously during ultrasound delivery was used to monitor the spatial temperature distribution in the prostate and its surrounding tissues. This temperature information is processed in a feedback control algorithm to modulate the rotation speed, frequency, and power output of each element of the transducer such that a desired temperature threshold is achieved along a predefined target region boundary.²⁰ Treatments were delivered in a Philips 3T Achieva MR scanner using a 32 channel cardiac coil for image reception. During treatment, MR images were acquired every 5.03 s using a gradient echo sequence ($TE = 15$ ms, $TR = 137$ ms, Flip angle = 30° , EPI factor = 13) on 7–9 contiguous transversal slices with 5 mm thickness, as well as on a unique sagittal/coronal slice including the heating applicator rotation axis, as depicted in Fig. 1. The initial acquisition setup of seven transversal slices (for patients #1 and #2) has been modified to nine transversal slices (for patients #4, #5, and #6) in order to guarantee the full coverage of larger prostates. For all patients, the field of view of each slice was

FIG. 1. Description of the system (Refs. 1 and 5) composed by a rigid transurethral heating applicator (A) and its multielement transducer (B) inserted into urethra until aligned with the prostate. Control of the temperature is achieved on 7–9 transverse imaging slices (C) and one sagittal/coronal slice (D) by performing the treatment within a MR scanner. An endorectal cooling device (E) is protecting the rectum from potential thermal damage with flowing water (Refs. 5 and 22).

256×256 mm with an acquisition voxel size of 1.16×1.23×5.3 mm and a reconstructed voxel size of 1.14×1.14×5.3 mm. The acquired matrix size was 220×208, and the reconstructed matrix size was 224×224. The slice acquisition was interleaved in order to optimize the signal-to-noise ratio (SNR). The algorithm named CLEAR was used. This algorithm uses a preliminary acquisition of the sensitivity distribution of each coil element to compensate for the intensity, inhomogeneity, and phase difference between each coil. In contrary to the direct summation of the signal received from each coil element, which can induce local signal cancellation if the coil elements are in opposite phase. This coherent summation optimizes the signal-to-noise ratio and homogenizes the distribution of the magnitude signal over the complete field of view. The receiver bandwidth was 59.7 Hz/pixel.

An endorectal cooling device, as seen on Fig. 1, was used to protect the rectum from thermal damage during treatment. Room temperature, degassed water doped with 5 mmol/l of manganese (II) chloride (MnCl₂) was circulated through the endorectal cooling device during treatment. This amount of doping was measured to result in an absence of signal (and hence artifact) from the water flowing through the device. Previous simulation²² study showed that this active cooling protocol should be sufficient to prevent thermal damage of the rectum.

2.B. Detection of artifacts in temperature maps

In the case of PRF shift MR thermometry, tissue temperature is measured with a gradient echo sequence providing a 2D complex MR image whose magnitude is associated with anatomical information and whose phase (φ) is associated with the local magnetic field variations.^{6,23} At each acquisition m , temperature maps T_m are estimated with regard to a reference phase map φ_0 estimated prior to the heating, an initial patient baseline temperature (measured physically) T_b ,

and the successive phase variations φ_k , through

$$T_m = \frac{1}{\alpha \cdot \gamma \cdot T_e \cdot B_0} \sum_{k=1}^m (\varphi_k - \varphi_{k-1}) + T_b, \quad (2)$$

where α is the temperature coefficient (0.094 ppm/°C), γ the gyromagnetic ratio (42.58 MHz/T), and B_0 is the main magnetic field (3 T).

Noisy areas of the temperature maps, usually located in regions with low SNR in the magnitude image (such as bone, fat, and air), are ignored by using a SNR mask with a spatially variable strength adapted to the heating pattern produced by the transurethral heating applicator. This is necessary because a reduced SNR is typically observed in the direction of ultrasound heating caused by signal loss due to the temperature dependence of T1 relaxation.²⁴ The spatially dependent SNR mask is derived using an approximate analytic expression for the ultrasonic heating beam Ψ_{simu} , described in Eq. (2), and built-up empirically from previous heating experiments in a tissue-mimicking gel phantom¹ and the knowledge of the transducer position at each acquisition

$$\psi_{\text{simu}}(x, y)_{x_T, y_T, \alpha_T} = \exp\left(-\frac{B}{D} \sqrt{(x - x_{T_{\max}})^2 + (y - y_{T_{\max}})^2}\right) \times \exp\left(-\frac{1}{D} \frac{(\tan^{-1}(\frac{y - y_T}{x - x_T}) - \alpha_T)^2}{2C^2}\right), \quad (3)$$

with $x_{T_{\max}} = x_T + 0.008 \times \cos \alpha_T$,

$$y_{T_{\max}} = y_T + 0.008 \times \sin \alpha_T. \quad (4)$$

x_T , y_T are the coordinates of the transducer center [m], α_T is the transducer angle (°), B is the empirical radial decay (mm⁻¹), C is the empirical parameter of the Gaussian angular fit, and D is an empirical parameter allowing to broaden the pattern. The empirical parameters used in this study have been set to $\{B, C, D\} = \{0.028 \text{ mm}^{-1}, 0.5^\circ, 4\}$ based on previous gel experiments using conservative values to ensure that

all heating is included in this region independently of the rotation speed. The heating experiments in tissue-mimicking gel phantoms used for the calibration of Ψ_{simu} involved sonicating without rotation for 46 s at a frequency of 4.64 MHz.

An estimation of the SNR (x, y) , can be obtained at each location (x, y) , by using the magnitude $\text{Mag}(x, y)$ and an estimation of the magnitude noise level, $\text{STD}(\text{Mag})$ through

$$\text{SNR}(x, y) = \text{Min}_{\forall \text{slices}, \forall \text{dynamics}} \left(\frac{\text{Mag}(x, y)}{\text{STD}(\text{Mag})} \right). \quad (5)$$

$\text{STD}(\text{Mag})$ represents the magnitude noise level whose minimum estimated value is updated each time a new dynamic is acquired and is common for all slices. At each update,

$\text{STD}(\text{Mag})$ is processed from the standard deviation of the difference of two successive dynamics, in the region of the images where both SNR masks are equal to 1.

The binary SNR mask is based on the estimation of the SNR of Eq. (5), and on a spatially dependent temperature uncertainty threshold θ , such as

$$\text{SNR}_{\text{mask}}(x, y) = \begin{cases} 1, & \text{if } \frac{1}{\text{SNR}(x, y) \cdot \alpha \cdot \gamma \cdot T_e \cdot B_0} < \theta(x, y)_{x_T, y_T, \alpha_T}, \\ 0, & \text{otherwise} \end{cases} \quad (6)$$

with the spatially dependent temperature uncertainty threshold θ defined by

$$\theta(x, y)_{x_T, y_T, \alpha_T} = \begin{cases} A_1 \times \psi_{\text{simu}}(x, y)_{x_T, y_T, \alpha_T}, & \text{if } A_1 \times \psi_{\text{simu}}(x, y)_{x_T, y_T, \alpha_T} > A_2, \\ A_2, & \text{otherwise} \end{cases} \quad (7)$$

where A_1 and A_2 are empirically tuned thresholds that have been set to $A_1 = 3^\circ\text{C}$, $A_2 = 1^\circ\text{C}$, with A_1 the temperature uncertainty threshold adapted to the heating zone and A_2 the temperature uncertainty threshold adapted to the background nonheated regions.

Figure 2 shows the spatially variable threshold $\theta_{(x, y)}$ used for the estimation of the SNR mask, as defined in Eq. (7), and the influence of the SNR mask on the resulting temperature maps.

The resulting thermal map is compensated for the baseline drift of the magnetic field B_0 by fitting a second degree

FIG. 2. An example of the spatially varying temperature uncertainty threshold $\theta_{(x, y)}$ used for the estimation of the SNR mask, as defined in Eq. (7), approximately 1 min after the beginning of sonication, with a transducer angular position indicated with a black dotted line, $A_1 = 3^\circ\text{C}$, $A_2 = 1^\circ\text{C}$ (A). The comparison of the temperature map processed without (B), and with (C) the spatial SNR mask, illustrates that the application of the SNR mask leads to reducing the noise obtained in low level areas.

polynomial function in 3D over a nonheated region at each dynamic time frame.²⁵

The temperature variations (or similarly phase variations) over time observed during *in vivo* thermotherapy can be modeled as arising from multiple sources:

- (F1) the heating of the targeted area,
- (F2) measurement noise, and
- (F3) the effects of tissue motion on the phase signal including the geometrical misalignment of the reference with the current phase maps as well as the magnetic susceptibility perturbations induced by breathing, gas, or devices in the vicinity of the image.

The goal of the filter is to remove F3 without increasing F2 and with no change to F1.

Figure 3(A), which presents an example of a MR thermal map obtained for a treatment that was affected by the presence of a moving air bubble, identifies the three typical areas of temperature fluctuations (F1, F2, and F3), and Fig. 4(A) shows their typical temperature variations over time, as observed on a voxel located in each of these areas, namely, in the heating zone (dominated by F1 fluctuations),

in the background noise (dominated by F2 fluctuations), and in the moving air bubble zone (dominated by F3 fluctuations).

The criterion proposed to estimate the temporal fluctuation (F1, F2, and F3) of the temperature is defined by

$$CR(x, y)_m = \sqrt{\frac{1}{M-1} \sum_{k=m-(M-1)}^m (dT_{k(x, y)} - \langle dT_m \rangle_{(x, y)})^2}$$

$$\text{with } \langle dT_m \rangle_{(x, y)} = \frac{1}{M} \sum_{k=m-(M-1)}^m dT_{k(x, y)}, \quad (8)$$

with m being the index of the m th acquired phase map, and M the size of the temporal window which has been set to three unless otherwise stated. This criterion, $CR(x, y)_m$, corresponds to the standard deviation of the temporal first derivative of the temperature dT over a sliding window of M dynamics and provides an estimation of the total temporal fluctuations being composed of F1, F2, and F3 fluctuations in our model.

FIG. 3. MR thermal maps (overlaid on top of the magnitude MR image) during transurethral ultrasound therapy. The presence and gradual motion of an air bubble in the rectum (Δ) resulted in a spatially localized temperature artifact (+) that interfered with the real heating caused by the high-intensity ultrasound beam aimed at the bottom left side of the image (*). (A) Temperature map without filter. (B), Temperature map with temperature variation forced to zero in the artifact-detected regions, which corresponds to the application of the filter correction step 1 only, as described in Eq. (15). (C), Temperature map with the criteria of (B) and with the temperature corrected based on neighbor voxels, which corresponds to the application of the filter with both corrections described in Eqs. (15) and (16). The dotted lines on (A) represent the typical zones of interest used for the calculation of the evaluation metrics TSTD, PCORR, and TDIFF: the rectum zone (bottom), the anterior prostate zone (top right), the posterior prostate zone (middle right) and the heated zone (top left).

FIG. 4. Typical temperature variations over time observed during treatment of patient #6, slice 3. (A), Temperature as a function of time, (B) fluctuation criterion (CR) defined in Eq. (9) as a function of time for the voxels tagged in: a voxel in the heating zone (*), a voxel in the moving air bubble zone (+), and a voxel in background noise (X).

Figure 4(B) presents the criterion CR's values over time observed on voxels whose variations are dominated by F1, F2, or F3 fluctuations. The CR's values for the moving air bubble zone are clearly higher than for both other zones, especially on the first 200 dynamics temporal range where the air bubble's motion along the rectum wall was the most pronounced.

In the absence of perturbation due to motion, F3 is equal to zero. The proposed detection of the areas affected by motion consists in setting the upper acceptable limit of the total fluctuations $CR(x, y)_m$ to a threshold that should be high enough to ensure that the trusted temperature information (F1 and F2 type temporal fluctuations) is never filtered out (in the heating zone for instance) and small enough to enable the maximum removal of the artifact (F3 type temperature fluctuations). This threshold can be dynamically estimated with F1 and F2 fluctuations components

$$CR_{\max}(x, y)_m = F1_{\max}(x, y)_m + F2_{\max}(x, y)_m. \quad (9)$$

The F1 type fluctuations upper bound $F1_{\max}(x, y)_m$ is approximated using the simulated ultrasonic heating beam pattern Ψ_{simu} described in Eq. (3)

$$F1_{\max}(x, y)_{xT, yT, \alpha T} = A \times \psi_{\text{simu}}(x, y)_{xT, yT, \alpha T}, \quad (10)$$

where A is the dynamically estimated maximum of the criterion CR [as defined in Eq. (8)] in the heating zone area defined as the 25% highest amplitude area of the Ψ_{simu} map.

For F2 type fluctuations, the upper bound pattern is related to the estimation of the measurement noise at each position of the thermal map. This $F2_{\max}(x, y)_m$ threshold is built as the upper limit of the fluctuation criterion's [CR as defined in

Eq. (8)] noise level's 99% confidence interval

$$F2_{\max}(x, y) = \sigma(dT)_{(x, y)} \times K_0 \times UL_{\text{Chi}^2}, \quad (11)$$

with $\sigma(dT)_{(x, y)}$ being an estimation of the standard deviation of the temperature variation at voxel (x, y) , obtained through

$$\sigma(dT)_{(x, y)} = \frac{\sqrt{2}}{\text{SNR}(x, y) \cdot \alpha \cdot \gamma \cdot T_e \cdot B_0} \quad (12)$$

and UL_{Chi^2} being defined as the upper limit of a chi-squared distribution with M degrees of freedom, such as

$$UL_{\text{Chi}^2} = \sqrt{\frac{2 \cdot \Gamma_{\text{inc}}^{-1}(p, \frac{M}{2})}{M - 1}}. \quad (13)$$

Γ_{inc}^{-1} is the inverse incomplete gamma function, with $p = 0.99$.

The relation described by Eq. (13) was derived from the cumulative chi-squared distribution estimating the probability p that the sum of M squared Gaussian noise draws overpass a threshold.

K_0 is a security factor empirically set to 2.2 which has been derived from the visual estimation of the optimal threshold on patient data and gel data. Indeed, lower values of K_0 ($1.0 < K_0 < 2.2$) lead to scattered correction being observed outside of the bubble artifact zone. Using this security factor proves to be useful to compensate for the additional fluctuations that can occur for mixed reasons like respiratory motion artifact over the complete image, inaccurate baseline drift correction, and underestimation of the standard deviation.

Based on the $F1_{\max}(x, y)$ and $F2_{\max}(x, y)$ fluctuation threshold maps, a mask of the artifact-detected region is finally estimated through

$$\text{Mask}(x, y)_m = \begin{cases} 0, & \text{if } CR(x, y)_m > CR_{\max}(x, y)_m \text{ and } \text{SNR}_{\text{Mask}}(x, y)_m = 1 \\ 1, & \text{otherwise.} \end{cases} \quad (14)$$

For the nonheated slices located at the periphery of the heating zones, the detection mask calculated in Eq. (14) is esti-

mated using the $F1_{\max}(x, y)$ fluctuation threshold map corresponding to the nearest heated slice. For the nonheated slices

that are not located in the neighborhood of a heated slice, the $F1_{\max(x,y)}$ fluctuation threshold map is set to zero.

2.C. Correction of the artifact-detected regions

The proposed correction method is based on a hybrid strategy composed of the cancellation of the temperature variation for the latest acquired MR data and a weighted spatial averaging using the valid neighbor voxels. This two-step strategy was chosen for its ability to eliminate the artifact-detected temperature variations while keeping the noise level low through spatial averaging.

The correction can be summarized as follows:

1. Cancellation of the current temperature variation for the masked voxels

$$\begin{cases} T_{C1}(x,y)_m = T(x,y)_{m-1}, & \text{if Mask}(x,y)_m = 0 \\ T_{C1}(x,y)_m = T(x,y)_{m-1} + \Delta T(x,y)_m, & \text{otherwise.} \end{cases} \quad (15)$$

2. Averaging of the voxel with its non SNR masked neighbors

$$\begin{aligned} T_C(x,y)_m &= \frac{1}{N_{VN}} \left(T_{C1}(x,y)_m + \sum_{(x_k,y_k) \in \{\text{valid neighbors}\}} T_{C1}(x_k,y_k)_m \right), \end{aligned} \quad (16)$$

with N_{VN} , the number of non SNR masked neighbor voxels including the temperature (x,y) obtained at the previous dynamic ($1 \leq N_{VN} \leq 9$).

2.D. Filter evaluation metrics

The evaluation of the above filter was accomplished by quantifying specific metrics over five zones of interest. The first zone of interest covered the entire acquired slice and included the prostate, bladder, rectum, muscles, and the rest of the abdomen. The second zone of interest was restricted to a smaller manually drawn zone containing the rectum only. This zone enabled evaluation of the effectiveness of removing F3 type heating. The third and fourth zones are defined as 9.12×9.12 mm square zones in the prostate where a high SNR is available and no heating is expected due to its remote location from the targeted zone. The third zone has been chosen preferably in the anterior part of the prostate, far away from the rectum so that no air bubble artifact is present in this zone. The fourth zone has been chosen preferably in the posterior part of the prostate, close to the rectum so that a possible artifact affecting the rectum would also be present in this more homogeneous prostate area. The third and fourth zones serve to whether the filter has any negative impact on F1 heating. The fifth zone is defined as a 11.4×11.4 mm square area located in a heated zone of the prostate, also present to evaluate the impact of the filter on F1 and F2 heating. The locations and sizes of these zones are depicted in Fig. 3(A) for the example of the slice 3 of patient #6.

The temporal standard deviation (TSTD), used to evaluate the performance of the filter, is defined for each position (x,y) in the map of a given slice S as

$$\text{TSTD}(x,y)_S = \sqrt{\frac{1}{N} \sum_{k=1}^N (T(x,y)_k - \langle T(x,y) \rangle)^2} \quad (17)$$

with N being the number of dynamics, and $T(x,y)_k$ the temperature for the position (x,y) at dynamic k .

The average of the temperature for a given slice is calculated over each position in the zone of interest by

$$\langle \text{TSTD} \rangle_{\text{slice}} = \frac{1}{N_{\text{zone}}} \sum_{(x,y) \in \text{zone}} \text{TSTD}(x,y)_S \quad (18)$$

with N_{zone} being the number of voxels included in the zone of interest.

The average temporal standard deviation of the temperature for a given patient by

$$\langle \text{TSTD} \rangle_{\text{patient}} = \frac{1}{N_{\text{Slices}}} \sum_{\text{Slices}} \langle \text{TSTD} \rangle_{\text{slice}} \quad (19)$$

with N_{Slices} the number of slices.

The $\sigma(\text{TSTD})_{\text{patient}}$ metrics provide an estimate of the spatial variance of the TSTD metric, through

$$\begin{aligned} \sigma(\text{TSTD})_{\text{patient}} &= \frac{1}{N_{\text{Slices}}} \sum_{\text{Slices}} \left(\frac{1}{N_{\text{zone}}} \sum_{(x,y) \in \text{zone}} (\text{TSTD}(x,y)_S - \langle \text{TSTD} \rangle_{\text{slice}})^2 \right). \end{aligned} \quad (20)$$

It has to be noted that voxels excluded by the SNR mask were not considered in the calculation of $\langle \text{TSTD} \rangle_{\text{patient}}$ and $\sigma(\text{TSTD})_{\text{patient}}$ since these correspond to previously identified noisy regions.

Another metric was introduced to evaluate the frequency with which the correction has applied throughout the course of the treatment on each voxel. For each position (x,y) in the temperature map, this metric is defined as the percentage of dynamics in which the temperature has been corrected over the entire therapy

$$\text{PCORR}(x,y)_S = \frac{100}{N} \sum_{k=1}^N (1 - \text{Mask}(x,y)_k). \quad (21)$$

The average percentage of correction for a given patient is calculated over each position in the zone of interest of every slice of the patient by

$$\begin{aligned} \langle \text{PCORR} \rangle_{\text{patient}} &= \frac{1}{N_{\text{Slices}}} \sum_{\text{Slices}} \left(\frac{1}{N_{\text{zone}}} \sum_{(x,y) \in \text{zone}} \text{PCORR}(x,y)_S \right), \end{aligned} \quad (22)$$

with N_{zone} being the number of voxels included in the zone of interest and N_{Slices} the number of slices.

Finally, the average difference between the filtered temperature and the unfiltered temperature is evaluated by patient in specific zones, in order to estimate the influence of the filter by comparison with the unfiltered result. For each voxel, the temperature difference is evaluated by

$$\text{TDIFF}(x, y) = |T_{\text{Filtered}}(x, y) - T_{\text{unfiltered}}(x, y)| \quad (23)$$

and the average difference by patient in a specific zone by

$$\langle \text{TDIFF} \rangle_{\text{patient}} = \frac{1}{N_{\text{Slices}}} \sum_{\text{Slices}} \left(\frac{1}{N_{\text{zone}}} \sum_{(x, y) \in \text{zone}} \text{TDIFF}(x, y) \right). \quad (24)$$

3. RESULTS

3.A. Performance of the filter on reprocessed patient data

Figure 3 presents thermal maps obtained for a treatment that were affected by the presence of moving air bubbles in the rectum. The thermal maps are centered and zoomed on the prostate and rectum zone, covering a 5×6 cm surface from the third slice imaged during the treatment. The acquisition time of this dataset corresponds to 29 min and 5 s (dynamic 277) after the beginning of the treatment, which is near the end of the sonication period. The maps show the current temperature with a color scale ranging from 40°C (blue) to 70°C (red) superimposed on the magnitude MR data presented in a gray color scale in order to provide the anatomical background. The light blue line starting at the center of the prostatic urethra corresponds to the angular position of the ultrasonic beam center, and the darker blue contours stand for

the prescribed contour to ablate. The temperature map processed without the filter (A) exhibits a heating pattern widely extending at the bottom right of the prostate and around the top and the right side of the rectum wall. This temperature artifact is due to the motion of an air bubble in the rectum which can be identified on the anatomical background magnitude image on the top right of the rectum. During therapy, this air bubble was gradually (and randomly) moving from one side of the rectum to the other (and possibly in and out of plane as well), leading to temperature artifacts of various amplitudes and geometric extent. The thermal maps processed with the filters [(B) and (C)] display modified temperature values mainly in the zone around the rectum. When only the first step [see Eq. (15)] of the correction is applied as in Fig. 3(B), which means that the artifact-detected regions have their current temperature variation canceled only, the resulting thermal map is scattered with noisy voxels. When the two-step correction is applied as in Fig. 3(C), the temperature is much lower at the proximity of the rectum with only few temperature artifacts remaining in the bottom right of the prostate. Qualitatively, the image in (C) appears to be the case with the best removal of the artifacts around the rectum.

The maps presented in Fig. 5 provide a spatial representation of the TSTD and percent of correction (PCORR) metrics introduced, respectively, in Eqs. (17) and (21). The temporal standard deviation of the third slice of the prostate and rectum centered zone is given in Fig. 5(A) for the processing without filter, in Fig. 5(B) for the processing with the filter featuring only the 1st step of the correction, and in Fig. 5(C) for the processing with the filter and the complete two-step correction. The temperature standard deviation map corresponding to the processing without filter in Fig. 5(A) exhibits

FIG. 5. TSTD maps ($^\circ\text{C}$) for patient #6, slice 3 over the prostate/rectum area, without filter (A), with temperature variation forced to zero in the artifacted regions (B), and with the temperature corrected based on neighbor voxels (C). Percentage of corrections map (%) for patient 6, slice 3 over the prostate/rectum area (D).

two zones of important fluctuations, one having its maximum in the middle of the heating zone and the other being distributed around the rectum. It can be observed in Figs. 5(B) and 5(C) that the standard deviation of the temperature is not significantly altered by the application of the filter in the heating zone, which is consistent with the filter design in so far as the detection threshold [see Eqs. (9) and (10)] is built in accordance with the temperature variations in the heating zone. On the other hand, in the zone around the rectum, using the filter lowers the standard deviation in both cases. For the first step correction only, a scattered number of voxels still have a TSTD over 10 °C, whereas for the two-step correction, nearly all voxels in the neighborhood of the rectum have a standard deviation similar to the originally nonartefacted prostate and rectum areas. Figure 5(D) shows the PCORR metric for the same slice with a color scale ranging from 0% to 20% however some values reach as high as 36.8% in this zone. The highest frequency of correction is located on the

top right and the right of the rectum, and the heating zone as well as the rest of the prostate is generally low with a percentage of correction under 4%.

Figure 6 presents a quantitative summary of the metrics TSTD and PCORR for each patient treatment [see Eqs. (19) and (22)] over two different zones of interest; Figs. 6(A)–6(C) focuses on the rectum zone, which is defined as a manually drawn circular zone around the rectum, depicted as a dotted line at the bottom of figure 3(A). Figures 6(D)–6(F) focuses on the global zone of interest, covering the whole slice for each patient.

Figures 6(A) and 6(D) show the average PCORR for each patient. For the rectum zone, the highest percentage of correction is obtained for patient #6 at 3.2% which is consistent with this patient being highly affected by the motion of an air bubble in the rectum, and the lowest value is obtained for patient #2 at 0.23% which was not concerned by this artifact.

FIG. 6. Comparison of the metrics $\langle \text{PCORR} \rangle_{\text{patient}}$, $\langle \text{TSTD} \rangle_{\text{patient}}$, and $\sigma(\text{TSTD})_{\text{patient}}$ for all patients, on the rectum zone [left: (A), (B), and (C)], and on the global zone [right, (D), (E), and (F)] without (left bar) and with filter (right bar).

For the global zone, the average percent of correction is always lower than 1.8% with the highest values reached for patient #1 and patient #4, both being highly affected by respiratory motion of the abdomen wall. Figures 6(B), 6(C), 6(E), and 6(F) compare the TSTD metrics with (right bar) and without (left bar) the use of the filter for each patient. The average temperature standard deviation ($\langle \text{TSTD} \rangle_{\text{patient}}$) values are generally lower with the filter than without the filter as the cancellation of the artifact-detected temperature variations tends to remove the most important variations. On the rectum zone, the highest gain is observed for patient #6 where the TSTD drops from 4.6 °C to 2.4 °C. On the global zone, the highest gain is observed for patient #2 where the standard deviation drops from 2.0 to 1.0. The application of the filter also strongly reduces the spatial variance of the temperature variation within each slice ($\sigma(\text{TSTD})_{\text{patient}}$) for all patients on both zones of interest. The improvement factor ranging from 3× to nearly 20× in the global zone (F) is due to the correction of voxels which had very high temperatures before the filter was applied.

The estimation of the influence of the filter, done by comparing the filtered and unfiltered temperature results in known zones of interest, is provided in Fig. 7. In each of these zones of interest, the severity of the artifact was assessed by using their known distance from the rectum and by using the knowledge of the severity of the artifact gathered during the treatment. Patients #2 and #4, who exhibited very low motion artifacts in the rectum during the treatment, show low $\langle \text{TDIFF} \rangle$ levels under 0.5 °C in all zones.

Patients #1 and #6, being affected by severe artifacts during the treatment, show average TDIFF levels higher than 2 °C in the rectum zone. For these patients, the prostate anterior and posterior zones, which are both nonheated with a high SNR, have the filter showing very different average TDIFF. In the prostate posterior zone which is located close to the rectum, $\langle \text{TDIFF} \rangle$ is 4–21 times higher than in the prostate anterior zone located far from the rectum zone where the artifact is taking place.

In the heating zone, the filter has only very low influence with TDIFF average levels systematically under 0.11 °C for all patients.

FIG. 7. Comparison of the metrics $\langle \text{TDIFF} \rangle_{\text{patient}}$ for all patients, on the rectum zone (1st bar from the left), the prostate posterior zone (2nd bar from the left), the prostate anterior zone (3rd bar from the left), and the heating zone (right bar).

3.B. Optimization of the detection criterion

The filter's detection criterion described by Eq. (8) has been studied for the optimization of its main parameter M , the size of its sliding window. Figure 8 presents the resulting metrics TSTD and PCORR for all patients, when the parameter M is varying between 3 and 25, in the rectum zone (A) and (B) as well as in the global zone (C) and (D). As reference, the $\langle \text{TSTD} \rangle_{\text{patient}}$ plots (B) and (D) also show this metric without the application of the filter on the ordinate axis. For both zones and for all patients, the average PCORR is a monotonically increasing function of the parameter M , meaning that a higher M parameter will lead to a less conservative filter with more temperature variations canceled as well as more loss in spatial resolution. On the rectum zone, the highest increase in the $\langle \text{PCORR} \rangle_{\text{patient}}$ metric is observed for patient #4 (1.3% for $M = 3$ and 5.2% for $M = 25$) and corresponds to an increasing ratio of approximately 410%. It can be observed that the higher the $\langle \text{PCORR} \rangle_{\text{patient}}$ metric is, the higher its increase is when M varies from 3 to 25. On the same range of values for the parameter M and for both zones, the $\langle \text{TSTD} \rangle_{\text{patient}}$ metric's variations are low when compared to PCORR's variations and mostly exhibit slow decrease when M is increased. In some cases (mostly for patient #2), this metric is slightly increasing with M , especially when M is higher than 16.

For the rectum zone, the highest decrease in the $\langle \text{TSTD} \rangle_{\text{patient}}$ metric is observed for patient #6 (2.52 °C for $M = 3$ and 2.21 °C for $M = 16$) and corresponds to an improvement ratio of 13%.

4. DISCUSSION

The design of this filter, based on the thresholding of a fluctuation criterion makes it a generic approach to detect artifacts manifested by phase variations higher than in a predefined area such as the heating zone in the case of thermotherapy. By simply considering the standard deviation of the first derivative of the temperature, the filter's detection process is able to account for many sources of variations without any assumptions on their nature. The design of the filter also enables it to be applied without any manual intervention or tuning, in so far as a dynamically estimated threshold is used to try to discriminate between changes due to artifact and heating. This enables it to respond to any sudden or random changes in temperature that exceed the noise threshold but also are outside the heated region. This type of change is often associated with the motion of gas in the rectum during prostate HIFU.

As described in the text, the algorithm also corrected voxels outside the region of heating in the prostate. The majority of these artifacts were caused by blood flow and respiratory motion of the abdomen. The high PCORR metric in Fig. 6 for patients #1, #4, and #5 include detection due to blood flow and respiratory motion of the abdomen. In the prostate zone, this PCORR metric better reflects the rectum air bubble artifact and was the highest for patients 1 and 6.

The peripheral slices that were not targeted by the ultrasound heating beam are useful for monitoring potential thermal damage in the surrounding of the targeted zone in the

prostate. The filter design using the closest heated slice to calibrate the filter's strength in the slices in the direct neighborhood of a heated slice led to results very similar to those obtained the heated slices when comparing the influence of the filter ($\langle \text{TDIFF} \rangle$ metric) in specific zones. The $\langle \text{TDIFF} \rangle$ metric averaged over the neighbor slices for all patients was less than 16% lower than the same metric for the heated slices in the rectum zone, less than 1% lower in the prostate posterior zone, and were both under 0.25°C in the heated and prostate anterior zones.

The size of the sliding window (M) used to estimate the fluctuation criterion [see Eq. (8)] is an important parameter and influences the performance of the filter in a number of ways. Shorter windows will lead to a quicker response time in the detection and a lower averaging effect on the criterion which will help in the detection of brief artifacts. Longer windows will have a higher averaging effect on the criterion and will thus help in discriminating the motion artifact from the background noise in so far as its threshold $F2_{\max}$ is a function

inversely proportional to the square root size of the sliding window [see Eq. (11)].

The results of the study on the effect of the size (M) of the sliding window on the performance of the filter conducted on the five patients' data, and presented on Fig. 8 led to the choice of $M = 3$ since the performance of the filter (TSTD metric) has not exhibited significant change with this varying parameter M . This choice of a lower M value also corresponds to a more conservative filter in so far as it allows keeping the PCORR metric as low as possible in order to prevent its associated loss in spatial resolution. It has to be noted that this optimized window size is valid in our specific case with an average measurement temperature uncertainty of 1.2°C across all patients. In the case of a higher level of measurement temperature uncertainty relatively to the heating temperature level, a longer window may have performed better through its noise averaging effect. Also, independently from the windows size choice, the higher the measurement

FIG. 8. Effect of the window size M [see Eq. (8)] on the performance of the filter. Comparison of the metrics $\langle \text{PCORR} \rangle_{\text{patient}}$ and $\langle \text{TSTD} \rangle_{\text{patient}}$ on the rectum zone [left: (A) and (B)], and on the global zone [right: (C) and (D)] when M is varying. The $\langle \text{TSTD} \rangle_{\text{patient}}$ metric given on the ordinate axis of (B) and (D) corresponds to the results obtained without the filter.

temperature uncertainty is the less effective this filter's artifact detection is expected to be in so far as the artifact's fluctuations become more and more comparable to the measurement noise related fluctuations.

The PRF thermometry method consists of building the temperature maps from the successive variations acquired at each dynamic [see Eq. (2)]. The correction of the artifact-detected zones at each dynamic aims at reducing the step by step propagation of the artifact during the treatment.

The first step leads to elimination of an invalid fluctuation with regard to the fluctuation criterion (CR). Figure 3(B) illustrates the fact that the cancellation of the current variation successfully removes the artifact but introduces noise due to the multiplication of the phase reference occurring each time the current variation is set to zero. Considering that each measured phase map φ_k is corrupted by some additional noise n_k , the calculation of the temperature expressed in Eq. (2) can, in the absence of correction, be rewritten

$$T_m = \frac{1}{\alpha \cdot \gamma \cdot T_e \cdot B_0} \left(\sum_{k=1}^m (\varphi_k - \varphi_{k-1}) + \sum_{k=1}^m (n_k - n_{k-1}) \right) + T_0, \quad (25)$$

and the term related to noise simplifies such as

$$\sum_{k=1}^m (n_k - n_{k-1}) = n_m - n_0, \quad (26)$$

with n_k being the noise component at dynamic k . The standard deviation of T_m can then be estimated by

$$\sigma(T_m) \approx \frac{\sqrt{2}}{\alpha \cdot \gamma \cdot T_e \cdot B_0} \sigma_\varphi, \quad (27)$$

with σ_φ being the standard deviation of the acquired phase maps.

If the number of canceled dynamics N_c is higher than zero, the term related to noise reaches its lower limit (respectively, upper limit) when all (respectively, none) of the cancellations happen at consecutive dynamics, and the standard deviation of T_m can be bounded such as

$$\frac{2}{\alpha \cdot \gamma \cdot T_e \cdot B_0} \sigma_\varphi \leq \sigma(T_m) \leq \frac{\sqrt{2+N_c}}{\alpha \cdot \gamma \cdot T_e \cdot B_0} \sigma_\varphi \quad (28)$$

with N_c being the number of canceled dynamics.

Equation (28) shows that the cancellations of the temperature variations related to the first step of the filter's correction procedure [see Eqs. (15) and (16)] lead to a degradation of the standard deviation of a factor of $\sqrt{2}$ in the best case (all cancellations are consecutive) which can grow with $\sqrt{N_c}$ in the worst case (no cancellations are consecutive).

According to Fig. 5(D), the percentage of correction exceeds 5% in most regions where temperature artifacts are present. This 5% threshold corresponds to approximately 16 dynamics for which a new phase reference is used. Equation (28) shows that the additional noise corresponding to 16 corrected dynamics could, in the worst case, be three times higher than the noise observed on an uncorrected temperature map. The latter [see Eq. (27)] being estimated at 1.2°C, the resulting standard deviation of the temperature would thus exceed 3.63°C in the zone where PCORR is higher than 5%.

The second step [described by Eq. (16)] has two major consequences on the filtered voxels. First, it limits the noise introduced by the cancellation of the current temperature variation at step 1 by a factor varying between 1 and 9 depending on the number of SNR masked voxels in its direct neighborhood. The average number of valid neighbors N_{VN} for patient #6, slice 3 has been estimated to 7.48 which results in, for example, an improved standard deviation of the temperature of 1.75°C (compared to 3.63°C) in the region where PCORR is equal to 5%. The hybrid correction of the temperature provides a much lower and smooth temperature standard deviation in the artifacted regions [see Figs. 3(A)–3(C) and Figs. 5(A)–5(C)], even if this second correction step averaging procedure also implies a spatial smoothing effect. This loss of spatial resolution is proportional to the percentage of correction in a given area. It will lead to a spatial smoothing of the temperature spreading across multiple voxels if the correction is applied at multiple successive dynamics in the same zone.

The application of this filter in real time is quite feasible as it is causal and as the average calculation time observed for the five patients with a MATLAB implementation was around 0.05 s per slice for the detection and correction tasks combined.

While the filter's original design is suited to the lowering of the temperature artifacts due to the motion of gas bubbles, it is not well-suited for correcting slow deformation of the prostate, for example, as the associated phase variation will not be detected in so far as it will be negligible compared to the noise or the heating.

The presence of moving air bubbles inside the hottest area of the heating region would result in a misquantification of coefficient A in Eq. (10), leading to a failure in the detection of this artifact. Indeed, the latter's temperature variation would be mistaken for real heating. Possible improvements of the filter in this case could consist in constraining the heating threshold $F1_{\max}$ by introducing some more specific knowledge of the ultrasonic applicator heating. Nonetheless, the current implementation may not be suitable for removing artifacts around the rectum when the ultrasound beam passes directly through this region.

Other application cases where this filter might be useful include externally applied HIFU treatments where there might be bowel-related artifacts during sonication, comprising the treatment of rectal, bone, liver, and kidney cancer. Different types of HIFU treatments related artifacts could be potentially addressed with this filter in that they feature sudden, random motion characteristics such as patient muscle contraction or displacement of the air corner formed by the interface between the coupling gel pad and the patient. If the removal of those type of artifacts can be useful for the typically short (30–60 s) scanning associated to ablation, it seems even more important for longer hyperthermia treatments where the errors accumulating over the treatment duration have to be kept around 1–2°C.

5. CONCLUSION

During MRI-controlled transurethral ultrasound therapy of the prostate gland, the accuracy of MR thermometry is critical for precise online feedback control as well as for monitoring potential thermal injury of surrounding tissues, especially the rectum wall. A filter based on the temporal variance of the temperature and using empirical and dynamic positional knowledge of the ultrasonic heating beam in the prostate tissue has been described and tested by postprocessing of MR thermometry data from five human transurethral ultrasound treatments. Most of the temperature artifacts due to the presence of moving air bubbles in the rectum have been detected and removed. A quantitative estimation of the filter capabilities showed a systematic improvement in the standard deviation of the corrected temperature maps in the rectum zone as well as in the entire acquired slice.

ACKNOWLEDGMENTS

The authors wish to acknowledge financial support from the National Institutes of Health (1R21CA159550) and Fed-DEv Ontario (Technology Development Program). The authors also acknowledge the technical support provided by Philips MR Finland with respect to the processing of MR temperature maps acquired on the Philips MRI.

^{a)} Author to whom correspondence should be addressed. Electronic mail: aschmitt@sri.utoronto.ca

¹K. Tang *et al.*, "Conformal thermal therapy using planar ultrasound transducers and adaptive closed-loop MR temperature control: Demonstration in gel phantoms and ex vivo tissues," *Phys. Med. Biol.* **52**, 2905–2919 (2007).

²R. Chopra *et al.*, "Analysis of the spatial and temporal accuracy of heating in the prostate gland using transurethral ultrasound therapy and active MR temperature feedback," *Phys. Med. Biol.* **54**(9), 2615–2633 (2009).

³W. H. Nau, C. J. Diederich, A. B. Ross, V. Rieke, K. Butts, and G. Sommers, "Evaluation of endorectal and urethral cooling devices during MR-guided ultrasound thermal ablation in canine prostate," *Conf. Proc. IEEE Eng. Med. Biol. Soc.* **4**, 2492–2495 (2004), <http://www.ncbi.nlm.nih.gov/pubmed/17270778>.

⁴R. Chopra, N. Baker, V. Choy, A. Boyes, K. Tang, D. Bradwell, and M. J. Bronskill, "MRI-compatible transurethral ultrasound system for the treatment of localized prostate cancer using rotational control," *Med. Phys.* **35**(4), 1346–1357 (2008).

⁵R. Chopra, A. Colguhoum, M. Burtnyk, W. A. N'Djin, I. Kobelevskiy, A. Boyes, K. Siddiqui, H. Foster, L. Sugar, M. A. Haider, M. Bronskill, and L. Klotz, "MR imaging-controlled transurethral ultrasound therapy for conformal treatment of prostate tissue: Initial feasibility in humans," *Radiology* **265**(1), 303–313 (2012).

⁶Y. Ishihara, A. Calderon, H. Watanabe, K. Okamoto, Y. Suzuki, K. Kuroda, and Y. Suzuki, "A precise and fast temperature mapping using water proton chemical shift," *Magn. Reson. Med.* **34**(6), 814–823 (1995).

⁷H. Cline, J. Schenck, K. Hynynen, R. Watkins, S. Souza, and F. Jolesz, "MR-guided focused ultrasound surgery," *J. Comput. Assist. Tomogr.* **16**(6), 956–996 (1992).

⁸L. J. Pisani, A. B. Ross, C. J. Diederich, W. H. Nau, F. G. Sommer, G. H. Glover, and K. Butts, "Effects of spatial and temporal resolution for mr image-guided thermal ablation of prostatic with transurethral ultrasound," *J. Magn. Reson. Imaging* **22**, 109–118 (2005).

⁹E. Ramsay, C. Mougenot, M. Kohler, M. Bronskill, L. Klotz, M. A. Haider, and R. Chopra, "MR thermometry in the human prostate gland at 3.0T for transurethral ultrasound therapy," *J. Magn. Reson. Imaging* **38**, 1564–1571 (2013).

¹⁰B. D. de Senneville, B. Quesson, and C. T. W. Moonen, "Magnetic resonance temperature imaging," *Int. J. Hyperthermia* **21**(6), 515–531 (2005).

¹¹S. Moricawa, T. Inubushi, Y. Kurumi, S. Naka, V. Seshan, and T. Tsukamoto, "Feasibility of simple respiratory triggering in MR-guided interventional procedures for liver tumors under general anesthesia," *ISMRM, 10th Annual Meeting*, Hawaii (2002).

¹²B. D. de Senneville, B. Quesson, P. Desbarats, C. Mougenot, and C. T. W. Moonen, "Image processing for on-line reduction of thermometry artifacts," *Proceedings of the 14th Annual Meeting of ISMRM (abstract 1425)*, Seattle, WA (2006).

¹³B. D. de Senneville, C. Mougenot, and C. T. W. Moonen, "Real-time adaptive methods for treatment of mobile organs by MRI-controlled high-intensity focused ultrasound," *Magn. Reson. Med.* **57**(2), 319–330 (2007).

¹⁴S. Roujol, M. Ries, B. Quesson, C. Moonen, and B. D. de Senneville, "Real-time MR-thermometry and dosimetry for interventional guidance on abdominal organs," *Magn. Reson. Med.* **63**(4), 1080–1087 (2010).

¹⁵S. Hey, G. Maclair, B. D. De Senneville, M. Lepetit-Coiffe, Y. Berber, M. O. Köhler, B. Quesson, C. T. Moonen, and M. Ries, "Online correction of respiratory-induced field disturbances for continuous MR-thermometry in the breast," *Magn. Reson. Med.* **61**(6), 1494–1499 (2009).

¹⁶V. Rieke, K. K. Vigen, G. Sommer, B. L. Daniel, J. M. Pauly, and K. Butts, "Referenceless PRF shift thermometry," *Magn. Reson. Med.* **51**, 1223–1231 (2004).

¹⁷W. A. Grissom, V. Rieke, A. B. Holbrook, Y. Medan, M. Lustig, J. Santos, M. V. McConnell, and K. B. Pauly, "Hybrid referenceless and multibaseline subtraction MR thermometry for monitoring thermal therapies in moving organs," *Med. Phys.* **37**(9), 5014–5026 (2010).

¹⁸V. Rieke, A. M. Kinsey, A. B. Ross, W. H. Nau, C. J. Diederich, G. Sommer, and K. B. Pauly, "Referenceless MR thermometry for monitoring thermal ablation in the prostate," *IEEE Trans. Med. Imaging* **26**(6), 813–821 (2007).

¹⁹K. B. Pauly, C. J. Diederich, V. Rieke, D. Bouley, J. Chen, W. H. Nau, A. B. Ross, A. M. Kinsey, and G. Sommer, "Magnetic resonance-guided high-intensity ultrasound ablation of the prostate," *Top. Magn. Reson. Imaging* **17**, 195–207 (2006).

²⁰R. Chopra, M. Burtnyk, M. A. Haider, and M. J. Bronskill, "Method for MRI-guided conformal thermal therapy of prostate with planar transurethral ultrasound heating applicators," *Phys. Med. Biol.* **50**, 4957–4997 (2005).

²¹M. Burtnyk, W. A. N'Djin, I. Kobelevskiy, M. Bronskill, and R. Chopra, "3D conformal MRI-controlled transurethral ultrasound prostate therapy: Validation of numerical simulations and demonstration in tissue-mimicking gel phantoms," *Phys. Med. Biol.* **55**(22), 6817–6839 (2010).

²²M. Burtnyk, R. Chopra, and M. Bronskill, "Simulation study on the heating of the surrounding anatomy during transurethral ultrasound prostate therapy: A 3D theoretical analysis of patient safety," *Med. Phys.* **37**, 2862–2875 (2010).

²³B. D. De Senneville, C. Mougenot, B. Quesson, I. Dragonu, N. Grenier, and C. T. W. Moonen, "MR thermometry for monitoring tumor ablation," *Eur. Radiol.* **17**(9), 2401–2410 (2007).

²⁴N. Todd, M. Diakite, A. Payne, and D. L. Parker, "In vivo evaluation of multi-echo hybrid PRF/T1 approach for temperature monitoring during breast MR-guided focused ultrasound surgery treatments," *Magn. Reson. Med.* **69**(1), 62–70 (2013).

²⁵A. M. El-Sharkawy, M. Schär, P. A. Bottomley, and E. Atalar, "Monitoring and correcting spatio-temporal variations of the MR scanner's static magnetic field," *Magn. Reson. Mater. Phys., Biol. Med.* **19**(5), 223–236 (2006).