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Markerless PET motion correction: tracking in narrow gantries through optical fibers

Rasmus Ramsbøl Jensen, Oline Vinter Olesen, Claus Benjaminsen, Liselotte Højgaard, and Rasmus Larsen

ABSTRACT

In a time with increasing resolution and signal-to-noise ratio of medical 3D brain scanners, there is also an increased need for tracking and motion correction of patient movements during acquisition time. To successfully implement a system for motion tracking in the clinic, the system should be accurate while only adding minimal complexity to the workflow. We present: Tracoline 2.0, a surface scanner prototype, which allows for markerless tracking in the clinic. The system uses structured light through optical fibre bundles, which easily fit in narrow gantries. The optical fibres also makes the system compatible with magnetic resonance (MR) imaging since all the electronics are moved away from the scanner. We demonstrate the system in a positron emission tomography (PET) study using the Siemens high resolution research tomography (HRRT). With two Ge/Ga-68 line sources fitted in a mannequin head mounted on a rotating stage we evaluate the system for stepwise motion with periods of rest and for continuous motion. Based on comparison with the ground truth of the rotating stage, we were able to accurately track the movement with a rotational error of -0.073° to 0.098° with a maximal SD of 0.031° for rotations up to $\pm 25^\circ$. Based on the tracking results the PET frames were also successfully corrected for motion by aligning 10 s frames without motion for the stepwise experiment and aligning 1 s frames for the experiment with continuous motion. We have demonstrated and evaluated a system for markerless tracking and motion correction. The system is a significant step towards markerless tracking and motion correction seamlessly implemented in the clinic.

I. INTRODUCTION

Medical 3D brain imaging has proven to be a valuable tool in diagnostics, treatment planing, and neuroscience. With improving scanner technology leading to increases in resolution and signal-to-noise ratio, the need for reliable motion correction also increases. In the clinic; fixation of patients during acquisition time is unpleasant and even impossible for a wide range of patients such as patients with claustrophobia,

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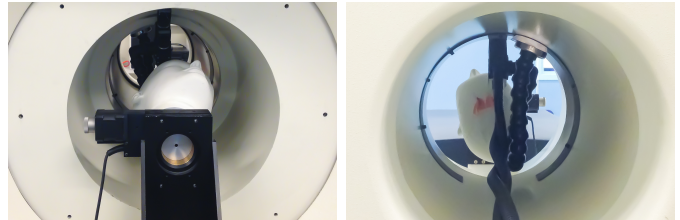


Figure 1. Tracoline 2.0 mounted in the narrow patient tunnel of the HRRT over and behind the phantom on a rotary stage. The fiber bundles can be seen running through the back entrance of the bore.

dementia, involuntary movements, and children. While several motion correction algorithms exist e.g. in brain positron emission tomography (PET) [1], [2], the problem of tracking motion robustly for clinical use has not been solved. Image based tracking may fail due to 1) lag of contrast and 2) the patient motion and the time courses of the functional pattern may confound. Marker based tracking systems such as the Polaris system for motion tracking in brain imaging [3] can accurately track markers. However, markers introduce two new essential problems; 1) The system only tracks the markers, which can move unnoticed relative to the skull during scan acquisition and cause distorted image reconstruction. 2) Adding a marker based system adds complexity to the patient preparation, reducing the costly scanner work flow. For these reasons it is hard to implement in the clinic.

We have previously demonstrated a markerless tracking system in PET brain motion correction based on structured light (SL) [4]. In this paper we introduce our new tracking prototype designed to compromise narrow bore holes and also being compatible with magnetic resonance imaging. Optical image fiber bundles remote all the electronics away from the tracking subject in bore. Allowing for easy access into narrow gantries such as the Siemens high resolution research tomography (HRRT) PET scanner. We evaluate our new system referred to as Tracoline 2.0 on phantom studies both on motion corrected PET frames and also against the ground truth of a rotation stage. The principals of the surface reconstruction and tracking pipeline are similar to our first prototype [5] but with improved system calibration, surface alignment, and speed now accomplishing realtime streaming of point clouds at up to 20 Hz.

II. EXPERIMENT AND METHODS

The developed Tracoline 2.0 tracking system consists of a light projector and a camera mounted in a radio frequency shielded box for MR compatibility. The projector and camera

are connected with optical fiber connections and two flexible image fiber bundles of 3 m. The fiber bundles both terminate with a set of low aperture optics. The line of sight for both optics is redirected with a mirror. The light source in the projector has been replaced with a near infrared (NIR) light source in order to remove discomfort from direct light onto the patients face. With both projector and camera synchronized the system is able to produce high quality 3D point clouds through phase shift interferometry (PSI) [6]. The system uses 6 patterns at two spatial wavelengths to produce point clouds at up to 20 Hz. The system is able to track from one point cloud to the next using the iterative closest point algorithm [7]. The algorithm continuously tracks the rigid point cloud transformation in relation to the point cloud at time zero, thus providing the tracking for the motion correction.

The Tracoline 2.0 system was applied for motion correction on the HRRT, Siemens dedicated brain PET scanner at the University Hospital of Copenhagen. Two typical motion scenarios were tested: long drift and stepwise rotary motion using a mannequin head mounted on a Thorlabs rotating stage (NanoRotator). The mannequin head was fitted with two Ge/Ga-68 line sources with 3.4 MBq. The rotations of the stage motor constitute a ground truth for the motion estimates of our tracking, while the uncorrected and corrected PET frames will show a quantitative measure of performance. The experiments with the rotating stage were carried out during 5-8 min PET acquisitions with simultaneous tracking at 8-11 Hz. In scenario 1, the stage rotated the phantom in increments of 5° with stops of 20s. The increments continued to a maximum of 25° left and right. This scenario simulates a situation, where the patient is piecewise still. In scenario 2, the phantom would move continuously to a maximum rotation of $\pm 25^\circ$ with an angular velocity of $5^\circ/s$. This is a worst case scenario, where there are no periods of rest, and all data has to be corrected in small time increments. Ideally, repositioning all the individual detected PET events. For this study the PET data was framed into frames of 1 s. As a reference a scan with no motion was also acquired and framed into 1 s frames.

The PET frames were reconstructed using the standard 3D-OSEM-PSF reconstruction algorithm (16 subsets, 10 iterations) without attenuation and scatter corrections due to the low attenuation of the air filled phantom. The PET image frames were motion compensated realigning the frames to a reference position according to the tracked motion [1].

In order to perform motion compensation the image volume and the tracking data has to be in the same coordinate system. This was done by acquiring a transmission (TX) scan of the phantom while also surface scanning it. The two data sets were used for coordinate system alignment [4]. If the Tracoline 2.0 system is fixed to the scanner this only has to be done once.

III. RESULTS

In the scenario with the rotating stage with periods of stationary rest, we were able to compare our tracking estimates with the ground truth. Figure 2 shows the estimated rotation (estimated as in [5]) as a function of time. The rotation estimates show a high agreement with the known rotations of

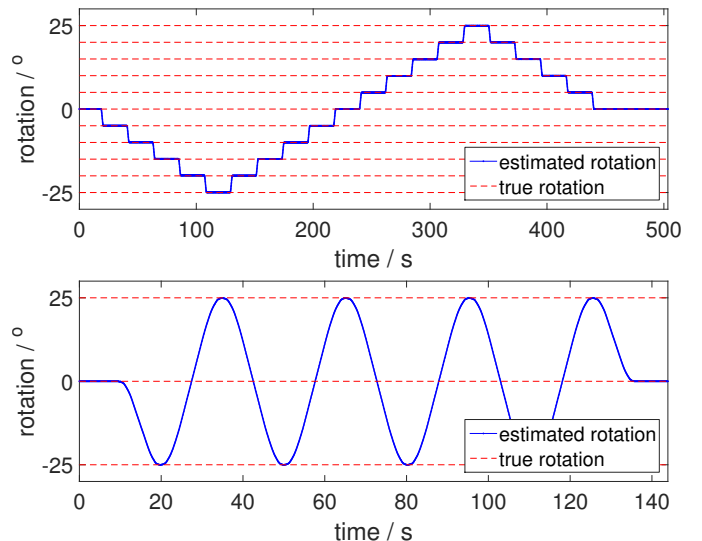


Figure 2. Estimated rotations of the stepwise motion (top) (error is listed in Table I) and the continuous motion (bottom). The tracking has been done at 8.33 Hz and 11.11 Hz, respectively.

the stage. Omitting the tracking while the stage was moving and comparing to the ground truth of the rotating stage, we find the standard deviation (SD), mean, and mean error (Table I). The maximum mean error is 0.098° with an SD of 0.031° . The total analysis was done with about 150 tracking points for each position of the stage. At our reference position, 0° , the error and SD are $< 10^{-5}$.

Figure 3 shows the summed uncorrected (top) and corrected (bottom) PET frames. From the corrected PET frames we estimate the average full-width-half-maximum (FWHM) of the cross section along the length of the line sources. The actual active diameters of the rods are 2.28 mm. For the stepwise motion the FWHM is 2.29 mm (Fig. 3(e)) and for the continuous motion 3.40 mm (Fig. 3(f)). As a reference the FWHM was found to be 2.17 mm for a 5 minute PET frame with no motion (Fig. 3(a)). To test our system, we also made a reconstruction of the no motion PET frame, where we used our tracking data (which should reveal no motion) to reposition 1 s frames (Fig. 3(d)). This resulted in an identical FWHM of 2.17 mm, revealing no added noise. The residual of the FWHM for the stepwise motion is only 0.12 mm compared to no motion. This should be explained by the interpolation of the realignment, partial volume effect, and some but less due to the accuracy of the tracking (c.f. Table I and Fig. 3(d)). The corrected image for the continuous motion is less focused. This is due to the fact that there was some interframe motion, the correction was with 1 s framing and the motion was up to $5^\circ/s$. However, the major part of the motion has been corrected.

IV. CONCLUSION

We have presented our new markerless tracking prototype and also applied it for PET motion correction on phantom studies. We have demonstrated that we can accurately track motion and apply this in PET motion correction. The Tracoline 2.0 prototype is a significant step towards markerless motion

Table I
TRACKING RESULTS OF THE STEPWISE ROTATION. MEAN, SD, AND ERROR ARE LISTED FOR EACH STAGE POSITION INCLUDING THE REPETITIONS (150-300 SAMPLES). THE ERROR IS THE MEAN OF THE DIFFERENCES BETWEEN THE GROUND TRUTH AND THE ESTIMATED ROTATIONS.

Rotation ($^{\circ}$)	-25	-20	-15	-10	-5	0	5	10	15	20	25
Mean ($^{\circ}$)	-24.944	-19.902	-14.927	-9.983	-5.000	0.000	4.966	9.947	14.927	19.966	24.945
SD ($^{\circ}$)	0.011	0.031	0.027	0.021	0.017	0.000	0.018	0.019	0.021	0.017	0.011
Error ($^{\circ}$)	0.056	0.098	0.073	0.017	0.000	0.000	-0.034	-0.053	-0.073	-0.034	-0.055

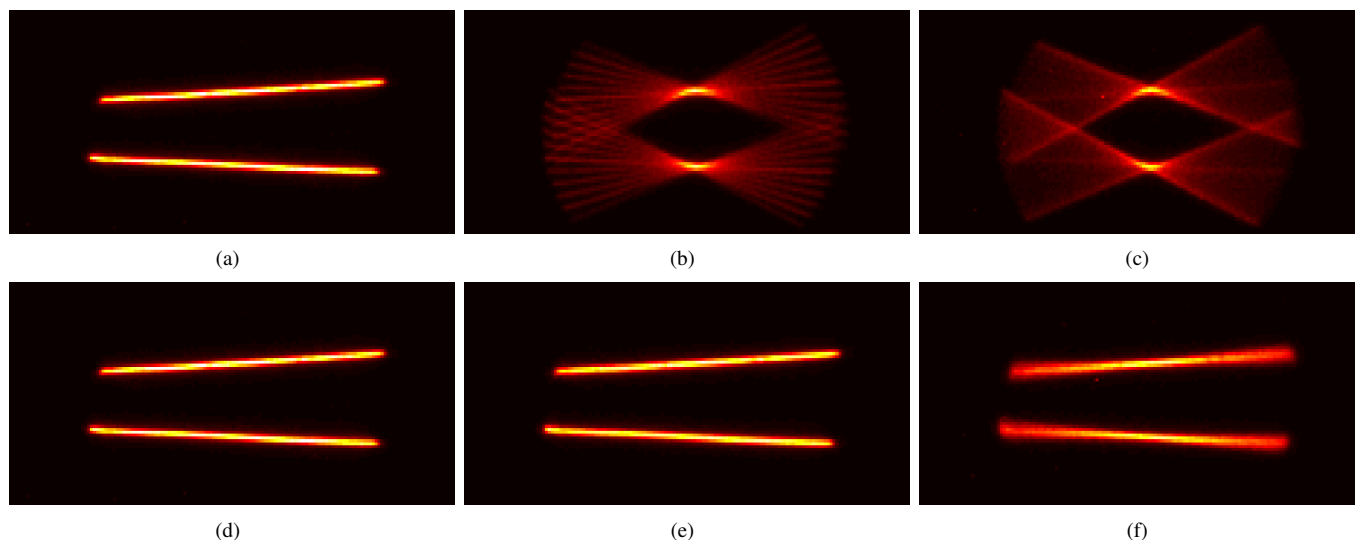


Figure 3. Line sources in uncorrected and corrected PET frames summed along the frontal image plan with a FOV of 86×172 mm. No motion: (a) and (d), Stepwise motion: (b) and (e). Continuous motion: (c) and (f). The line sources has a diameter of 2.28 mm. The average full-width-half-maximum (FWHM) for the reference frame with no motion (a) and with our motion correction applied (d) have both been estimated to 2.17 mm revealing no added noise, the stepwise corrected (e) to 2.29 mm, and finally the corrected continuous motion (f) to 3.40 mm.

tracking and correction, which can be seamlessly added to the clinical workflow.

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