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A New Method for Temperature-Field Reconstruction during Ultrasound-Monitored Cryosurgery using Potential-Field Analogy

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Abstract

The current study aims at developing computational tools in order to gain information about the thermal history in areas invisible to ultrasound imaging during cryosurgery. This invisibility results from the high absorption rate of the ultrasound energy by the frozen region, which leads to an apparent opacity in the cryotreated area and a shadow behind it. A proof-of-concept for freezing-front estimation is demonstrated in the current study, using the new potential-field analogy method (PFAM). This method is further integrated with a recently developed temperature-field reconstruction method (TFRM) to estimate the temperature distribution within the frozen region. This study uses prostate cryosurgery as a developmental model and trans-rectal ultrasound imaging as a choice of practice. Results of this study indicate that the proposed PFAM is a viable and computationally inexpensive solution to estimate the extent of freezing in the acoustic shadow region. Comparison of PFAM estimations and experimental data shows an average mismatch of less than 2 mm in freezing-front location, which is comparable to the uncertainty in ultrasound imaging. Comparison of the integrated PFAM+TFRM scheme with a full-scale finite-elements analysis (FEA) indicates an average mismatch of 0.9 mm for the freezing front location and 0.1 mm for the lethal temperature isotherm of -45°C . Comparison of the integrated PFAM+TFRM scheme with experimental temperature measurements show a difference in the range of 2°C and 6°C for selected points of measurement. Results of this study demonstrate the integrated PFAM+TFRM scheme as a viable and computationally inexpensive means to gain information about the thermal history in the frozen region during ultrasound-monitored cryosurgery.

Keywords

Cryosurgery; Bioheat; Simulation; Imaging; Ultrasound

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1. Introduction

Prostate cryosurgery is an alternative surgical procedure to destroy a portion or the entire gland by freezing. Prostate cryosurgery is most frequently performed under ultrasound imaging guidance, by means of a trans-rectal ultrasound transducer (TRUST) [6,11,12]. This minimally invasive procedure is performed by means of cryoprobes, each in the shape of a long hypodermic needle with a cylindrical cooling surface extending backwards from its tip. During prostate cryosurgery, a number of cryoprobes are inserted such that their active cooling surfaces are strategically placed within the gland in order to match frozen region shape with the organ contour (i.e., the target region) [23,25,26]. In addition, a urethral warmer is used during prostate cryosurgery to protect the urethra from freezing injury. While prostate cryosurgery represents only one application of minimally invasive cryosurgery, it is arguably one of the most challenging applications of its kind, due to the large number of simultaneously operating cryoprobes and difficulties with TRUST imaging. The current study uses prostate cryosurgery as a developmental model and TRUST imaging as a choice of practice.

The minimally invasive nature of the clinical procedure poses many technical difficulties, some of which can be alleviated by means of computational tools. Those difficulties can be conveniently classified as being related to either surgical planning [22,26] or real-time monitoring [6,7,12,21]. With these technical difficulties in mind, cryosurgery success is highly dependent on the skill and personal experience of the clinician. The current study addresses difficulties in cryosurgery monitoring by developing a computational approach to improve the reconstruction of the frozen-region shape in real time.

The primary challenge with ultrasound imaging for cryosurgery is its wave reflection from solid surfaces, such as the cryoprobe shell and the freezing front. As a consequence, the frozen region appears opaque on the ultrasound monitor, while acoustic shadowing is formed behind the frozen region [12]. Using TRUST, the freezing front next to the rectal wall can be clearly observed during prostate cryosurgery, but no imaging information may be obtained about the remaining portion of the freezing front. This lack of information may affect the quality of the medical treatment and lead to post-operational complications. The current study aims at that portion of the frozen region hidden from the ultrasound transducer.

While the freezing front location is the controlled parameter during clinical practice, with the onset of freezing around 0°C, clinical interest extends to lower temperatures as well. It is widely accepted that a lethal temperature threshold must be surpassed in order to maximize cryoinjury, with a commonly assumed value typically in the range of -40°C and -50°C [6]. Since temperature measurements during prostate cryosurgery are rarely done (at one or a couple of locations, if at all), the location of the lethal temperature during clinical practice represents a speculation at best. The current study is further aimed at improving the prediction of the location of the lethal-temperature isotherm.

The current study presents a proof-of-concept for temperature-field reconstruction during minimally invasive cryosurgery, using a potential-field analogy method (PFAM). The current study aims at ultrasound-monitored prostate cryosurgery, and builds upon a recently

developed temperature-field reconstruction method (TFRM) [27]. The current study presents: (i) a numerical technique to estimate the location of the freezing front in the acoustic shadowed region, (ii) an integrated approach to reconstruct the temperature field, and (iii) validation of the integrated method with published experimental data [18].

2. Mathematical Formulation

2.1 A new potential field analogy method (PFAM)

This study presents a new method to estimate the freezing front location in the acoustic shadowed area during prostate cryosurgery, as schematically illustrated in Fig. 1. The proposed method is based on following principles:

- a. The temperature field within the simulated cryosurgery domain is analogous to an electric-potential field, where isotherms are analogous to isoelectric lines.
- b. Cryoprobes are analogous to static electric charges.
- c. Thermal inertia effects are negligible, where the problem is essentially a quasi-steady problem [27].
- d. The location of the cryoprobes and their thermal history are known.
- e. The location and temperature of the urethral warmer are known.
- f. The location of some portion of the freezing front is known, as illustrated in Fig. 1(b).
- g. The problem is solved in two dimensions (2D), representing the prostate cross-section displayed on the ultrasound monitor. While no limitation exists on extending the PFAM to three dimensions (3D), the 2D formulation represents a choice of practice for method demonstration.

The potential field in presence of a number of static electric charges is given by [10]:

$$V(x, y) = \alpha \sum_{i=1}^m \frac{q_i}{r_i} \quad (1)$$

where α is a field constant, q_i is magnitude of the i^{th} charge, r_i is distance from the i^{th} charge to a field point of interest (x, y) , and m is the total number of charges.

In analogy, a quasi-steady, pseudo-temperature field is modeled by:

$$\theta(x, y) = \kappa \sum_{i=1}^n \frac{c_i}{r_i} \quad (2)$$

where κ is a field constant, c_i is a cooling constant of the i^{th} cryoprobe, and n is the total number of cryoprobes. In principle, c_i may be time dependent, temperature dependent,

cooling-rate dependent, or exhibit any other implicit behavior. Nevertheless, since Eq. (2) is applied on an ultrasound snapshot, and since c_i is evaluated based on data from the same snapshot, an *a priori* knowledge of the functional behavior of c_i is not needed.

The strategy for evaluating the instantaneous values of c_i in the current study relies on the conditions that all cryoprobes are operated simultaneously, and that all cryoprobes have similar physical dimensions and cooling capabilities. These similarities suggest c_i values in a close range for all cryoprobes. Hence, a simplifying assumption is made in the current study that all c_i are actually identical at any given point in time, where the implications of this assumption are discussed below. For this simplified case, the pseudo-temperature field becomes:

$$\theta(x, y) = \beta \sum_{i=1}^n \frac{1}{r_i} \quad ; \quad \beta \equiv c_i \kappa \quad (3)$$

where the value of β is selected such that one of the isotherms of the pseudo-temperature field optimally matches the US-visible portion on the freezing front. This is essentially a minimization problem, characterized by an explicit set of algebraic equations.

2.2 Application of a temperature-field reconstruction method (TFRM)

In principle, the PFAM allows to estimate the freezing front location, as presented in the results and discussion section below. However, in order to gain more information about the temperature field within the frozen region, a recently developed quasi-steady temperature-field reconstruction method (TFRM) [27] can be integrated. In particular, the location of the so-called lethal temperature isotherm may be of particular interest to the clinician. While many alternatives can be chosen to reconstruct the quasi-steady temperature field, the algorithm presented in [27] is potentially suitable for real-time simulations and represents a choice of practice. This algorithm is presented here in brief only, for the completeness of presentation, while the reader may refer to [27] for more detail.

It is customary to assume that heat transfer during cryosurgery can be modeled with the classical bioheat equation [13]:

$$C_t \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) + w_b C_b (T_b - T) + q_{met} \quad (4)$$

where C_t is the volumetric specific heat of the tissue, T is the temperature, t is the time, k is the thermal conductivity, w_b is the blood perfusion rate, C_b is the volumetric specific heat of the blood, T_b is the blood temperature entering the treated region, and q_{met} is the metabolic heat generation. It has been previously suggested [15] that the bioheat transfer equation is suitable for prostate cryosurgery, in the absence of major blood vessels, and that the metabolic heat generation effect is negligible during cryosurgery [15]. Typical thermo-physical properties for Eq. (4) are listed in Table 2.

The TFRM is based on the fact that the heat transfer problem during cryosurgery is characterized by a small Stefan number:

$$St = \frac{C_t \Delta T}{H} \quad (5)$$

where T is the maximum temperature difference within the frozen region and H is the latent heat of freezing. The heat transfer process is primarily dominated by the volumetric latent heat effect when Stefan number is small and, hence, the transient term in Eq. (4) may be neglected [21]. This observation also supports the underlying principles of the PFAM, where inertia terms are neglected. Nevertheless, the overall transient nature of freezing process is not neglected, but introduced through boundary conditions. These boundary conditions may be external, such as at the outer surface of the domain, the cryoprobe surface, or the urethral warmer surface. These boundary conditions may also be internal such as at the imaged freezing front [27]. Data collected from tissue-embedded temperature sensors can be used for model verification but cannot be regarded as a boundary condition [27].

The TFRM attempts to use all the available data in order to evaluate the instantaneous temperature field, using a modified Laplacian interpolation scheme based on the following explicit formulation [27]:

$$0 = \sum_{l,m,n} \frac{T_{l,m,n} - T_{i,j,k}}{R_{l,m,n-i,j,k}} + (w_b C_b)_{i,j,k} (T_b - T_{i,j,k}) \quad (6)$$

where i,j,k are spatial numerical grid indices, l,m,n are indices of the neighboring grid points, and R is the thermal resistance to heat transfer by conduction between grid point i,j,k and its neighbor l,m,n . For a regular Cartesian geometry, the thermal resistance to heat conduction can be simplified as:

$$R_{l,m,n-i,j,k} = \left[\frac{\Delta \eta}{2kA} \right]_{l,m,n} + \left[\frac{\Delta \eta}{2kA} \right]_{i,j,k} \quad (7)$$

where η is the distance between the point of interest and its neighbor, and A is the representative cross-sectional area perpendicular to the direction of heat conduction. Since the thermophysical properties in the tissue are temperature dependent (Table 2), Eqs. (6)–(7) are solved using an iterative predictor-corrector technique.

2.3 Integration of PFAM with TFRM

Figure 2 displays a schematic illustration of a computation framework to integrate the proposed PFAM with the established TFRM. With reference to Fig. 2, input data may come from various sources. In a clinical procedure, partial freezing front location would come from ultrasound imaging, cryoprobe thermal histories would come from embedded

temperature sensors at the cryoprobe tips, urethral warmer temperature would come from the warming device, and the layout of cryoprobes and warmer would come from preplanning or ultrasound registration. For the bench-top setup described in the next section, the partial freezing front is simulated from experimental data (discussed below), the cryoprobe thermal histories are measured similar to clinical practice, and the cryoprobe layout is set with a special placement grid.

The framework in Fig. 2 is based on two stages. In the first stage, partial freezing front location and cryoprobe thermal histories are used to estimate the hidden portion of the freezing front using the PFAM. Being at the center of the frozen region, the urethral warmer data is not needed for that estimation. In the second stage, the evaluated freezing front location in the acoustic shadow region is integrated with all other input data, in order to evaluate the temperature field in the domain by means of TFRM. Note that TFRM has already been validated for a complete or partial freezing-front contour [27], while the current study focuses on the improvement of this scheme by creating additional data by means of PFAM.

3. Experimental Verification

While experimental verification with clinical results would serve as the ultimate test, no such data is available for the current study. In particular, frozen region scanning in combination with TRUST registration are major challenges in clinical practice. Instead, the experimental verification in the current study is based on a bench-top physical simulator of the procedure [18]. This simulator has been created to validate computer simulations of the cryoprocure [18], and to verify computerized planning of prostate cryosurgery [19]. The experimental system and image analysis techniques have been presented in detail previously [18] and are overviewed here in brief, for the completeness of presentation.

3.1 Experimental setup

The experimental system consists of a transparent phantom material container, an array of proprietary liquid nitrogen-based cryoprobes (1.3 mm OD) [18], and a proprietary electrical urethral warmer (7 mm OD) also termed *cryoheater* [16]. A side view of the phantom-embedded cryoprobes and cryoheater is displayed in Fig. 3 (parallel to the cryoprobe axes). The container is made of thick Plexiglas walls (5 mm), having overall dimensions of 25 mm $\times$ 200 mm $\times$ 200 mm. In particular, the internal width of the container equals to the length of the active surface of the specific cryoprobes (15 mm), such that the heat transfer process is essentially 2D in a plain transverse to the cryoprobe axes (see [18] for detailed thermal analysis). The heater and cryoprobes are inserted through special holes in the container, which enable variable cryoprobe layouts. The combined heat transfer process generated by the cryoprobes and cryoheater is simulative of the largest cross-section of the prostate during cryosurgery.

The image displayed in Fig. 3 is a snapshot from a high-resolution movie recorded throughout the experiment. The phantom material displayed in Fig. 3 is distilled water mixed with 1.3% gelatin and blue food dye. Once cured, the gelatin prevents natural convection currents in the distilled water, and pure heat conduction process is created in a

well characterized material. The food dye is used as a visual enhancement means, where the frozen dyed water changes color from blue to white.

Up to six cryoprobes can be used concurrently in the specific setup, which is fed by a portable liquid nitrogen container (1.8 l) and powered by compressed air (30 psi) [18]. The resulting thermal history at the cryoprobes is a cooling rate of about 33°C/min in the first three minutes of operation and a minimum operation temperature of about -110°C. The actual thermal history for each cryoprobe is measured during experimentation by means of thermocouples embedded at the cryoprobe-tips, and the continuous data streams are recorded with a computerized data logger. Urethral warming in the current setup is simulated with the self-controlled cryoheater set to a constant temperature of 25°C, which equals to the initial temperature of the phantom material. In selected experiments, clinical-grade hypodermic thermocouples (type T, 1.5 mm OD) are also strategically placed within the domain, in order to measure the thermal history at specific locations for comparison with simulation results.

3.2 Simulating ultrasound imaging from experimental data

The current study uses previously generated frozen region reconstructions from selected experiments [18]. In brief, individual images from the recorded video of each experiment were captured and post-processed. Reconstruction of the frozen region from this image was performed in four stages: (i) manual registration by identifying the locations of each cryoprobe and the cryoheater, (ii) applying selected MATLAB filters for image enhancement, (iii) segmentation by a MATLAB implementation of a region-growing method [3,14], and (iv) application of a dilation transformation [24] to increase the quality of the reconstruction frozen region.

Figure 3 displays a representative captured image from a recorded movie, where the inset on the top-right displays the reconstructed frozen region. That inset was used to extract the freezing front, previously displayed in Fig. 1 to discuss the acoustic shadowing effect. For the purpose of the current study, the TRUST image location was assumed such that one-third of the frozen region contour becomes 'visible', Fig. 1(b).

3.3 Comparison approach of numerical results with experimental data

Comparison of numerical results with experimental data is based on (i) the evolution of the frozen-region shape over time, and (ii) the thermal history at pre-selected points in space. For the comparison of the frozen-region shape, the PFAM predictions are compared with the freezing front location reconstructed from experimental data. The mismatch between those curves is quantified by the so-called Hausdorff distance, which is commonly used in medical imaging [2], curves analysis [1], and computer vision [8].

In brief, the Hausdorff distance between two curves is defined as the maximum distance from one curve to the nearest point on the other curve [20]. For two sets of points representing curves $A = \{a_1, a_2, \dots, a_p\}$ and $B = \{b_1, b_2, \dots, b_q\}$, the Hausdorff distance from A to B is given by:

$$H(A, B) = \max_{a \in A} \min_{b \in B} \|a - b\| \quad (8)$$

where an Euclidean norm is used to calculate $\|a - b\|$ [22]. The current study also makes use of a modified Hausdorff distance, which represent an average distance:

$$H(A, B)_{\text{mod}} = \frac{1}{|A|} \sum_{a \in A} \min_{b \in B} \|a - b\| \quad (9)$$

where $|A|$ is the number of points in A . The modified Hausdorff distance is also widely used in image processing applications [4]. Since the modified Hausdorff distance is an average quantity, it decreases the impact of outliers and hence it is more useful in pattern comparing applications. In this study, S_{max} stands for $H(A, B)$ where A is the estimated freezing front location and B is the actual (observed) freezing front location. Consistently, S_{avg} stands for the modified Hausdorff distance, $H(A, B)_{\text{mod}}$.

4. Results and Discussion

Validation of the proposed PFAM and its integration with the TFRM are studied in three steps representing increasing-complexity: (i) comparison of the freezing front location based on the PFAM with experimental results of simplified cases; (ii) comparison of the freezing front location based on the PFAM with simulated ANSYS results of more complex cases, and (iii) comparison of the estimated thermal history based on the integrated PFAM and TFRM scheme with experimental data.

4.1 Experimental validation of PFAM

Experimental validation of the PFAM is performed for three typical cases. Case A represents freezing process as a result of four cryoprobes symmetrically arranged around a urethral warmer, Fig. 4. Note that the experimental results from Fig. 4(a) are extracted from Fig. 3. Cases B and C represent similar results for six cryoprobes in an increasing layout complexity, with results displayed in Figs. 5 and 6, respectively. Table 1 lists mismatch values between experimental data and PFAM results for cases A–C.

Note that the PFAM is applied only after the merge of the frozen regions around the individual cryoprobes into a continuous frozen region around the urethral warmer, which is typical in clinical practice at an advanced stage of the procedure. The planned cryoprobe layout in clinical practice is aimed at creating such a continuous region, otherwise significant areas within the prostate would be left untreated [16,23,26]. Further note that neither of the cryoprobe layouts is considered optimal for prostate cryosurgery, which is only selected for a proof-of-concept study. Hence, while the principles of acoustic shadowing have been maintained consistent with clinical practice, the actual selection of the visible portion of the freezing front is somewhat arbitrary.

From the results displayed in Figs. 4–6 and Table 1, the following observations can be made:

1. The best match between PFAM results and experimental data is obtained for the symmetric Case A. As could be expected, the quality of this match degrades with the increased complexity of the cryoprobe layout.
2. Regardless of the specific cryoprobe layout, the match between PFAM results and experimental data improves as the freezing process progresses. This trend can be explained by the observation that the frozen region becomes smoother with less concaved regions with the progression of freezing.
3. Regardless of the specific configuration tested, an average Hausdorff distance of less than 2 mm is found after 400 s., and less than 1.2 mm after 1000 s. This value is equal or better than the typical uncertainty in ultrasound imaging, which suggests that the PFAM is adequate for clinical operation when a real-time application is developed.

In order to put the above observations in the right context, estimating the freezing front location in the acoustic shadowed area represents an unmet need. The PFAM represents an alternative approach to meet this need, and real-time operation appears feasible. However, reducing the PFAM to clinical practice requires further development, and the current study is aimed only at developing a low computation-cost method to meet that need. Reducing this method to practice can take advantage on a recently developed method for real-time segmentation of the visible portion of the frozen region [5].

4.2 Computer validation of the integrated PFAM+TFRM scheme

While the PFAM has shown its effectiveness in approximating the freezing-front location in areas where imaging analysis may prove ineffective, temperature-field reconstruction within the frozen region calls for additional computation measures. In particular, the location of lethal-temperature isotherm is often desired but no physical artifact can serve as its indicator, which calls for the proposed integrated scheme to be validated with robust bioheat transfer simulation tools. For this purpose, ANSYS is selected to generate a benchmark case consistent with a previous study [27] (Case D), which is based on eight cryoprobes and a urethral warmer. The simulated domain is assumed infinite from heat transfer consideration, having an initial temperature of 37°C. All cryoprobes are assumed to cool simultaneously from to -145°C in 30 s (simulative of Argon-based cryoprobes), while the urethral warmer is maintained at 37°C throughout the simulation. Thermophysical properties used in the current analysis are listed in Table 2.

Validation results for the integrated PFAM and TFRM scheme are displayed in Figs. 7 and 8, with mismatch distances summarized in Table 1. It can be seen from Fig. 7 that the PFAM dramatically improves the prediction of the freezing front location (the jagged nature of the curves is consistent with the 1 mm grid used for TFRM evaluation). The maximum and average mismatch distances for the TFRM alone are 15.2 mm and 8.1 mm, respectively. However, the maximum and average mismatch distances for the combined PFAM+TFRAM are 2.9 mm and 0.9 mm, respectively. Taking into account an ultrasound analysis uncertainty of 1 mm to 2 mm, the combined PFAM+TFRAM scheme appears to be a clinically relevant mathematical scheme for real-time analysis.

The maximum and average mismatch distances for the lethal temperature isotherm are 10.1 mm and 5.3 mm, respectively, when only TFRM is used solely. However, when the integrated mathematical scheme is applied, these values decrease to 3.1 mm and 0.1 mm, respectively. These results appear quite relevant for the development of computer-assisted tools for cryosurgery, especially when emphasizing that they are based on only about one third of the freezing-front location (PFAM), while neglecting the inertia term in the bioheat transfer process (TFRM).

The integrated scheme has been solved using MATLAB on an Intel® Quad Core™ i7 machine with 9GB RAM, running at 3.07GHz. The PFAM is computationally inexpensive, which required less than 0.5 s for any of the cases presented above. As described in [27], the TFRM is solved by applying a sparse matrix format and internal solvers for efficiency in calculations, with a typical runtime under 3 s. These runtime numbers represent non-optimized computation frameworks, where GPU implementation is likely to reduce the overall runtime to milliseconds [9]. Hence, the integrated numerical scheme is relevant for clinical practice and warrants further development. For example, once a clinician marks the location of the cryoprobes and freezing front at a few points along its visible portion, a complete temperature reconstruction can be obtained in real time.

4.3 Experimental validation of the thermal history of the PFAM+TFRM scheme

Thus far, scheme validation has been demonstrated for the ability to predict the freezing front location based on experimental results, and for the ability to predict the lethal-temperature isotherm based on computer simulations. The discussion now turns to the ability to estimate the thermal history at preselected points in space, with Case B as a test case. Temperature measurements were obtained with T-type thermocouples ($\pm 0.5^\circ\text{C}$ uncertainty), as illustrated in Fig. 5. Comparison results are presented in Fig. 9 starting at 400 s, after which the temperature sensors became embedded in the frozen region. Differences between predicted and measured temperatures for TC1 range between 6°C at 400 s to 3.6°C at 850 s. Temperature differences for TC2 range between 2°C at 400 s to 0.8°C at 850 s. The main reason for the lower-quality of temperature predictions for TC1 is that it is located further than TC2 in relation to the observed freezing front. When the magnitude of 6°C is considered, it should be noted that (i) no real temperature measurement means are available during cryosurgery in the respective areas, and (ii) the maximum temperature variation in the domain can become two orders of magnitude larger than that value.

5. Summary and Conclusions

The primary challenge in ultrasound monitoring of cryosurgery is the high absorption of the ultrasound energy by the frozen region, which leads to (i) opacity of the frozen region on the ultrasound monitor, (ii) shadow effect behind the frozen region, and (iii) clear demarcation of the freezing front in areas close to the ultrasound transducer. The current study aims at developing computation means in order to gain more information about the temperature distribution in the cryotreated area and the location of the freezing front in invisible areas to US imaging. The current study is based on US image analysis as input, but does not rely on

the development of additional sensors or specialized instrumentation. This study uses prostate cryosurgery as a developmental model and TRUST imaging as a choice of practice.

A proof-of-concept for freezing-front estimation is demonstrated in the current study, using the new PFAM. This method is further integrated with a recently developed TFRM to estimate the temperature distribution within the frozen region. Results of the study indicate that the proposed PFAM is a viable and computationally inexpensive solution to estimate the extent of freezing in the acoustic shadow region. Comparison of PFAM estimations and experimental data shows an average mismatch of less than 2 mm in freezing-front location, which is comparable with the uncertainty in ultrasound imaging. Comparison of the integrated PFAM+TFRM scheme with full-scale finite elements analysis (FEA) indicates an average mismatch of 0.9 mm for the freezing front location, $T=0^{\circ}\text{C}$, and 0.1 mm for the lethal temperature isotherm, $T=-45^{\circ}\text{C}$. Comparison of the integrated PFAM+TFRM scheme with experimental temperature measurements show a difference in the range of 2°C and 6°C for the selected points of measurement.

It appears that the integrated PFAM+TFRM scheme is adequate to estimate the location of the freezing front, the location of the lethal temperature isotherm, and the temperature distribution in a clinical setup. Furthermore, given the associated computer runtime (0.5 s for PFAM and 3.5 s for the integrated scheme) and the lack of alternative cost-effective means, further development of the integrated PFAM+TFRM scheme appears warranted. While the above runtime results were obtained using MATLAB and general-purpose subroutines, it is possible to dramatically decrease the computation time by GPU-based computation in a C++ environment.

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Figure 1. Schematic illustration of a representative cross section during prostate cryosurgery displaying: (a) the ultrasound transducer, the freezing front, and the urethral warmer; and (b) the acoustic shadow by the frozen region and the partially visible freezing front on the ultrasound (US) monitor.

Figure 2. Integration of PFAM and TFRM for the purpose of (I) freezing front estimation in the invisible area of the US image, and (II) thermal history evaluation.

Figure 3.

A representative snapshot from a video recording of a simulated cryosurgery procedure [18], where the medium is distilled water mixed with 1.3% gelatin and blue food dye. The reconstructed frozen region is shown on the top-right corner of the image, which is the basis for the displayed freezing front contour in Fig. 1.

Figure 4.

Comparison between PFAM estimation for the 0°C isotherm (green curve) and freezing-front reconstruction from experimental results (blue curve) for Case A, where the PFAM results are only presented within the simulated acoustic shadow area (Fig. 4(a) is also presented in Figs. 1 and 3).

Figure 5. Comparison between PFAM estimation for the 0°C isotherm (green curve) and freezing-front reconstruction from experimental results (blue curve) for Case B, where the PFAM results are only presented within the simulated acoustic shadow area. TC1 and TC2 are thermocouples embedded in the freezing medium and used to validate the integration of PFAM and TFRM scheme in the context of Fig. 9.

Figure 6. Comparison between PFAM estimation for the 0°C isotherm (green curve) and freezing-front reconstruction from experimental results (blue curve) for Case C, where the PFAM results are only presented within the simulated acoustic shadow area.

Figure 7. Comparison of a benchmark generated with finite-elements analysis (ANSYS) and the estimated location of the freezing front (0°C isotherm) using TFRM, with and without integration with PFAM (Case D in Table 1).

Figure 8.

Comparison of a benchmark generated with finite-elements analysis (ANSYS) and the estimated location of the lethal-temperature isotherm, -45°C using TFRM, with and without integration with PFAM (Case D in Table 1).

Figure 9. Comparison of measured and estimated temperatures, using an integrated PFAM and TFRM scheme, where sensor locations are illustrated in Fig. 5.

Table 1

Summary of freezing front mismatch between the PFAM, TFRM, and experimental results

	Analysis Method	Elapsed time, s	Hausdorff distance, mm	
			S_{max} , Eq. (7)	S_{avg} , Eq. (8)
Case A (Fig. 4)	PFAM	420	4.5	1.1
	PFAM	760	2.8	0.8
	PFAM	1100	1.9	0.5
	PFAM	1440	1.9	0.5
Case B (Fig. 5)	PFAM	430	8.6	1.9
	PFAM	650	6.3	1.4
	PFAM	870	5.7	1.2
	PFAM	1080	5.3	1.1
Case C (Fig. 6)	PFAM	540	8.8	1.6
	PFAM	680	4.5	1.1
	PFAM	820	3.6	1.0
	PFAM	960	3.0	0.9
Case D, $T=0^{\circ}\text{C}$ (Fig. 7)	TFRM	N/A *	15.2	8.1
	TFRM+PFAM	N/A *	2.9	0.9
Case D, $T=-45^{\circ}\text{C}$ (Fig. 8)	TFRM	N/A *	10.1	5.3
	TFRM+PFAM	N/A *	3.1	0.1

* Optimal planning based on computation means described in [27]

Table 2

Representative thermophysical properties of soft biological tissue used in the current study [17]

Thermal property	Value	
Thermal conductivity, k , $\text{W m}^{-1} \text{K}^{-1}$	0.5	$273\text{K} < T$
	$15.98 - 0.56T$	$251\text{K} < T < 273\text{K}$
	$1005T^{-1.15}$	$T < 251\text{K}$
Volumetric specific heat, C , $\text{kJ m}^{-3} \text{K}^{-1}$	3,600	$273\text{K} < T$
	15,440	$251\text{K} < T < 273\text{K}$
	$3.98 T$	$T < 251\text{K}$
Blood perfusion rate, $C_b w_b$, W m^{-3}	40,000	$273\text{K} < T$
	0	$T < 273 \text{ K}$
Metabolic heat generation, q_{met} , W kg^{-1}	0.67	$273\text{K} < T$
	0	$T < 273\text{K}$