Latency compensation for real-time 3D HIFU beam-steering on moving targets

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Purpose/Introduction

MR-guided high intensity focused ultrasound (HIFU) is a promising candidate for the non-invasive ablation of pathological tissue in abdominal organs such as liver and kidney. However, due to the high perfusion rates of these organs, sustained sonications are required to achieve a sufficiently high temperature elevation to induce necrosis, which in turn requires sub-second 3D HIFU-beam steering under MR-guidance to achieve near real-time compensation of respiratory motion [1, 2]. A particular problem of this technique remains the intrinsic latency between the position measurement and beam update which leads to undesired energy dispersion and potentially to the destruction of non-pathological tissue. In this study, a robust Kalman-predictor [3] for 3D motion anticipation is suggested and it's usefulness for the compensation of the beam update latency is demonstrated experimentally.

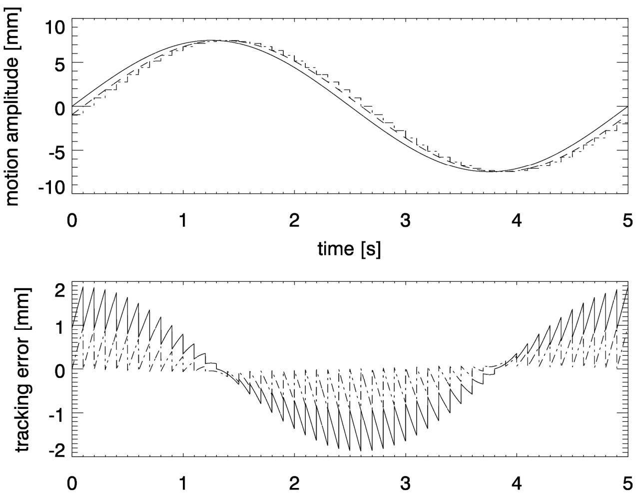

Figure 1. top: Simulated influence of a processing latency of 100ms (top, dashed line) and a sampling delay of 100ms (top, dash-dotted line) on the tracking error (bottom) of an ideal (i.e. sinusoidal, no noise) motion pattern. The tracking error is evaluated without motion anticipation (bottom, solid) and using a Kalman-predictor (bottom, dash-dotted).

Material and Methods

Tracking strategy: The target position is observed in 3D space by coupling rapid 2D MR-imaging with prospective slice tracking (PST) based on pencil-beam navigator echoes. The in-plane target position is obtained using real-time image registration as proposed by Roujol et al. [4] and combined with the current slice position obtained by the pencil beam navigator. The latency arising from the MR-acquisition and reconstruction, the image transport and the data processing is compensated using the linear predictor of an adaptive Kalman filter [3] which uses the current 3D target location and velocity as the state variable. Subsequently, the position is transformed from the MR-imaging coordinate system to transducer coordinates and send to the HIFU-transducer.

MRI: All imaging was performed on a 1.5T Philips Achieva scanner (Philips Healthcare, Best, The Netherlands) using a gradient recalled single shot echo-planar sequence (TE=46 ms, image matrix=128×96, flip angle=35°, readout BW=2.1 kHz, single-slice with a thickness of 5 mm. For MR-image reconstruction and processing a dual processor (INTEL 3.1 GHz Penryn, four cores) workstation with 8GB of RAM and dual 1GB/s network interface cards was used. All latency contributions were obtained by measuring the RF-excitation pulse of the MR-scanner with help of a dedicated pick-up coil, a TTL-signal generated by the reconstructor and the acoustic signal of the HIFU-transducer directly using an digital oscilloscope. Heating was performed with a 256 element phased-array transducer (radius=120mm, aperture=126mm) and allowing a lateral displacement of the focal point of 20mm peak-to-peak. A physiological phantom with relaxation times matched to the human kidney was mounted on a motorized platform to simulate an abdominal organ (displacement 15 mm peak-to-peak, motion period 5s to match the human respiratory cycle).

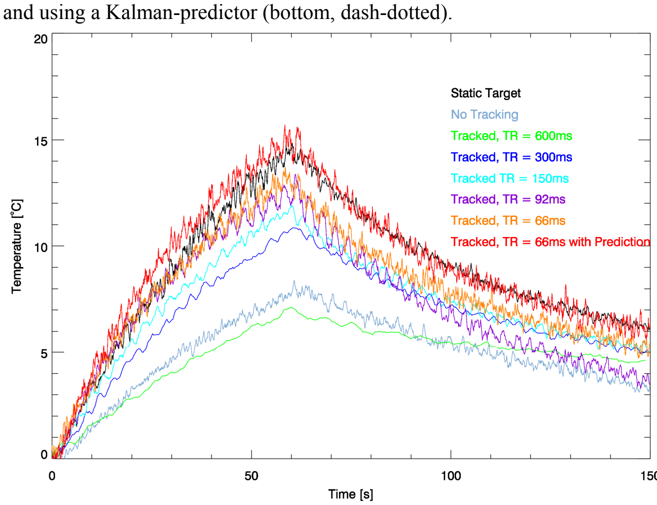

Figure 2. Comparison of the temperature evolution in the target area during motion compensated HIFU experiments using different sampling times and a fixed latency of 100ms.

Results and Discussion

The simulation in figure 1 explains the influence of the sampling frequency and the processing latency on the targeting precision. The resulting tracking error for the case of an imaging frame-rate of 10 images/s and a processing latency of 100ms is found to be ±1.9 mm (solid line in 1.b). The Kalman predictor (dashed line in 1.b) reduces the tracking error below the employed MR-image resolution (2.5mm) to ±0.9mm. These simulations are reconfirmed by the experimental findings shown in figure 2.: Only for imaging frequencies above 10Hz and beam update latencies below 100ms are more than 90% of the beam energy is deposited in the designated target area if no additional motion anticipation is used. A further acceleration of the data acquisition to 15Hz does not lead to an significant improvement since the remaining beam update latency remains as a limiting factor. The two major contributions to this intrinsic latency is the remaining post-TE k-space acquisition time (44ms) and the time required for MR-data transport, reconstruction and image processing (45ms) which both can not be significantly shortened due to the limitations of the available SNR and the available processing power. The suggested Kalman-predictor evaluates the velocity of a Kalman filtered target trajectory to anticipate the future target position to compensate for this processing latency. This allows to achieve an energy deposition which is comparable to the static reference experiment (fig. 2, red line). However, the predictor is not suitable for parts of the motion pattern which are subject to high accelerations, since it will produce overshots and deviations from the true trajectory. However, due to the high frame-rate and the low latency the predictor is only required to anticipate the target position over short epochs of 50-100ms before newly measured data is available, which limits the impact of such events.

References

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