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 is the occurrence of strong pose differences between two 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. In this paper 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. The method generally assumes that no 3D image data is available from the patient. The accuracy of pose estimation is validated with chest phantom images acquired under controlled geometric conditions. A subtle interval change simulated by a piece of plastic foam attached to the phantom becomes visible in subtraction images generated with this technique even at strong angular pose differences like an anterior-posterior inclination of 13 degrees.

Keywords: Digital radiography, temporal subtraction, pose estimation, image registration, chest

1. INTRODUCTION

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 automatic 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 artefacts ("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) cannot compensate for those image distortions that lead to crossings of shadows due to the perspective nature of radiography. If, for instance the position of the clavicle (anterior) and the posterior rib is flipped in two images due to a strong patient A-P inclination between both acquisitions, only either the clavicle or the posterior rib can be aligned by image registration, but never both. Still, these 2D registration methods as proposed in [3,7] can compensate to a certain extend.

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for the 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 to each of these images and it is perspective 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.

![Image](image1.png)

**Figure 1:** (a) Radiograph of the thorax phantom with subtle, almost invisible opacity in the central right lung field (marked by a box) caused by a piece of plastic foam attached to the phantom. Some fiducial markers are also visible in the image that were used to determine the reference pose (see text). They were attached to the phantom at the nape of the neck, the frontal shoulders, the belly button, and the posterior hips. (b) Conventional subtraction image with 1 deg rotation between both images. Despite some anatomical noise from posterior ribs and the scapula border the interval change is visible. (c) Conventional subtraction image with 13 deg rotation. Anatomical noise strongly masks the interval change invisible in this case.

## 2. MATERIAL AND 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. This corrected image is used in our proposed subtraction method. A subsection is dedicated to each of these two steps below.

### 2.1. Pose estimation

To estimate the phantom's pose from a single PA radiograph, a 3D surface model of the phantom's rib cage is matched to the radiograph such that the pose of the 3D model yields a good approximation of the phantom's pose during image acquisition. We used a triangulated surface model of seven rib pairs build from a number of diagnostic CT images from
previous work [11]. For the experiments reported here we used a version of it that was adapted to the thorax phantom by use of a CT image of that phantom.

We used the approach proposed in [2] for the application to a digital reconstructed radiograph (DRR). In the work reported here, we applied it to real radiographs of the phantom. 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

$$d(x_m, S_r) = \min_{x_r \in S_r} (\|x_m - x_r\|).$$

The distance is defined between contours of corresponding ribs only. Thus, the ribs need to be labelled 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

$$D_S = \int_{x_m \in S_m} d_e(x_m, S_r)^2 dx_m + \int_{x_r \in S_r} d_e(x_r, S_m)^2 dx_r.$$

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

$$T_p = P_a^{-1} P_p$$

with $P_a$ actual pose, $P_p$ previous pose, both with respect to an initial model pose.

2.2. Image correction

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 $T_f$, apply the approximated patient transformation $T_p$ and transform it back with $T_f$. This can be expressed in a single transformation $T$ when using homogeneous coordinates by
Its projection back to the image by $T_3$ gives $i'$, $\theta_x$ (A-P inclination) and $\theta_z$ (rotation around the patient longitudinal axis) are the most relevant components of $T_p$ (see Fig. 2). As $d_f$ is undetermined, it may be adjusted interactively while viewing its influence on the compensated subtraction image. It is expected that anatomical noise originating from structures located in the coronal $d_f$-plane disappear from the subtraction image.

![Figure 2: Compensating a pose difference: Original image position $i$ is projected by $T_1$ to $p$ in the coronal patient plane at $p,=d_f$. Estimated patient pose difference $T_p$ is applied to $p$ giving $p'$. Its projection back to the image by $T_3$ gives $i'$. $\theta_x$ (A-P inclination) and $\theta_z$ (rotation around the patient's longitudinal axis) are the most relevant components of $T_p$ in practice.](image)

## 2.3. Experimental settings

We acquired PA chest radiographs with a Philips Digital Diagnost X-ray machine with a distance source to detector $d_s=2000$mm. The pixel spacing was $144\mu m$. We used 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 placed at posterior-anterior (PA) position thus facing the detector and almost touching it with the breast. It 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 $\theta_z=+5$deg, $\theta_z=-5$deg, and A-P inclination $\theta_x=+3$deg). $\theta_z$ could be accurately adjusted with a high precision optical turn table, where the phantom was located on. Additional images with even...
higher angular pose differences to the baseline image were also made. 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. We achieved a mean error as low as 0.0014° between \( \theta_z \) read from the turn table scale and from the results of the point-based registration.

The rib silhouettes were manually segmented from these radiographs and the 3D rib cage model was adapted automatically with the method explained in the previous section 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. The translational difference between initial pose and reference was in the range of -30mm to 30mm for each axis, and the angular difference was in the range of -15° to 15° for each Euler angle.

![Image](image.jpg)

Figure 3: Exemplary pose reconstruction results of two phantom images: The reference surface in red illustrates the pose of the phantom during image acquisition; the beige surface shows the reconstruction result. (a) depicts one result of average quality of Image 3 and (b) one result of Image 4, where a slight error in translation along the sagittal axis \( t_y = -6\text{mm} \) can be observed in the lateral 3D model view (right).

3. RESULTS

Fig.3 shows exemplary results for two of the sample images. It gives a qualitative impression on the pose estimation accuracy. We also determined the quantitative 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. \( T_{p\text{-error}} \) is composed of a displacement error \( (t_x, t_y, t_z \text{ in mm}) \) and angular error \( (\theta_x, \theta_y, \theta_z, \text{ the Euler angles of } \theta \text{ in deg}) \). 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 Image 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 plane 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.4 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 difference \( T_p \) was determined and then used to generate a corrected subtraction image. \( T_p \) was obtained with an accuracy of \( T_{p-error} = (0.53, 2.74, 0.04, 0.14, 0.29, 0.78) \) and an average surface distance of 1.4mm.

The choice of \( d_f \) strongly determines the quality of the subtraction image of our method. 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 \( df \)-plane actually holds true. Other structures still 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 structures are (1st) a posterior rib and (2nd) the medial scapula border. Fig.4. shows subtraction images for those \( d_f \) values that correspond with the depth of these anatomical structures in the back of the patient, and the more anterior located clavicles, respectively. Compared to a conventional subtraction image calculated from the same image pair as shown in Fig.1c, both variants as shown in Fig.4b and Fig.4c appear superior, because the interval change is visible despite the strong pose difference.

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![Fig. 4: Novel 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, respectively. The interval change as shown in Fig.1 is clearly visible in (b) and (c), but masked by a posterior rib shadow in (a). Still, the scapula border appears as central vertical line in (b) and the posterior ribs appear in (c). Note: The depth of the simulated tumor cannot be determined as it is present in one image only.](image-url)
4. DISCUSSION

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 1.4mm to 6.5mm had no strong impact on the final subtraction image quality due to its comparably little influence on the projection image. Whenever the pose $T_\text{P}$ is estimated with a sufficient accuracy the method can only improve the quality of a subtraction image because the transformation $T$ does not cause any artifacts in the image. A combination with a subsequent classical non-rigid 2D-2D registration appears reasonable.

In its current implementation, the method relies on the manual segmentation and labeling 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 in the temporal subtraction framework yet.

In the experiments reported here, a rib model was used that was previously adapted to the shape of the ribs in the phantom via a CT image. In a typical clinical context, it cannot be expected that a thorax CT is available from a patient who should get a radiographic thorax examination. A fully automatic version of this method would thus further require the adaptation of a statistical shape model to an image by variation of both pose and shape parameters. Doing this for a single projection image seems quite challenging. Some preliminary work towards this end is reported in [2]. The experiments reported in this paper can be considered as an intermediate step towards a fully automatic method for clinical images.

5. 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. A temporal subtraction image is calculated for this compensated image. Experiments with a thorax phantom acquired under well controlled geometric conditions demonstrated the accuracy of the pose difference estimation. The 3D bone surface distance error is in the range of 1.4mm to 6.5mm in our experiments. As most of this error is oriented along the projection axis the error on the final subtraction image is less than this.

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 all known methods, this pseudo-3D method was able to compensate for a strong anterior-posterior inclination of 13deg and still render the subtle interval change. The results promise to make interval changes detectable and assessable easier than without this method even in those frequent cases where the patient pose differs strongly between follow-up acquisitions.
REFERENCES


