Results: We were able to demonstrate the feasibility of our new acquisition technique for peak-inhale only, peak-exhale only, and mid-ventilation only short (200°) and full scanning protocols. In the figure, a significant reduction in motion-related artefacts at the tumour surface is demonstrated. Reconstruction artefacts stemming from the presence of the EM-array were greatly reduced. Inherently, the method leads to a smooth distribution of projection angles over the entire gantry rotation. A typical CTDI imaging dose for a peak-exhale (360°, 375 fr., 5 min)/peak-inhale (200°, 106 fr., 2:47 min) protocol is ~8/2 mGy.

Conclusions: We have implemented a new dose-saving 4D CBCT scanning protocol which reduces motion-artefacts by selecting the breathing phase(s) desired for reconstruction prior to image acquisition. We are currently implementing the acquisition of multiple phases in one gantry rotation.

OC-0335
Tumour motion tracking technique based on dynamic surface scanning and 4D CT breathing motion model
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Purpose/Objective: To develop and evaluate a tumour tracking method for the dynamic functionalization of extracranial targets during radiotherapy treatments, providing the continuous monitoring and compensation of intra-fraction organ motion due to breathing. The proposed approach is based on external surface surrogates estimated from non-invasive optical devices, and on patient-specific adaptive motion models derived from time-resolved planning and in-room X-ray imaging systems.

Materials and Methods: A patient-specific motion model, parameterized as a function of the respiratory tumour baseline, amplitude and phase, is estimated from 4D CT planning images, by applying B-spline deformable registration between each 4D CT phase and the mid-position CT volume. The tumour baseline is adapted at each treatment fraction according to daily information on target localization derived from volumetric CBCT images. The breathing amplitude and phase parameters are retrieved from the external surface motion, acquired with 3D surface imaging systems. Deformable mesh registration is applied to derive the spatial correspondence between markerless optical surfaces. The obtained 3D trajectories of all thoraco-abdominal surface points are summarized into a single respiratory surrogate signal through k-means clustering techniques. The instantaneous values of the respiratory phase are extracted from the surface surrogate using the Hilbert transform. The amplitude scaling factor is obtained by comparing surface motion amplitudes during treatment planning and delivery. The adapted breathing parameters are integrated in the 4D CT motion model to estimate intra-fraction 3D tumour motion.

Results: The developed tumour tracking method was tested on a clinical database of seven lung cancer patients, including the synchronized information on the external surface and internal tumour breathing motion during CBCT scans. About 30 seconds of synchronized acquisition of CBCT projections and optical surfaces, captured with the VisionRT system, were analyzed for each patient. The tumour trajectories estimated from surface displacement combined with the a priori 4D CT motion model were compared to the real target trajectories identified on CBCT images. The resulting absolute differences between real and estimated tumour motion ranged between 0.7 and 2.4 mm, with median values of 1.5 mm both along the horizontal and vertical image dimensions (Table 1). The measured phase shifts did not exceed the 7% of the breathing cycle length.

Table 1. Tracking errors between tumor trajectories estimated from external surface displacement and real tumor motion identified on CBCT projections along the horizontal and vertical image directions. Results are expressed as median value [25% ~ 75% percentile].

<table>
<thead>
<tr>
<th>Patient</th>
<th>Horizontal tracking error (mm)</th>
<th>Vertical tracking error (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>P1</td>
<td>2.2 [1.2 ~ 3.8]</td>
<td>1.6 [0.9 ~ 2.7]</td>
</tr>
<tr>
<td>P2</td>
<td>1.7 [0.8 ~ 3.0]</td>
<td>1.2 [0.6 ~ 1.9]</td>
</tr>
<tr>
<td>P3</td>
<td>1.2 [0.6 ~ 1.7]</td>
<td>2.0 [1.6 ~ 2.4]</td>
</tr>
<tr>
<td>P4</td>
<td>0.7 [0.4 ~ 1.0]</td>
<td>1.4 [0.9 ~ 1.9]</td>
</tr>
<tr>
<td>P5</td>
<td>1.7 [0.8 ~ 3.1]</td>
<td>2.4 [1.4 ~ 3.5]</td>
</tr>
<tr>
<td>P6</td>
<td>1.3 [0.8 ~ 1.9]</td>
<td>1.5 [0.9 ~ 2.0]</td>
</tr>
<tr>
<td>P7</td>
<td>1.5 [0.9 ~ 2.3]</td>
<td>1.4 [0.6 ~ 2.2]</td>
</tr>
</tbody>
</table>

Conclusions: The developed tumor tracking method proved to be effective in estimating tumour motion from the external surface displacement even in presence of breathing irregularities, as depicted in Figure 1. The innovative methodological aspects, related to the use of patient-specific adaptive motion models and to the redundancy of markerless surface data, are put forward to improve the accuracy and robustness of targeting techniques for intra-fraction organ motion compensation.

Figure 1. Comparison between real and estimated tumor trajectories obtained for patient P2. The breathing irregularities both for amplitude and phase parameters. Vertical image-dimension corresponds to the projection of the superior-inferior component of tumor motion, while horizontal image-dimension expresses the combinations of tumor motion projection along antero-posterior and medio-lateral directions.

OC-0336
Inter- and intrafractional relative motion of implanted fiducial markers in the liver
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Purpose/Objective: In radiotherapy, fiducial markers are often implanted in the liver and used as x-ray visible surrogates for the tumour position. The basic assumption is that the tumour maintains a constant position relative to the marker throughout the treatment course. The aim of this study was to investigate the validity of this assumption by quantifying the geometric accuracy by which one implanted marker can predict the position of another marker in the liver.

Materials and Methods: 26 patients with 2-3 implanted gold markers received stereotactic body radiation therapy to liver lesions in 3-6 treatment fractions. At each fraction a cone-beam CT (CBCT) scan was acquired and used for marker based patient setup. After the treatments each marker was segmented in the projections of the CBCT scans, and the resulting 2D marker trajectories were used to estimate the 3D trajectory of each marker in patientspace by a probability based method. In total 198 3D marker trajectories with a mean duration of 51 seconds were recorded at 11 Hz over 90 treatment fractions. The markers in the data set constituted 54 marker pairs in total. For each marker pair, the error in estimating the 3D position of one marker from the 3D position of the other marker was calculated for each time point as the difference in the 3D shifts the two markers away from their respective mean positions. The root-mean-square (RMS) of this error was calculated for each marker pair as a measure for the intrafraction prediction error, during all fractions. Similarly, the interfraction RMS estimation error was calculated for each marker pair from the changes in the relative 3D mean marker positions from fraction to fraction.

Results: The 3D position of an implanted marker was estimated from the position of another marker with a mean RMS error of 0.35 (LR), 0.59 mm (CC), 0.47 mm (AP), and 0.86 +0.57 mm (3D) intra-fractionally, and of 0.52 mm (LR), 0.60 mm (CC), 0.59 mm (AP), and...
Pancreatic motion is estimated on MR images, as compared to internal tracking based on the motion of the diaphragm. External monitoring by means of a respiratory pressure belt, is therefore investigate two (fast) surrogate motion markers for temporal resolution required for real-time beam steering. We imaging, however, is currently not fast enough to achieve the track OARs during treatment (Lagendijk, 2008). Volumetric (3D) MR-at risk (OAR). MR-guidance opens up the way to visualize tumors and for lung cancer using log files.

Purpose/Objective: Radiotherapy for pancreatic tumors is currently limited by respiratory induced motion. Precise motion management is vital to ensure full dose coverage and limit toxicity in nearby organs at risk (OAR). MR-guidance opens up the way to visualize tumors and track OARS during treatment (Lagendijk, 2008). Volumetric (3D) MR-imaging, however, is currently not fast enough to achieve the temporal resolution required for real-time beam steering. We therefore investigate two (fast) surrogate motion markers for pancreatic motion to be used in conjunction with (slower) 3D MRI. External monitoring by means of a respiratory pressure belt, is compared to internal tracking based on the model conformity.

Materials and Methods: 2D radial cine MR scans (bSSFP; TE/TR = 1.29ms/2.6 ms) were acquired in six healthy subjects during free-breathing (2m50s, 3 frames/sec). Scan planes were angulated parallel to the principal axis of motion. Simultaneously, the breathing signal was recorded using a respiratory pressure belt (surrogate 1). 1D MR navigators were monitored for 40 s with the IR camera every 16.7 ms and an orthogonal kV X-ray imaging subsystem every 80 or 160 ms, respectively. The 4D model predicted the future target position from the displacement of the IR markers in real-time, and the gimbaled x-ray head then tracked the moving tumour continuously, based on the predicted target position. In clinical practice, we updated the 4D model at least once during each treatment session to ensure predictive accuracy. This study evaluated the 4D modelling error (E4DM) and influence of the intrafractional baseline drift of the IR marker position on the predicted target position (E4DM). E4DM was defined as the difference between the predicted and detected target positions during the modelling duration and EIR was defined as the difference between the predicted target positions generated from parameters of previous and updated 4D model. For E4DM and EIR, the overall mean (M), systematic (σ), and random (ε) errors were calculated from the log files of ten patients who underwent IR tracking. A total of 112 and 55 log files were analyzed for the E4DM and EIR, respectively.

Results: The respiratory motion amplitudes of the lung tumours ranged from 1.0–7.5, 4.7–28.5, and 2.4–10.5 mm in the left-right (LR), anterior-posterior (AP), and cranio-caudal (CC) directions, respectively. For the E4DM, (M, σ) (in mm) were (0.0, 0.0, 0.4), (0.0, 0.0, 0.8), and (0.0, 0.0, 0.6) in the LR, CC, and AP directions, respectively. The local maximum E4DM commonly appeared around the peak of the respiratory pattern (Fig. 1). The median time elapsed until the 4D model was updated was 13 (range 2–23) min. For the EIR, (M, σ) (in mm) were (0.2, 0.3, 0.2), (0.2, 0.9, 0.5), and (0.4, 0.6, 0.3) in the LR, CC, and AP directions, respectively. If the 4D model was not updated in the presence of intrafractional baseline drift, the predicted target position deviated from the detected target position systematically (Fig. 1).