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Evaluation of Rigid-Body Motion Compensation in Cardiac Perfusion SPECT Employing Polar-Map Quantification

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Abstract

We have recently been successful in the development and testing of rigid-body motion tracking, estimation and compensation for cardiac perfusion SPECT based on a visual tracking system (VTS). The goal of this study was to evaluate in patients the effectiveness of our rigid-body motion compensation strategy. Sixty-four patient volunteers were asked to remain motionless or execute some predefined body motion during an additional second stress perfusion acquisition. Acquisitions were performed using the standard clinical protocol with 64 projections acquired through 180 degrees. All data were reconstructed with an ordered-subsets expectation-maximization (OSEM) algorithm using 4 projections per subset and 5 iterations. All physical degradation factors were addressed (attenuation, scatter, and distance dependent resolution), while a 3-dimensional Gaussian rotator was used during reconstruction to correct for six-degree-of-freedom (6-DOF) rigid-body motion estimated by the VTS. Polar map quantification was employed to evaluate compensation techniques. In 54.7% of the uncorrected second stress studies there was a statistically significant difference in the polar maps, and in 45.3% this made a difference in the interpretation of segmental perfusion. Motion correction reduced the impact of motion such that with it 32.8 % of the polar maps were statistically significantly different, and in 14.1% this difference changed the interpretation of segmental perfusion. The improvement shown in polar map quantitation translated to visually improved uniformity of the SPECT slices.

Index Terms

Cardiac Imaging; Motion Correction; SPECT

I. Introduction

Since the introduction of the first commercially available cardiac perfusion agent [1] and the demonstration of the combined use of computed and emission tomography [2], technological advances in cardiac perfusion single photon emission computed tomography (SPECT) enhanced patient throughput by clever gamma camera designs while the timely introduction of sequential multimodality imaging (SPECT-CT) brought to full realization Ter-Pogossian and colleagues [2] concluding remarks that both emission and transmission CT would be used in the assessment of cardiac pathology. However, patients are inevitably prone to move during a 10 to 15-minute acquisition due to the uncomfortable positioning (arms above head) or the distractions provided by the motion of the acquisition system especially between SPECT and CT (transmission). Furthermore, the left ventricle (LV) heart muscle thickness of a normal individual is only on the order of 2–3 voxels (9–14 mm)

depending on the SPECT acquisition matrix (64 or 128) and easy to erroneously relocate fully or misplace a portion to the left lung of the attenuation map when relatively small motions occur.

Until recently, motion detection in cardiac perfusion SPECT, focused on the consistency of projections or sinograms to estimate motion with application of the corrections in projection or image space [3–16]. The methods most frequently used were calculating projection by projection cross-correlations of the acquired perfusion data [3–4, 11, 13–14], usage of a radioactive marker placed on the patient [13], tracking the center of the heart using diverging squares [4, 14], or implementation of a projection re-projection algorithm which determines the best match between the original and re-projected projections [12, 15–16]. All the above-mentioned methods estimate motion with 3 degrees-of-freedom (3-DOF).

We have developed an integrated motion tracking and acquisition strategy [17–24], which enables rigid-body and respiratory motion compensation, either separately [19–22] or in some combination [17,18]. These corrections run with minimal initial user interaction employing a visual tracking system (VTS) from Vicon Motion Systems, Inc. (Lake Forest, CA) for motion tracking during SPECT acquisition. In preliminary investigations with patients who volunteered to undergo a second session of imaging where they were instructed to move our methods have shown good visual agreement when compared to the initial no motion induced baseline acquisitions [23–24]. In this study we report on the next phase of our investigations by evaluating the effectiveness of our rigid-body motion compensation methodology in 64 patient volunteers employing polar-map quantification [25]. Correction for respiratory motion was not included in this investigation as it focused on the estimation and correction of rigid-body motion occurring between the initial baseline clinical imaging and repeat imaging where the patients intentionally moved.

II. Methods and Materials

A. Patient Acquisitions

Patient volunteers (n=64, 27 females) with Institutional Review Board (IRB) approved written consent between the ages of 29 and 74 (54.4 10.9) and body mass indices varying from 19 to 47 (31.9 5.9) undergoing scheduled Tc-99m sestamibi cardiac perfusion rest-stress SPECT exams were asked to either execute some predefined body motion during an additional second stress perfusion acquisition (48 patients), or stay perfectly still during the second imaging study (16 patients). Seventeen patients were asked to execute one simple motion during the second stress acquisition such as moving to their left, right, or towards their feet. Two patients in this group also had the pillow under their heads pulled out during imaging, resulting in unbending the body. The rest of the patients (31) performed more complex motion regimes consisting of between two and five motions during the second acquisition. The complex motions were based on those technologists reported seeing patients performing during imaging in our clinic. The motions included patients moving one or both arms / shoulders, rolling to the left or right, sliding to the right or left, twisting slightly, and shifted the pillow supporting them. The motions performed by each patient in this group varied depending on their ability to move, and many of them might be expected to result in non-rigid motion between markers on the chest and the heart within the body. All

was a unique short pulse sequence 450 ms in duration introduced 25 seconds after the start of the visual tracking system and easily detected with heart rates up to 120 beats per minute. The third option did not need additional hardware as it only used the translational motion of the bed recorded by the VTS aligned with the bed motion recorded in the list-mode data file. This was accomplished by cross correlation and calculation of the minimum squared error between the two sets of data. The best alignment was obtained when the minimum squared error reached its smallest value.

C. Rigid-Body Motion Estimation

Processing was fully automatic and followed the following steps [19]. First the list-mode data was binned into 100 ms time frames for each of the two camera heads. Next the 30 Hz VTS positional measurements were down sampled to 10 Hz (100 ms) and synchronized with 100 ms emission data frames (as described in Section B.). Then the rigid-body and respiratory motion components in the 100 ms VTS frames of each retro-reflective marker were separated using total-variation based iterative smoothing modified to include a non-linear function to better preserve edges [21]. The 100ms time frames corresponding to each of the 32 projection angles acquired by each head were combined together to form the projections and the average rigid-body motion of the patient's chest relative to the first time frame was determined. This was followed by estimating the 6 degrees-of-freedom (DOF) rigid-body motion of the markers on the chest using singular-value decomposition also as described in [21]. Finally, reconstruction with rigid-body motion compensation was performed as detailed in the following section.

D. Reconstructions

All data were reconstructed with an ordered-subsets expectation-maximization (OSEM) algorithm using 4 projections per subset and 5 iterations. During reconstruction, all physical degradation factors were compensated for (attenuation, scatter, and distance-dependent spatial resolution) [26]. Additionally, during reconstruction 6-DOF rigid-body motion was corrected using a 3-dimensional Gaussian rotator as detailed in [27] by mimicking in iterative reconstruction the motion of the patient. The method is an extension of using a 2-dimensional Gaussian to rotate transverse slices with projection angle [28]. It combines in a single interpolation both patient motion and the rotation of the coordinate system with acquisition angle. With this method 3-dimensional Gaussian interpolation is employed to move the current emission voxel estimates and attenuation maps in the reference position in the global coordinate system to the moved patient location in the rotating coordinate system when calculating the expected projection. Gaussian interpolation also is then employed to move back the backprojection of the ratio of the measured projection to the expected projection and backprojection of the unit value (sensitivity factor) to the reference location. Rigid-body motion compensation was also performed on baseline acquisitions while second stress acquisitions were reconstructed without motion compensation for comparison. All reconstructed data were post filtered using a 3D Gaussian with a standard deviation of 0.466 cm [26].

E. Image Analysis and Evaluation

All reconstructed transaxial slices were reoriented to short-axis, horizontal long-axis, and vertical long-axis slices. Polar maps were subsequently generated using our own software developed in IDL [25, 29]. Polar maps were compared to a normal database created with 20 patients acquired and reconstructed in the same manner as the clinical stress patient volunteers described earlier. The normal database consists of two polar maps. The first is the normalized average values for each polar map pixel over the 20 patients and the second the standard deviation (SD) among the 20 patients also on a pixel-by-pixel basis. The normalized baseline, second stress, and motion-corrected second stress polar maps were subtracted from the normal database polar map of normalized average values and negative pixel values (those larger than the average) set to zero. The resulting difference polar maps were then divided pixel-by-pixel by the SD polar map. The values thus obtained reflected the number of SD's below the average pixel values on the normal database polar map. Averages of these values were then determined for 17 segments of the polar map defined according to the ASNC guidelines [30]. A one-tailed paired Student t-test was used on each volunteer's study to determine if there was a statistically significant difference in the SD's of the 17 segments between the second-stress study with and without motion compensation compared to the initial clinical or baseline study. A further check was then made to determine if this statistically significant difference resulted in a change in interpretation as to whether the segmental perfusion was normal or not. This was accomplished by determining whether the number of SD for any of the segments crossed the threshold of 2.5 separating normal from abnormal [30]. In cases where the baseline polar map was normal this indicated a false indication of the presence of disease and in cases where patients were already classified with disease on the initial baseline polar map, a reversal to above 2.5 SD's indicated a masking of the disease. Thus two levels of failure in motion correction were determined. The first was a statistically significant change in the map form that of the baseline, and the second was when this change also resulted in a change in clinical interpretation.

III. Results

Fig. 2–7 show reconstructed slices, polar-maps, and motion graphs from three example patients. The numerical results for the number of SD's below the average of the database of 20 normal studies for each of the 17 polar-map segments for the initial clinical acquisition (baseline) and the second acquisition either with or without motion correction are given in Table 1 for these three patients. All images are displayed to the maximum in the heart for the specific method used. No lower thresholding or other additional manipulation of display contrast was employed.

For the first example patient, motion was estimated in the 1 cm range with up to 7 degrees of rotation around the z-axis (head-to-feet) as seen in Fig. 3 and the deterioration of image fidelity is visible in both the bottom row of short-axis slices and the polar-map on the right for reconstruction without motion correction (Fig. 2). SD's in Table 1 show the anterior-lateral segment crossover the 2.5 SD threshold indicating disease present, while the segments surrounding that area also deteriorate but not beyond 2.5 SD's. Furthermore, the

inferior segments also show a marked deterioration. The one-tailed paired t-test shows a significant difference between the baseline and no compensation ($p < 0.001$). No significant difference is observed when the baseline is compared with motion compensation ($p = 0.106$).

Body motion for the second patient example was estimated in the range of 3 cm with up to 7 degrees of rotation (Fig 5) around the z-axis in the opposite direction than example 1. The horizontal long axis on the bottom row and polar maps (center and right) (Fig. 4) clearly show the deterioration even when motion is compensated for. The one-tailed paired t-test shows significant differences for both no motion compensation ($p = 0.008$) and motion compensation ($p = 0.002$), however only the no motion compensation polar map indicates disease (> 2.5) in the apex and anterior apex.

Significant uncorrected motion ($p < 0.001$) is present in the last patient example (Fig. 6) in 7 segments, while motion correction improves image fidelity ($p = 0.351$) although some deterioration can be seen in the inferior and inferior septal regions. In summary, the three examples demonstrated that the type and duration of motion have an influence in where the motion artifact appeared.

In analyzing the impact on polar map quantification of motion during the second stress acquisition in comparison to the baseline acquisition it was determined that 43.8% of the 64 patients showed a change in diagnostic outcome (i.e., normal changing to abnormal or abnormal changing to normal). This was reduced to 14.1% with motion correction. Of the 56.3% which showed no change in outcome in the absence of motion correction in 10.9% the SD's in the 17 segments were statistically significantly different. In the 85.9% that showed no change in outcome when the motion corrected second stress studies were compared to the baseline studies 18.8% were determined to be statistically different. Thus the monitoring of external markers was seen to definitely restore the diagnostic value of many, but not all of the studies to that of the baseline study.

IV. Discussion

The goal of this investigation was to determine if we could restore the images acquired when patients intentionally moved to having the same diagnostic value as judged by polar map quantification as studies acquired when the patients did not intentionally move (the original or baseline studies). Patients acquired with no intentional motion were also included to determine if our methodology would detect fallacious movement in these patients and thereby degrade the slices such that there also was a change detected by polar map quantification. Correction of the rigid-body motion estimated by our VTS resulted in a decrease from 43.8% to 14.1% in the studies in which the diagnostic outcome changed between the baseline and second stress acquisitions. This would be a definite improvement clinically. The 14.1% of acquisitions where the outcome changed after motion compensation represented the fraction of studies where the VTS failed to accurately predict the internal motion of the heart from the exterior. This is likely due in part to the inherent non-rigid nature of the motion between the surface of the body and heart for motions such as arm / shoulder motion, twisting, and bending / unbending which we requested to be performed by the patients. These pose a challenge to exterior marker based estimation of interior motion

[31]. The 14.1% also included practical difficulties such as slippage of the stretchy belt holding the markers around the chest causing an over or underestimation of motion [31], and IR camera LED reflections off the collimator face and other camera components causing errors in marker tracking. We are currently investigating ways to decrease the impact of such practical difficulties.

Three markers are the minimum necessary to compute 6-DOF rigid-body motion. Retro-reflective marker motion during execution of prescribed motion should be correlated in the sense that those markers in close proximity of each other will have similar directional motion (x, y, z, displacement). The VTS used five markers on the chest which gave some redundancy. The additional retro-reflective marker on the right lower rib was an additional location on the rib cage ('assumed' to be rigid object in our case) to provide a third independent motion vector together with that on the right and left chest.

This study employed actual patients who volunteered to undergo a second SPECT acquisition. Although we requested the patients to perform specific motions the actual nature and extent of the performed motion was beyond our control. This was to some extent a limitation in the experimental design of this study. Further, we corrected based on rigid-body motion based on the average motion between the given projection and the first as opposed to correcting for the varying intra-projection motion of the patient which may have improved our ability to correct motion as the patients were requested to move during camera acquisition and not during the time period between frames when the camera rotated [32]. Also, respiration between the baseline and second stress study might have changed (although rarely according to our observations). This might explain in part some of the variation between the results of the two imaging sessions. However, since respiratory motion correction was not the focus of this study employing repeated patient imaging, it was not included herein. Respiratory motion correction will be the topic of a subsequent paper employing just the clinical stress imaging of a much larger population of patients. Correction of any incidental patient motion will be also included in that study. Also we did not investigate motion compensation in cardiac gated slices. Since the motion estimates derived from the markers would be the same one would expect a similar level of success with employing them to correct cardiac gated studies.

V. Conclusions

In this study of 64 cardiac perfusion patients who volunteered to undergo a second SPECT acquisition in which 75% of them intentionally moved, we determined through polar map quantification that in the second acquisition there was a change in diagnostic outcome (i.e., normal changing to abnormal or abnormal changing to normal) in 43.8% of the cases. This was reduced to 14.1% with motion correction. We have also provided the results of three example patient studies which illustrate mild to major artefactual changes in slices which were either completely or in large part compensated for by our VTS based correction. Thus motion correction based on monitoring the motion of external markers was determined to be able to correct motion in many of the studies, but not all. Since we corrected based on the average motion for each projection relative to the first frame, motion correction might have helped reduce the number in which there was a change in diagnostic outcome, as perhaps

also would have correction for respiratory motion. However, practical limitations in external marker motion tracking, estimation of motion from these markers, and the non-rigid nature of the relationship between motion of markers on the chest surface and the internal motion of organs likely was a restraining factor in many of these studies when a change was determined. In future investigations we plan to determine if the practical limitations can be addressed to provide even more robust motion compensation.

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Fig. 1.
The latest placement of the retro-reflective markers on a patient, 5 on the chest, one on the right rib, and one on the abdomen.

Fig 2.

Patient example 1 with motion graphed in fig. 3 is depicted. Short-axis slices of baseline, motion corrected, and no motion correction are displayed from top to bottom, with the polar maps in the same order from left to right. The green arrows point to the location of an artefactual inferior wall decrease caused by motion, and removed by motion compensation.

Fig 3.

Translational and rotational motion of patient example 1 during the SPECT acquisition is shown in the graphs. X in the legend refers to left-to-right motion and rotation around that axis, Y referred to inferior-posterior motion as well as rotation around that axis, and Z referred to axial (head-to-feet) motion and rotation around that axis. Thirty-two motion values (48–80) are shown on the x-axis coinciding with the number of projections acquired on one head out of a total of 64 projections over 180 degrees. The same motion occurs in the 32 projections acquired on the other head.

Fig 4.

Patient example 2 with up to ~3 cm motion is depicted. Horizontal long-axis slices are shown as they best demonstrated the presence of motion. The baseline, motion corrected, and no motion correction are displayed from top to bottom and polar maps in the same order from left to right. The green arrows point to the location of a substantial artefactual apical defect caused by motion, and removed in large part by motion compensation.

Fig 5.

Translational and rotational motion of patient example 2 during the SPECT acquisition is shown in the graphs. X in the legend refers to left-to-right motion and rotation around that axis Y referred to inferior-posterior motion as well as rotation around that axis, and Z referred to axial (head-to-feet) motion and rotation around that axis. Thirty-two motion values (48–80) are shown on the x-axis coinciding with the number of projections acquired on one head. The same motion occurs in the 32 projections acquired on the other head.

Fig 6.

Patient example 3 with up to 1.6 cm motion in the axial direction and 0.9 cm lateral motion is depicted. Short-axis slices of baseline, motion corrected, and no motion correction are displayed from top to bottom, with a similar ordering of the polar maps in the same order from left to right. The green arrows point to the location of a major artefactual inferior defect with an associated inferior / lateral apparent increase in localization caused by motion, and with both significantly diminished by motion compensation.

Fig 7.

Translational and rotational motion of patient example 3 during the SPECT acquisition is shown in the graphs. X in the legend refers to left-to-right motion and rotation around that axis Y referred to inferior-posterior motion as well as rotation around that axis, and Z referred to axial (head-to-feet) motion and rotation around that axis. Thirty-two motion values (48–80) are shown on the x-axis coinciding with the number of projections acquired on one head. The same motion occurs in the 32 projections acquired on the other head.

Table 1

Shown are the number of standard deviations below average of the normal database for the 3 example patients of figures 2–7, respectively, for the Initial (Base) study, and the second-stress study with (Cor) or without (No) motion correction.

#	Example 1			Example 2			Example 3		
	Base	Cor	No	Base	Cor	No	Base	Cor	No
1	0.72	0.76	1.57	0.59	0.61	0.19	0.97	1.00	2.55
2	0.03	0.02	0.07	0.51	0.48	0.14	0.09	0.09	0.94
3	0.03	0.35	0.00	1.13	1.00	0.85	0.38	0.52	0.50
4	0.83	0.79	0.99	0.46	0.79	0.98	1.05	1.10	5.43
5	0.33	0.89	0.53	0.00	0.00	0.07	1.00	0.62	0.93
6	0.02	0.22	1.19	0.09	0.18	0.74	1.00	0.84	3.88
7	0.97	0.98	1.91	0.02	0.61	1.27	1.16	1.00	2.98
8	0.57	0.36	1.69	0.01	0.02	0.04	1.03	0.94	3.03
9	1.00	1.38	1.22	0.76	1.40	0.25	0.98	1.93	1.73
10	0.89	1.29	1.89	0.60	1.08	0.59	1.11	1.52	5.84
11	0.43	0.54	0.61	0.03	0.40	0.67	0.35	0.79	0.80
12	0.51	0.39	3.15	0.00	0.73	1.28	0.94	0.97	5.95
13	0.95	0.76	1.26	0.00	0.51	3.22	1.00	0.39	1.84
14	0.91	0.65	1.23	0.01	0.51	0.91	1.00	0.46	1.45
15	0.96	1.87	2.45	0.63	1.71	1.01	0.82	0.99	2.05
16	0.44	0.57	1.84	0.00	0.99	1.67	0.45	0.44	1.49
17	0.96	0.88	1.00	0.00	0.87	2.75	0.99	0.01	1.01

is the polar-map segment number with: 1 – base anterior, 2 – base anterior-septal, 3 – base inferior-septal, 4 – base inferior, 5 – base inferior-lateral, 6 – base anterior-lateral, 7 – mid anterior, 8 – mid anterior-septal, 9 – mid inferior-septal, 10 – mid inferior, 11 – mid inferior-lateral, 12 – mid anterior-lateral, 13 – apical-anterior, 14 – apical-inferior, 15 – apical-septal, 16 – apical-lateral, and 17 – apex.