Localized FFT phase-correction algorithm for improved real-time PRF shift thermometry

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Introduction

Thermal therapy is a minimally invasive alternative for treatment of a variety of cancers that benefits immensely from MR-guidance for planning, targeting, monitoring treatment delivery and treatment verification. Proton resonance frequency (PRF) shift MR temperature imaging (MRTI) [1] can be used to qualitatively or quantitatively monitor progression of treatment in real-time, potentially enhancing the safety and efficacy of these emerging therapies. Unfortunately, PRF based MRTI relies on phase-subtraction techniques and is thus highly susceptible to time varying errors in the background magnetic field, such as those caused by susceptibility changes at interfaces or motion induced changes. Examples of these errors include respiratory motion in liver treatments, or tissue swelling in prostate treatments, which both obscure the ability to visualize the treatment area and can potentially add substantial error to quantitative estimations of the extent of damage [2]. Correction of motion artifact for the PRF consists of two interleaved problems. When possible, the treatment regions need to be registered for subtraction as well as for monitoring cumulative exposure during therapy. Since this registration does not compensate for the induced distortions in the background field, the background phase must also be corrected to avoid phase errors. In this work, we further our work from last year and investigate the latter problem of phase correction using an iterative fast Fourier extrapolation technique for estimation of the background phase [4]. We test the method during laser ablation in canine prostate, canine brain (control), and human liver and bone.

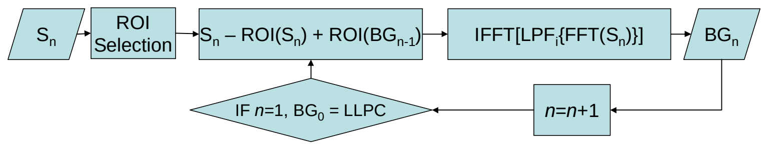

Fig 1: Iterative phase-correction algorithm for temperature imaging

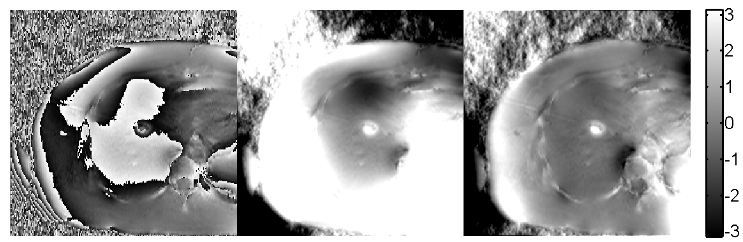

Figure 2: Raw phase images (a) from an MRgLITT liver patient (a) can be unwrapped [6] (b) and the background removed using the FFEA algorithm (c) to provide a "referenceless" approach to MRTI as described by [3]. However, in order to minimize artifacts from sharp interfaces, generate only modest background phase errors,and provide cumulative thermometry, subtraction is still usually desired.

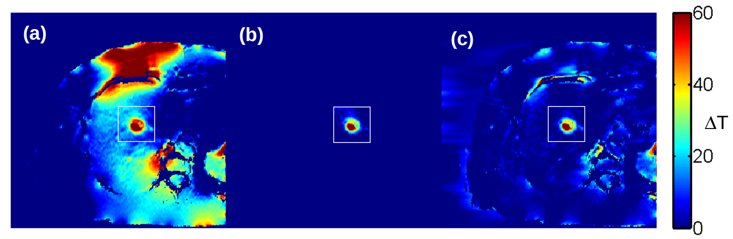

Figure 3: Retrospective correction of motion. The complex phase difference from an MRgLITT liver procedure in a patient on a 1.5T Siemens Espree (a) demonstrates substantial artifact from motion and field distortion. The LLPC algorithm approximates and removes the background phase within the prescribed ROI without removing the data (b) while the FFEC algorithm (4 iterations) removes the data from an ROI in order to remove effects of heating from the background estimation (c). Both algorithms can be performed with less than 0.5 seconds of processing.

Materials

All imaging described in this study was performed on 1.5T whole body MR scanners (Signa and Excite HD platforms, GE Healthcare Technologies) or (MAGNETOM Espree, Siemens Medical Solution, Inc.) using various receive only coils. All animals and human subject studies were conducted in accordance with the local independent review panels for ethical execution of these studies. MR-guided laser ablation procedures in humans, animals and phantoms were performed using 980-nm actively cooled diode laser system (Visualase®, BioTex, Inc, Houston, TX) with MRTI feedback to monitor and control therapy delivery. Single-plane fast spoiled-gradient echo imaging or multi-planar, fat suppressed, multi-shot interleaved EPI techniques were used for acquisition.

Like other "referenceless" background techniques [3], the algorithm assumes a slowly varying background and so may have problems near sharp susceptibility interfaces. The algorithm follows the flowchart in Fig. 1. First an ROI surrounding the area of heating is selected to be as small as possible while covering the expected extent of heating. It is assumed that the field of view is large compared to this area The image for which the background is to be determined can be complex amplitude, complex difference, phase, temperature or frequency maps as long as the background is slowly varying. Next the data within the ROI is removed and replaced with an approximation of the background data. For this we use the locoregional linear phase-correction (LLPC) estimate described previously. If zero is used in the ROI as opposed to LLPC, preliminary data indicate that the choice of low pass filter (LPF) in k-space as well as iterating the algorithm is critical to avoid underestimation. Next the data (with the heated region removed) is Fourier transformed, low pass filtered, then inverse Fourier transformed to produce an estimate of the background for the ROI. This new background estimate replaces the LLPC estimate. The process is iterated using increasing bandwidth for the LPF, but must remain smaller than the frequency content of the ROI window. Because only a key-hole of k-space is being used and we are not concerned with resolution, our choice for the LPF has been a Gaussian window to minimize ringing in the background (BG) phase estimates. The choice of the LPF minimum and maximum bandwidths are critical. We use a Gaussian where the minimum bandwidth is based on the width of the ROI dimensions.

Results

The algorithm can be applied in a manner congruent with referenceless PRF techniques (Fig. 2), but we run with subtractions for maximal suppression of background (Fig. 3). In general, 4 iterations or less were needed (more failed to contribute to suppression of background or started to diverge). In liver and prostate the phase-correction algorithm decreased background B0 errors in the ROI substantially, depending on the extent of motion, without significantly affecting heating in experiments with stationary tissue (i.e., brain). The algorithm suppressed small oscillations due to respiratory motion in the prostate. The algorithm generally increased the correlation between cumulative Arrhenius dose based on the time-temperature history and relevant post-therapy imaging. Like the LLPC algorithm, human liver showed a large range of results depending on location and degree of motion (Fig 3)

Discussion

We expand upon our previously described LLPC technique for PRF thermometry with one based on Fourier extrapolation into the heating zone. This technique is essentially a reference technique with a different method of extrapolation utilized. Like the LLPC presented last year, this technique can be applied in real-time and substantially increases the quality of both the resulting temperature map and thermal dose estimate. By using a large region of interest which does include the treatment zone, processing time is kept to a minimum and phase-correction results are loco-regional.

References

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