

Temporal subtraction of chest radiographs compensating pose differences

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Abstract. Temporal subtraction techniques using 2D image registration improve the detectability of interval changes from chest radiographs. Although such methods are well known for some time they are not widely used in radiologic practice. The reason are strong pose differences between these follow-up acquisitions with a time interval of months to years in between. Such strong perspective differences occur in a reasonable number of cases. They cannot be compensated by available image registration methods and thus mask interval changes to be undetectable. A method is proposed to estimate a 3D pose difference by the adaptation of a 3D rib cage model to both projections. The difference between both is then compensated for, thus producing a subtraction image with virtually no change in pose. No 3D image data is used. The accuracy of pose estimation is validated with chest phantom images under controlled geometric conditions.

Purpose. Chest radiographs are the most frequently used medical images worldwide. During the diagnosis of a chest radiograph it is recommended to compare the actual image with a given previous one whenever available. Temporal subtraction, that means registration of both images and subsequent image subtraction, facilitates this diagnosis with the result of a better receiver operating characteristic (ROC) for the detection of interval changes as it was shown for solid lung nodules [4] or hazy opacities [10]. In practice, the difference in patient pose between acquisitions may be considerable. In fact, this difference is the main prohibiting factor for a wide use of temporal subtraction techniques [5]. Especially anterior-posterior (A-P) inclination and rotation around the patient's longitudinal axis lead to strong artifacts (“anatomical noise”) in the subtraction images that counteract interval change detection. Fig.1 illustrates the effect of pose differences on interval change detection with a thorax phantom. Image registration in the image plane (2D) as proposed in [3,7] can compensate to a certain extend for this anatomical noise, but restricted to a few degrees of rotation only. Unfortunately, there is a considerable number of image pairs in clinical practice showing strong angular differences, where 2D registration does not compensate well.

Model based pose estimation from few projections has been reported in literature. Typically, the model pose parameters are optimized to minimize a distance measure between its simulated projection and the object in the image. In [6], a signed distance measure is minimized, while an iterative closest point approach is followed in [1]. In [2], a rib cage model is adapted to binary digital reconstructed radiographs using a two-sided distance measure. In the novel approach presented here, the difference in 3D pose between both acquisitions is estimated by the adaptation of a 3D model and it is perspectively corrected for in the subtraction image. In the remainder of this article we describe the method for estimating the pose of a patient's chest by adaptation of a 3D rib cage model to radiographs. We present the method how to use the estimated pose difference between follow-up images for compensating for it in the subtraction image. We show the results for a novel experimental design where a thorax phantom was imaged under well controlled geometrical conditions and interval change was simulated by plastic foam attached to it. This subtle interval change was rendered visible even under severe pose differences.

Methods. The procedure to obtain subtraction images compensated for pose differences is made up of two stages. First, the 3D transformation that approximates the difference in patient pose is determined. Second, it is applied to simulate an image corrected for those pose differences. To estimate the patient's pose from a single PA radiograph, a 3D surface model of the patient's rib cage is matched to the radiograph such that the pose of the 3D model yields a good approximation of the patient's pose during image acquisition. We used the approach proposed in [2] for the application to a digital reconstructed radiograph (DRR) and applied it to real radiographs. To this end, simulated projections of the model were generated and compared to the segmented radiograph on the basis of distances between silhouette contours of the ribs. The silhouette distance between a contour point $x_m \in S_m$ in the model's projection to a set of contours S_r in the radiograph is given by

$$d(\mathbf{x}_m, S_r) = \min_{\mathbf{x}_r \in S_r} (\|\mathbf{x}_m - \mathbf{x}_r\|).$$

The distance is defined between contours of corresponding ribs only. Thus, the ribs need to be labeled in the radiograph. As previous experiments have shown that the posterior ribs are misaligned more often than the anterior or lateral parts of the ribs, we weighted their contribution to the distance measure in the model stronger than that of the other parts. Dissimilarities of the model's projection to the radiograph are iteratively reduced by adapting parameters that control the six degrees of freedom of the model's 3D pose, i.e., a rigid transformation P consisting of a rotation matrix b and a translation vector t . The adaptation is driven by the minimization of a two-sided distance measure

$$D_S = \int_{\mathbf{x}_m \in S_m} d_e(\mathbf{x}_m, S_r)^2 d\mathbf{x}_m + \int_{\mathbf{x}_r \in S_r} d_e(\mathbf{x}_r, S_m)^2 d\mathbf{x}_r.$$

For the optimization of D_S a gradient-based optimization procedure was applied. Based on two pose estimations from a previous and an actual radiograph, the patient pose difference between two acquisitions is given by $T_p = P_a^{-1} P_p$ (P_a actual pose, P_p previous pose). Once the patient pose difference is approximated by T_p we are able to simulate the patient position in the acquisition geometry (see Fig.2). As we do not have volumetric properties of the patient, but only projection images, we apply the following constraint initially proposed in [9] to reconstruct a radiograph: We assume that all interactions of the X-rays with the patient occur in a single coronal plane only. This plane shall have a distance d_f to the X-ray source. Under this "flat patient" assumption we may transfer a point on the image to the d_f plane by T_1 , apply the approximated patient transformation T_p and transform it back with T_3 . This can be expressed in a single transformation T when using homogeneous coordinates (see Fig.2 and equation there). Its projection back to the image by T_3 gives i' . θ_x (A-P inclination) and θ_z are the most relevant components of T_p . As d_f is undetermined it may be adjusted interactively while viewing the compensated subtraction image.

Results. We used radiographs of a thorax phantom made of human bones acquired under well-determined pose conditions to verify the pose estimation as well as the quality of the subtraction images. The phantom was imaged once with a foam plastic piece attached to it that was hardly visible in the projection image without subtraction (baseline image, Fig.1a). The phantom was imaged again without that piece at four different poses (baseline again, rotation around the longitudinal axis $\theta_z = +5\text{deg}$, $\theta_z = -5\text{deg}$, and A-P inclination $\theta_x = +3\text{deg}$). Fiducial markers attached to the phantom and a CT image of it allowed for a very precise reference designation of the pose geometry by point-based registration. The rib silhouettes were manually segmented from these radiographs and the 3D rib cage model was adapted automatically to fit that silhouette. For each of the four images, 20 different reconstructions were performed with randomly initialized poses of the phantom's 3D model. We determined the accuracy of pose estimations from these PA radiographs. Adaptation errors are given in 3D space as deviation $T_{p\text{-error}}$ from the correct pose - displacement (t_x, t_y, t_z in mm) and angular error ($\theta_x, \theta_y, \theta_z$ in deg). Furthermore, the 3D surface distance (vertex to corresponding vertex) between the surface of the adapted model and the reference surface reconstructed by point-based registration, which represents the correct pose, is given (cf. Table 1). These errors appear acceptable apart from relative strong displacements in t_y along the optical axis especially for Image 2 and 4 that cause mean displacement errors above 5mm. However, given a source to detector distance $d_s=2000\text{mm}$ such a displacement causes only displacements in the projection that are in the range of the direct t_x and t_z errors. The t_y error thus does not extraordinarily affect the quality of a subtraction image. Fig.3 illustrates the result of the generation of subtraction images compensating an angular displacement determined between a previous and a follow-up radiograph. For an especially difficult case, in which the follow-up image exhibits a difference in pose of $\theta_z = 13\text{deg}$ rotation, the pose movement T_p was determined and then used to generate a corrected subtraction image. T_p was obtained with an accuracy of $T_{p\text{-error}} = (0.53, 2.74, 0.04, 0.14, 0.29, 0.78)$ and an average surface distance of 1.4mm. The choice of d_f strongly determines the quality of the subtraction image. Anatomical structures close to the coronal plane at d_f are compensated perfectly, because for them the underlying assumption that interaction of the X-ray with the patient occurs in the coronal d_f -plane actually holds true. Other structures cause anatomical noise depending on their contrast and on their distance to the d_f -plane. The best

results are achieved if those anatomical structures that mask the interval change are compensated. For the phantom image these are (1st) a posterior rib and (2nd) the scapula as shown in Fig.3.

Table 1. Results for the pose reconstruction of the phantom’s thorax from manually segmented PA radiographs.

Phantom X-ray	$T_{p-error}=(t_x, t_y, t_z, \theta_x, \theta_y, \theta_z)$	Surf. dist. in mm mean \pm std (max)
Baseline Image	0.22, 6.41, 0.34, 0.17, 0.10, 0.13	2.94 \pm 0.74 (3.97)
Image 2, $\theta_z=+5$ deg	0.09, 13.65, 0.20, 0.21, 0.10, 0.22	6.40 \pm 1.39 (10.80)
Image 3, $\theta_z=-5$ deg	0.19, 2.45, 1.04, 0.11, 0.23, 0.10	1.35 \pm 0.80 (3.06)
Image 4, $\theta_x=+3$ deg	0.07, 11.03, 0.58, 0.22, 0.03, 0.33	5.28 \pm 1.69 (10.99)

New or Breakthrough Work. We see a strong virtue in a fully automatic method of temporal subtraction that compensates for angular pose differences, because they appear to be the most prohibiting factor for interval change detection by available temporal subtraction techniques. To our knowledge a combination of both methods, model-based pose estimation and compensation, has never been pursued before. We validated our method quantitatively in a phantom study. Inaccuracies in the correct estimation of the patient-to-detector-distance of about 6mm had no strong impact on the final subtraction image quality due to its little influence on the projection image. In its current implementation, the method relies on the manual segmentation of the ribs in the PA radiograph. Automatic segmentation seems feasible, as it is suggested by the authors in [8], but has not been implemented yet. A fully automatic version of this method would further require the adaptation of a statistical shape model to an image by variation of both pose and shape parameters.

Conclusion. A method is presented that estimates the 3D pose of a patient from a single PA radiograph in two follow-up images by the use of a 3D thorax model. Following the idea expressed in [9], the estimated patient pose difference between both acquisitions is used to compensate for this pose difference in a temporal subtraction image. Experiments with a thorax phantom acquired under well controlled geometric conditions demonstrated the accuracy of the pose difference estimation. With the simulation of a subtle interval change by plastic foam attached to the phantom this method was shown favorable over deformable 2D registration in cases of strong angular pose differences. Unlike 2D methods this pseudo-3D method was able to compensate for a rotation of 13deg. The results promise to make interval changes detectable and assessable easier than without that method even in those cases where patient pose differs strongly between follow-up acquisitions.

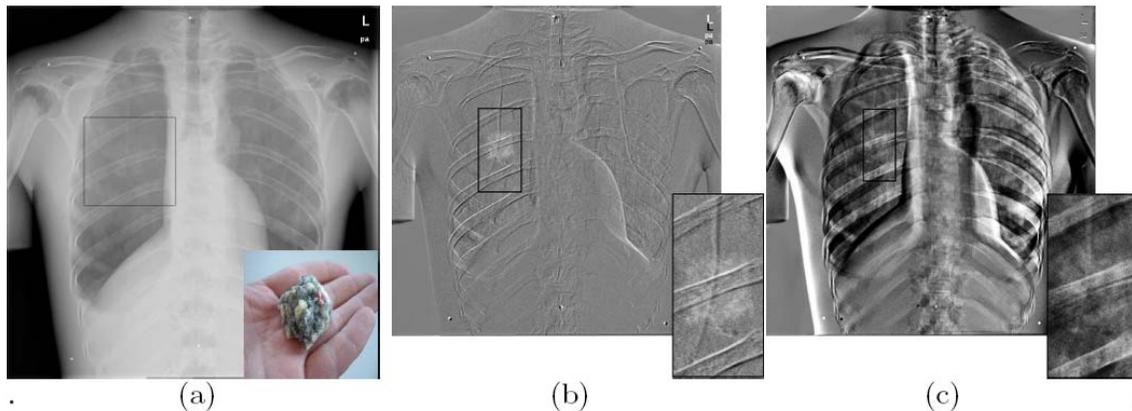


Figure 1: (a) Radiograph of the thorax phantom with subtle, almost invisible opacity (see box) caused by attached plastic foam. (b) Subtraction image with 1 deg rotation. Despite some anatomical noise from posterior ribs and the scapula border the interval change is visible. (c) Subtraction image with 13 deg rotation. Anatomical noise strongly masks the interval change.

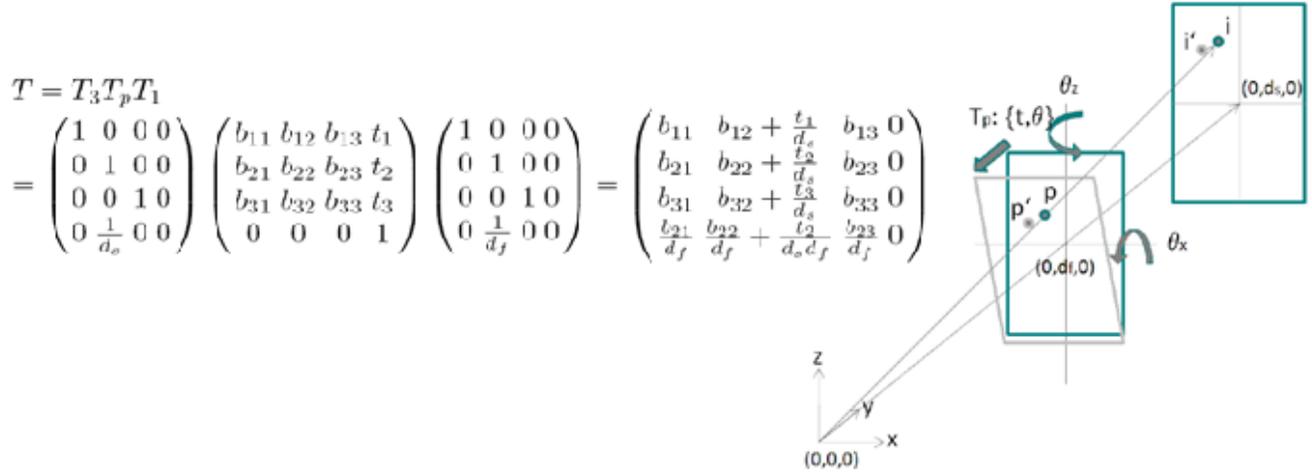


Figure 2: Compensating a pose difference: Image position i is projected by T_1 to p in the coronal patient plane at $y=d_f$. Estimated patient pose difference T_p is applied to p giving p' . Its projection back to the image by T_3 gives i' . θ_x (A-P inclination) and θ_z (rotation around the patient's long axis) are the most relevant components of T_p .

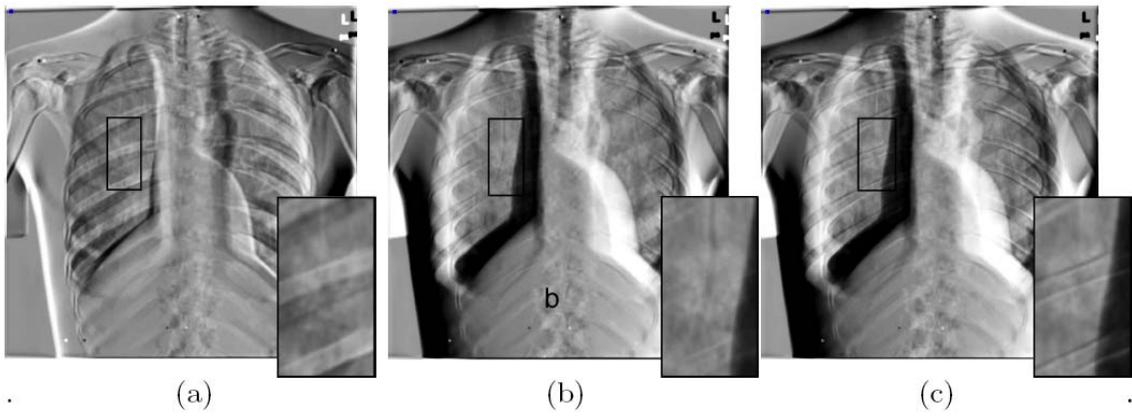


Figure 3: Subtraction images compensating a strong ($\theta_z = 13\text{deg}$) phantom rotation. d_f is chosen to compensate ideally for (a) the clavicles, (b) posterior ribs, and (c) the scapula. The interval change as shown in Fig.1 is clearly visible here in (b) and (c), but masked by a posterior rib shadow in (a). Note: The depth of the simulated tumor cannot be determined as it is present in one image only.

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