Use of MRI to assess the prediction of heart motion with gross body motion in myocardial perfusion imaging by stereotracking of markers on the body surface

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Purpose: The aim of this study is to determine using MRI in volunteers whether the rigid-body-motion (RBM) model can be approximately used to estimate the gross body-motion of the heart from that of external markers on patient's chest. Our target clinical application is to use a visual-tracking-system (VTS) which employs stereoisimaging to estimate heart motion during SPECT/CT and PET/CT myocardial perfusion imaging.

Methods: To investigate body-motion separate from the respiration the authors had the volunteers hold their breath during the acquisition of a sequence of two sets of EKG-triggered MRI sagittal slices. The first set was acquired pre-motion, and the second postmotion. The motion of the heart within each breath-hold set of slices was estimated by registration to the semiautomatic 3D segmentation of the heart region in a baseline set acquired using the Navigator technique. The motion of the heart between the pre- and postmotion sets was then determined as the difference in the individual motions in comparison to the Navigator sets. An analysis of the combined motion of the individual markers on the chest was used to obtain an estimate of the six-degree-of-freedom RBM from the VTS system. The metric for judging agreement between the motion estimated by MRI and the VTS was the average error. This was defined as the average of the magnitudes of the differences in the vector displacements of all voxels in the heart region. Studies with the Data Spectrum Anthropomorphic Phantom and "No-Motion" studies in which the volunteer did not intentionally move were used to establish a baseline for agreement. With volunteer studies a t-test was employed to determine when statistically significant differences in Average Errors occurred compared to the No-motion studies.

Results: For phantom acquisitions, the Average Error when the motion was just translation was 0.1 mm. With complex motions, which included a combination of rotations and translations, the Average Error increased to 3.6 mm. In the volunteers the Average Error averaged over all No-Motion acquisitions was 1.0 mm. For the case of translational motion, which might be expected to be RBM, the Average Error averaged over all volunteer studies increased to 2.6 mm, which was statistically different from the No-Motion studies. For the case of bends and twists of the torso, which would be expected to challenge the RBM model, the Average Error averaged over all such volunteer studies was 4.9 mm and was again statistically different. Investigations of motion of the arm including just bending at the elbow and leg motion resulted in Average Errors which were not statistically different from the No-Motion studies. However, when shoulder movement was included with arm motion the Average Error was near that of torso bends and twists, and statistically different.

Conclusions: Use of the RBM model with VTS predictions of heart motion during reconstruction should decrease the extent of artifacts for the types of patient motion studied. The impact of correction would be less for torso bends and twists, and arm motion which includes the shoulders. © 2013 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4824693>]

Key words: SPECT, PET, motion, stereoisimaging

1. INTRODUCTION

Patient motion during imaging can impact diagnosis by causing a loss of resolution due to blurring and artifacts resulting from inconsistencies in projection data. Motion can be physiological in origin such as heart contraction and respiration,¹ or gross body-motion, perhaps to relieve discomfort or as a result of coughing. Gross body-motion is of significant concern in emission imaging (SPECT and PET) due to the extended period of time the patient must remain motionless while the projections are acquired.²⁻⁹ In sequential hybrid imaging it also can impact attenuation correction when motion occurs between emission imaging and CT or transmission imaging employed to estimate the attenuation maps.¹⁰⁻¹⁹ In emission imaging body-motion can be estimated from the imaging data itself,²⁰⁻³¹ or by some surrogate such as markers on the external surface of the patient.^{24,32-37} The later approach has the advantage that it can also be applied to track patient motion in sequential hybrid-imaging. The drawback of using such surrogates is that some model of the relationship between the movement of the organ(s) of interest clinically inside the body and the external markers must be employed as part of motion correction. In the case of brain imaging where the brain is encased in the rigid skull, the use of a rigid-body-motion (RBM) model has been successfully employed for motion correction using the tracked motion from markers fastened in various ways externally to the head.³³⁻³⁶ The internal organs of the chest are bounded by the ribcage, spine, and sternum, thus in the absence of motion caused by respiration one might hypothesize that RBM estimated from markers on the chest surface might be useful at least as a starting point for predicting

the motion of individual organs within the chest. Many studies have been performed to investigate modeling the internal motion of structures within the chest from markers on the body surface as a result of respiratory motion.³⁸⁻⁴⁷ However, we have not found any investigations of the linkage between external markers and internal organ motion for gross body-motion during imaging.

In this study we investigate the accuracy with which the gross body-motion of the heart as determined by MRI can be predicted from simultaneously tracked markers on the chest of volunteers. Knowledge of the accuracy with which such motion can be predicted externally could have applications to organs in the chest other than the heart, and therapeutic applications. Our target clinical application is motion estimation and correction for patients during list-mode SPECT and PET myocardial perfusion imaging (MPI) and sequential transmission/CT imaging. As illustrated in Fig. 1, we have developed a stereoisimaging based visual-tracking-system (VTS) for this purpose.⁴⁸ From the stereotracking of markers (~1 cm diameter spheres covered with a retroreflective material) on stretchy bands wrapped about the patient's chest and abdomen we obtain information for use in the correction of both respiratory and gross body-motion. To do this we separate the tracked marker motion into periodic respiratory motion and residual body-motion components.⁴⁹ The body-motion component of the chest-marker motion is then employed to correct the heart for gross body-motion making usage of the RBM assumption in iterative reconstruction.⁵⁰ The respiratory component of the motion of the abdominal markers is then used to correct for respiratory motion of the heart as detailed in Ref. 51. Our clinical results with body-motion correction in

FIG. 1. (Left) Volunteer undergoing simulated SPECT/CT imaging as seen through lead-glass window of control room. Shown are the backs of two of the five near-infrared cameras of the VTS. (Right) Close-up view showing markers on volunteer's chest (arrows).

SPECT thus far have been encouraging in terms of the improvement in image quality with motion correction;⁴⁸ however, we are left with questions regarding the suitability of the RBM model for such correction and for applications beyond ours.

Therefore the aim of this study is to determine by MRI in volunteers the extent to which the stereomaging predictions of body-motion based on tracking external markers matches the actual motion of the heart within the chest. This knowledge can then be used to understand the merits and deficiencies of such an approach, and where deficiencies are observed point out areas requiring further investigation to improve the tracking of the heart during gross body-motion with external markers.

2. MATERIALS AND METHODS

2.A. VTS operation and calibration in MRI room

The cameras employed by our VTS were the MX-3 cameras from Vicon, Inc (Los Angeles, CA) with near-Infrared emitting LED's whose emission are reflected back from the retroreflective coating on the markers we employ to the CCD detectors of the cameras. We have previously described these cameras in Ref. 48. The marker layout was the same as we employ for clinical SPECT imaging and illustrated in Fig. 2 (left). The chest markers were fastened to a Velcro strip which was approximately 10 cm wide by 25 cm long. This was in turn attached to the 10 cm wide stretchy band of self-sticking bandage material wrapped around the volunteers body, and stretched to hold in place but still allowed the volunteer to breath freely. The 1 cm diameter markers we use clinically were however replaced by 2 cm diameter markers which were hollowed-out and filled with a copper sulfate solution for visualization in MRI as well as by the VTS. This switch to the 2 cm markers was essential for establishing the transformation between the VTS and MRI coordinate systems. Two markers

were positioned on the right side of the chest, and three more were on the left side. No markers were placed in the central portion of the chest as due to the shallow angle of imaging they would have been obstructed by markers on the abdomen band which were positioned medially. The markers on the chest band covered the width of the Velcro in the axial direction to allow determination of rotation about the lateral axis. Two markers were similarly attached to a 7.5 cm wide band wrapped around the abdomen. We have previously reported our methodology for determining the six-degree-of-freedom (6-DOF) motion of the markers on the chest of patients.⁴⁹ Thus we will not detail this herein. Within iterative reconstruction when calculating the projections of the estimated source distribution, 3D Gaussian interpolation⁵⁰ was used to apply this 6-DOF transformation to the current source estimate and attenuation map slices in the reference position to match the pose of the patient at the time of acquisition. Gaussian interpolation is also employed to transform the backprojection of the ratio of the measured projection to the estimated projection and backprojection of the unit value (sensitivity factor) to the reference location.

Before bringing the VTS cameras into the room with the strong magnetic fields of our 3T Philips Achieva MRI System, the cameras were tested for the presence of ferromagnetic components. Only the connector on the MX-3 cameras for the signal cables from the acquisition box exhibited attraction to a test magnet. Even with this small amount of ferromagnetic material being present, care was taken to stay along the walls of the MRI room where the field strength is at its lowest when moving the cameras into and out of the MRI room. Aluminum mounts were purchased and installed on the walls of the MRI room by aluminum screws. The 3-way pan/tilt heads upon which the cameras were attached to the wall mounts were also made of aluminum with the exception of ferromagnetic handles, which were replaced with aluminum ones made in our machine shop. The cables and connectors between the cameras and the acquisition box were

FIG. 2. (Left) Layout of markers used herein for volunteers and in clinical studies. (Right) VTS cameras, 3-way pan/tilt heads, wall mounts, and cables on wall MRI room.

FIG. 3. Calibration phantom consisting of seven noncoplanar 2-cm diameter retroreflectively coated spheres filled with a copper sulfate solution used to develop the transformation to convert measurements in the VTS coordinate system into MRI coordinate system.

passed out of the room through a wave-guide to minimize RF interference with MR imaging. The three cameras of the VTS system with pan/tilt heads, wall mounts, and cables are shown in imaging position on the wall of the MRI room at the right in Fig. 2. The pan/tilt heads were adjusted with a volunteer on the table of the MRI wearing bands with markers to align the MX-3 cameras for marker tracking.

Before each usage the VTS system is calibrated for stereo-motion tracking by waving a calibration tool with three markers on it within the volume of interest for imaging while capturing images from all three MX-3 cameras at 30 frames per second.⁴⁸ The Vicon, Inc. software is then employed to provide the 2D to 3D mapping of marker locations from the images acquired by the cameras. The calibration tool used in the MRI room was specifically built of nonferromagnetic materials by our machine shop. We have previously shown that the VTS agreement with the SPECT system for measuring the motion of a set of seven markers with a small amount of Tc-99 m inside is well sub-mm and sub-degree.⁴⁸

To convert the VTS estimated 6-DOF RBM estimate in the VTS coordinate system to that of the MRI coordinate system a phantom consisting of seven of the 2 cm diameter markers was constructed such that the markers were not coplanar (Fig. 3). The phantom was positioned with laser guidance such that a marker centrally located in the phantom was approximately at the isocenter of the MRI system. The phantom was imaged by both MRI and the VTS systems. The location of the centers of the markers in the MRI was determined as the center-of-mass of the MR intensities in the slices. The correspondence of the markers in MRI and VTS was determined by matching them according to their distance from the center of the seven marker locations. Once matched, as we have previously detailed⁵² SVD analysis of the centers of the seven spheres was used to determine the transformation between the VTS coordinate system and the native MRI coordinate system. After applying the transformation, both the MRI

and VTS axes were centered at the isocenter of the MRI. The resulting x-axis was in the vertical direction (posterior to anterior), the y-axis was in the axial direction (head to foot), and the z-axis was in the lateral direction (right to left).

We accessed the accuracy of this calibration by determining the difference in 3D between the centers of the markers as determined by the VTS after applying the transformation and the MRI for each of the seven spheres of the phantom over the period our studies were acquired. The average agreement was 1.28 mm with a standard deviation (SD) of 0.49 mm. The finite extent of the difference probably relates to MRI voxel size and the impact of potential imperfections in hollowing out of the markers so that the centers for VTS and MRI could have actually been slightly different.

2.B. Methodology of MRI acquisition for studying body-motion in volunteers

Our methodology for investigating body-motion separate from the influence of respiratory motion was as follows. We first acquired an initial slab of 60 EKG triggered sagittal slices with a matrix size of 128×128 centered on the heart with VTS tracking. A delay was used in triggering such that the acquisitions occurred during the portion of diastole while the heart was relatively quiescent prior to the atrial kick. The slices consisted of cubic voxels of 3 mm on each side and there were no gaps between the slices. The Navigator methodology⁵³ was used to restrict the presence of respiratory motion between slices as illustrated in Fig. 4. With this method slice acquisition occurred only when the dome

FIG. 4. (Top) Shown is the plot as a function of time of the one-dimensional signal acquired in the superior/inferior direction from a Navigator sampling volume centered on the dome of the right diaphragm. (Bottom) Shown is a coronal slice made from the acquisition of sagittal slices within the Navigator acceptance window acquired over a period of minutes.

of the right diaphragm was within a 3 mm wide Navigator acceptance window about its end-expiration (EE) location within the Navigator sampling volume. The acceptance window is shown as horizontal blue lines in the figure. The EE location of the dome was determined as the average location of the lung/soft tissue interface during the first eight seconds of MRI acquisition. As illustrated in the figure, the volunteers were instructed to hold their breath at a comfortable EE position during this period of time to enable efficient acquisition. The volunteers were trained to thereafter rapidly breathe out and back in, returning to the same EE state as that established at the start of MRI. Verbal feedback was provided to the volunteers during imaging as to whether they should breathe out more or less so that their diaphragm level was within the acceptance window. The thick horizontal red line segments in the figure show the MRI system determined locations of the diaphragm at each of the EKG acceptable potential MRI acquisition time points. With training, we were able to achieve greater than a 75% gating efficiency. That is, 75% of potential acquisitions had the Navigator determined locations of the right diaphragm within the acceptance window and were thus acquired. Note in the coronal slice shown at the bottom of Fig. 4 how the use of the Navigator technique has effectively “frozen” respiratory motion for the acquisition of sagittal slices acquired over a period of minutes. The parameters for these and all subsequent MRI acquisitions were 2D T1-Weighted-Fast Field Echo, with TR/TE = 5.5 ms/3.1 ms and a matrix size of 128×128 . Solely the body coil of the system was employed for acquisition.

Following the Navigator acquisition, we acquired a sequence of two sets of 11 sagittal slices centered on the heart as above except with 9 mm gaps between slices. One set was acquired premotion and the other postmotion. Throughout both MRI acquisitions and while performing the requested motion, the volunteers were requested to hold their breath at EE. The period of breath-hold was 30 s or shorter depending on heart rate and time taken in performing the motion. If the volunteer had difficulty holding their breath for this duration the number of sagittal slices was reduced.

We linked the acquisition of the two sets of MRI slices with acquisition initiated by the MRI operator for each set. Experiments were then carried out by the operator:

1. Giving the volunteer instructions as to the type of motion they will be asked to perform between acquisition of the two sets of slices and coaching the volunteer while they practiced the motion;
2. Asking the volunteer signal when holding their breath at EE and thus ready for acquisition to begin by squeezing a “ball” attached to an alarm;
3. Initiating the first acquisition when the alarm sounds;
4. Asking the volunteer to perform the motion upon the completion of the premotion acquisition;
5. Starting the postmotion acquisition when the alarm sounds indicating the volunteer has moved and was now stable in the second position; and
6. Informing the volunteer they can now breathe freely at the end of the postmotion acquisition.

During imaging the volunteers laid on the MRI table with their hands over their heads on the same support as employed during cardiac SPECT imaging in our clinic. VTS imaging is performed throughout the MRI acquisition period for each motion, for comparison of the estimated motion by VTS of the surface of the chest with that by MRI of the heart within the body.

The motions performed by each volunteer were of six types. The first was that they did not move between the two sets of MRI acquisition. These “No-Motion” studies provided a baseline for the agreement between the VTS and MRI when there was little volunteer motion. The second was for axial or lateral translations, which represent the majority of patient body-motion as noted in the literature,^{4,6,7} and observed by us when analyzing motion in a VTS investigation of 110 cardiac SPECT studies.⁹ The third type of motion was to have the volunteer bend such that their shoulders moved to the right or left while their hips remained relatively in place. Alternatively they were asked to twist about their backbone to the right or left such that they rolled their shoulders up off the imaging table on one side while keeping their hips in place. Although these were not commonly identified as motions of patients^{4,6,7} or observed in our study,⁹ it was believed these motions would include some non-RBM by the volunteer thereby challenging our usage of the RBM motion model for correction.⁵⁴ Through discussions with our imaging technologists we empirically determined that arm and leg motion are the most common types of patient motion, other than respiratory. We therefore included the following three types of motion in our investigations to determine if they affected the location of the markers or heart, and if they did to what extent the motions were similar. The fourth type of motion was to have the volunteer move their left or right upper arm involving the elbow but not the shoulder. This was the most common of these types of motion as per our discussions. The fifth type of motion was to have the volunteer move their left or right arm including the shoulder by stretching one of their arms above their head from its normally bent position. Finally the sixth type of motion was for the volunteer to move one of their legs from being next to the other to the side of the imaging table. Other types of motion our technologists reported as commonly encountering were coughing and gradual drifts in position. We did not investigate these as volunteers would be unable to perform them while holding their breath.

2.C. Anthropomorphic phantom studies of agreement between VTS and MRI

Prior to the above human volunteer studies, we performed studies with the anthropomorphic phantom from Data Spectrum Corporation (Hillsborough, NC) to select the extent of the gaps between breath-hold slices employed in the human volunteer studies. The phantom acquisitions also were to serve as a way of determining the degree to which our methodology for estimation of motion from the MRI slices agreed with the VTS estimates for a truly rigid-body. For VTS motion tracking, the 2 cm diameter markers filled with

copper sulfate were attached to the anthropomorphic phantom in the same layout as employed in the volunteer studies. The phantom's surfaces were covered with nonreflective paper to avoid reflections which interfered with correct marker identification by our VTS. The "myocardial" wall chamber of the heart phantom inserted within the body phantom was filled with oil to highlight its difference from the other organs simulated within the phantom. Similarly the lungs were filled with a mix of small diameter Styrofoam beads and water, and the backbone was simulated by a Teflon rod which was very low in MR signal as is bone. An experimenter lying on the MRI table on the end away from the VTS moved the phantom after the premotion set of slices was acquired and signaled with the alarm to begin the acquisition of the postmotion slices. This was done to mimic the acquisition process with the human volunteers. Given the tight "fit" within the bore of the MRI system and the limitations imposed by the MRI environment on what we could bring into the MRI room, we made no attempt to provide another method of measuring the motions of the phantom independent of the VTS and MRI. Thus such measurements are unavailable for usage as a truth to compare both VTS and MRI estimates against. Since we did not have to worry about either respiratory or cardiac motion of this phantom, cardiac triggering and Navigator gating were not included in acquisition. Otherwise the voxel dimensions and MRI acquisition parameters were as employed with the human volunteers, except as follows. A "Navigator-less" set of 60 sagittal slices with no inter-slice gaps spanning the heart in the phantom was first acquired. This was followed by the pre- and postmotion acquisitions consisting of 41 slices each with no gaps between them. The experimenter on the imaging table performed the following six motions to the phantom between the pre- and postmotion MRI acquisitions: (1) No-Motion; (2) axial translation, (3) rotation about the vertical or AP axis, (4) rotation about the axial axis, (5) rotation about the lateral axis, and (6) a mix of translations and rotations. For motions which lifted one side or end of the phantom off the imaging table, padding was placed under the phantom so that it could remain stationary during the period of acquisition. This process imparted some translational motion to the phantom besides the desired rotation, as will be illustrated in Sec. 3.

MRI estimation of heart motion is detailed in Sec. 2.D. Using this methodology we investigated the impact of slice gaps by estimating the motion in the absence of gaps, and then with gaps of 3–27 mm in steps of 3 mm by zeroing the slices which were to correspond to gaps. The MRI estimated motions were also compared to those estimated by the VTS to determine the agreement between the two which could be expected in the case of true RBM.

2.D. Estimation of motion of the heart from the MRI slices

The motion between the breath-hold slice sets (volumes) and the Navigator volumes was estimated by 6-DOF intensity-based registration. The criterion used to determine the transformations was the minimization of the sum-squared-

difference (SSD).⁵⁵ The method is similar to that we previously performed to estimate motion of the heart in respiratory gated CT slices.⁵⁶ Instead of trying to register the sparse slices of the pre- and postmotion breath-hold datasets directly, we used the dense Navigator slices as an intermediary to find the transformation between them. This strategy is based on the work of King *et al.*⁵⁷ who used it to enable respiratory motion correction in x-ray guided catheterization using a motion model derived from MRI. Since we were interested in an estimate of the 6-DOF motion of the heart, we segmented the heart in the Navigator slices and then registered this segmented volume with the pre- and postmotion datasets. Then from the transformations obtained in this way, we calculated the transformations to align the pre- and postmotion datasets themselves (i.e., the 6-DOF motion of the heart within the chest).

The specific steps in this process were as follows. First we filled the 9 mm gaps in the breath-hold pre- and postmotion slice sets with three 128×128 slices of zeros, and then zero padded before the first and after the last slices of these sets so that the breath-hold volumes were the same size as the Navigator volumes.

In the second step we segmented the heart in the Navigator volumes so that it is only the heart to which we were registering the breath-hold volumes. For segmentation we assumed an ellipsoid as a prior model for the 3D shape of the heart.⁵⁶ The ellipsoid was placed roughly about the heart while carefully excluding the liver region and the major blood vessels superior to the heart using a user-interactive interface. The level-set based distance map evolution used in⁵⁶ did not add significant benefit in this study. This was because of the excellent contrast we obtained between the heart and the subdiaphragmatic structures as illustrated in the coronal slice at the bottom of Fig. 4. Thus solely the ellipsoid was used to define the volume employed in registration.

The third step was to register the pre- and postmotion breath-hold volumes to the Navigator volumes using 6-DOF rigid-body registration. When registering we kept the sparse-sliced pre- and postmotion volumes stationary and transformed the segmented Navigator volume with complete slice sampling in an iterative registration process to match them.⁵⁶ What we estimated by this process was the transformation required to move each breath-hold volume to the corresponding Navigator volume. The SSD and the 6-DOF parameters converged within 3000 iterations for most cases, while some required 4000–6000 iterations.

In the final step the two transformations from prebreath-hold and postbreath-hold to the Navigator volumes were then used directly to calculate the desired transformation of the heart inside the chest as follows. Let R be the desired 3×3 rotation matrix of the transformation between the pre- and postmotion volumes, T be a three valued translation vector between the premotion and postmotion volumes, and (R, T) be the 6-DOF transformation. Also let (R_1, T_1) denote the estimated transformation between the premotion and Navigator volumes, and (R_2, T_2) that between the postmotion and Navigator volumes. Then to find the transformation from pre- to postmotion we can apply the first transform (to go from

TABLE I. Average error (mm) for VTS versus MRI 6-DOF estimates for motion of Data Spectrum Anthropomorphic Phantom with indicated gaps employed pre- and postmotion for sagittal slices registered to initial reference acquisition.

Motion type	No gap	3 Slices (9 mm gap)	6 Slices (18 mm gap)	9 Slices (27 mm gap)
No-Motion	0.1	0.1	0.1	0.1
Axial translation	0.5	0.5	0.4	0.4
Rotation around AP axes	1.9	1.8	1.9	2.0
Rotation around axial axes	1.3	1.3	1.3	1.3
Rotation around lateral axes	3.7	3.6	3.5	3.6
Mix of rotations and translations	2.2	2.2	2.2	2.2

premotion to Navigator) and invert the second transform (to go from Navigator to postmotion). The rotation R between the pre- and postmotion datasets can be calculated as $R = R_2^{-1} R_1$ and the translation is given by $T = R_2^{-1}(T_1 - T_2)$. The center of rotation in these operations was considered the mid-point of datasets but the transformation was easily converted to rotation with respect the MRI isocenter knowing the isocenter offset in mm from the center slice of the Navigator datasets.

Besides comparing the agreement in the individual 6-DOF parameters of the heart as estimated by the VTS and MRI we also performed an overall quantitative assessment of agreement between the VTS and MRI estimated motions for the voxels of the segmented heart volume as a whole. For this we determined the Average Error as the mean of the differences in the vector displacements between the 6-DOF MRI and VTS estimated motions over all voxels in the heart region. These Average Errors were further averaged over all volunteers and repetitions of a given type of motion. This provided an overall measure of the average agreement in the motion of the heart as estimated by VTS and MRI for each of the types of motion investigated. We report the number of measurements, and mean, standard deviation, and range for the Average Errors for each type of motion in the volunteers. For the volunteers we compared the Average Errors for each type of motion to that of the No-Motion studies using a two-tailed unpaired t-test with unequal SD and Bonferroni correction for multiple comparisons.⁵⁸ We employed a $p < 0.05$ as an indication of a significance difference.

3. RESULTS

3.A. Anthropomorphic phantom studies

As shown in Table I, there was very little change in the accuracy as measured by the Average Error of MRI estimation

of motion of the heart in comparison to the VTS estimate for gaps from 0 (no gap) to 27 mm for all six motions investigated. Note only gaps of 0, 9, 18, and 27 mm (i.e., gaps of none, 3, 6, and 9 slices) are shown in Table I because of the small extent of change. We selected the usage of 9 mm gaps with the human volunteers based on usage of this gap decreasing the time the volunteers would need to hold their breath by a factor of approximately four, and causing no discernible loss in estimation accuracy for a truly rigid-body. Also as the thorax in humans is not rigid, we preferred to error on the side of conservatism and selected a 9 mm gap as opposed to a larger one.

As can be seen in Table I the agreement (Average Error) between MRI and VTS is excellent when No-Motion occurs (0.1 mm). For the other five motions whose results are presented in that table, we report in the text major translations (those > 1 mm) and rotations (those $> 1^\circ$) as estimated by the VTS. We also provide in Table II all motion components for both the VTS and MRI for the study in which we observed the largest Average Error. For the axial-translation of 20.3 mm the Average Error is still small (0.5 mm). The next motion investigated was a rotation of -5.3° around the AP-axes, which also inadvertently included an axial-translation of 11.0 mm, and resulted in a larger Average Error of 1.8 mm. An Average Error of 1.3 mm was determined for the next motion which was a rotation of -6.8° about the axial-axis with a lateral-translation of 16.1 mm. The largest Average Error observed was 3.6 mm for rotation about the lateral-axis of -5.0° which also included an axial-translation of 24.4 mm and a vertical translation of -5.9 mm. Table II reports these results in more detail. In our previous studies^{59,60} it was determined that estimation of the angle of rotation about the lateral-axis was difficult due to our usage of markers positioned in a narrow “line” laterally on the chest. Thus for both the phantom and volunteer investigations reported herein we increased the

TABLE II. VTS versus MRI 6-DOF estimates for rotation of Data Spectrum Anthropomorphic Phantom about its lateral axis coupled with an axial translation for case of 9 mm gap between MRI slices as employed for human volunteers.

Estimated by	AP translation (mm)	Axial translation (mm)	Lateral translation (mm)	AP rotation (deg)	Axial rotation (deg)	Lateral rotation (deg)	Distance moved (mm)	Average error (mm)
VTS	-5.9	24.4	0.0	-0.6	-0.2	-5.0	25.1	
MRI	-5.6	23.0	-0.2	0.4	-0.4	-4.3	23.7	
Difference	-0.3	1.3	0.2	-0.9	0.3	-0.8	1.4	3.6

FIG. 5. Shown are sagittal MRI slices through the heart region of the Data Spectrum Anthropomorphic Phantom which underwent a combination of rotation primarily about the lateral axis and translation primarily along the axial axis (see Table II for details). (Top) Starting from the left are the premotion slice, the “Navigator” slice at the position of the premotion slice, the corresponding “Navigator” slice after 6-DOF registration of the “Navigator” slices to match the premotion slices, the difference between the premotion and the “Navigator” slice, and the difference between the premotion slice and “Navigator” slice after 6-DOF registration. (Bottom) Shown is the same sequence of slices for the postmotion slice matching the location of the premotion slice of the top row.

width of the region axially over which the chest markers are positioned, as detailed in Sec. 2 and illustrated at the left in Fig. 2. The difference between the estimated rotations about the lateral-axis for this acquisition estimated by the VTS and MRI is now less than 1° as shown in Table II. Notice that individually the disagreements between the parameter estimates are small; however, the combination resulted in this study having the largest Average Error among our phantom studies. Figure 5 illustrates a comparison of one set of corresponding sagittal slices from this study. Easily seen in the figure is the bright oil filled myocardial wall of the heart phantom. Also seen are the signal absent plastic base of the heart phantom and contour of the liver compartment. Notice the improved matching of the inferior wall of the heart relative to the dashed horizontal line with registration. Also notice the obvious large axial-shift of the phantom from its premotion position illustrated by the difference in location of the inferior wall of the heart relative to the dashed horizontal line as well as the rotation of the superior end of the phantom anteriorly by insertion of support padding underneath it. The success of the registration process can be judged by looking at the difference between the pre- and postmotion slices in comparison to the Navigator slice with and without being moved to match the pre- or postmotion slice using the registration estimate. The final entry in Table I was for a rotation about the AP-axis of 6.6° and an axial-rotation of -2.6° with a 24.2 mm lateral-

shift and a 4.3 mm axial-shift. The Average Error is less than that in the previous study being approximately 2.2 mm.

3.B. Human volunteer studies

Table III summarizes our measurements in the motion studies of the five healthy volunteers (two males and three females) who participated in this study. It lists for each of the six motion types investigated the number of studies included in our analysis, and then the range, average and standard deviation in the VTS estimated motion, MRI estimated motion, and the Average Error between these two estimates of motion. In addition an indication of when there is a statistically significant difference at the level of $p < 0.05$ in comparison to the results of when the volunteers were requested to remain stationary is provided. The measured motions are listed to provide a gauge for the relative size of the Average Errors in comparison to the magnitude of the motions which occurred. The listed VTS and MRI estimated motions are the norm of the translations in 3D. It does not include any differences in rotation which would also contribute to misalignment. The Average Error is provided to allow an overall assessment of the agreement in the motion of the voxels of the heart by the two methods including both amount and direction of movement.

TABLE III. Summary of motion estimates by VTS and MRI of the heart in the breath-hold human volunteer studies.

Type of motion	Number of studies	VTS estimated motion (mm)				MRI estimated motion (mm)				Average error (mm)			
		Min	Max	Ave	SD	Min	Max	Ave	SD	Min	Max	Ave	SD
No-Motion	7	0.1	1.9	0.7	0.6	0.2	1.00	0.7	0.3	0.2	2.4	1.0	0.8
Translation	7	7.6	21.6	13.6	5.5	6.1	22.8	14.5	6.5	1.0	3.9	2.6 ^a	1.0
Bend/twist	12	3.7	17.0	10.2	4.3	5.0	15.0	9.7	3.0	2.3	7.9	4.9 ^a	1.7
Arm	11	0.4	2.9	1.1	0.8	0.2	2.6	1.1	0.7	0.5	3.4	1.5	0.9
Shoulder	7	1.6	6.5	4.5	1.9	1.5	5.1	3.4	1.5	2.5	5.6	4.3 ^a	1.0
Leg	8	0.6	2.1	1.3	0.4	0.8	3.7	1.6	0.9	1.1	4.6	2.2	1.1

^aSignificant difference in the Average Error at $p > 0.05$ in two-tailed unpaired t-test with unequal SD and Bonferroni correction for multiple comparisons.

FIG. 6. Plots of the vertical location of one abdominal marker as determined by the VTS versus time post the start of acquisition in 1/30 of second increments. (Top) No-Motion study. (Bottom) Motion of leg between MRI acquisitions. Note the difference in scale of the vertical locations between the plots.

In the case of the No-Motion studies for the volunteers the estimated motions were small but larger than for the anthropomorphic phantom, as expected. The Average Error averaged over all volunteer No-Motion acquisitions was 1.0 mm, and the maximum Average Error among the volunteer acquisitions was 2.4 mm. These numbers provide a baseline for the differences seen due to inherent errors in both the MRI and VTS measurement methods, and the heart not necessarily being perfectly stationary during breath-hold in humans. This lat-

ter point is illustrated in the upper plot in Fig. 6 which shows the gradual downward drift in marker position of ~ 1.5 mm of an abdominal marker during the 27 s duration of the breath-hold acquisition for a volunteer asked to perform a No-Motion study. Also note the high-frequency low magnitude variation in marker height due to the pulsatile blood-flow underneath the marker. This gradual downward drift in marker location is typical of our baseline studies and corresponds to the gradual loss in respiratory volume previously reported in breath-hold MRI studies.⁶¹

For the case of translational motions which might be expected to be RBM, the Average Error averaged over all studies was 2.6 mm, and the maximum Average Error among the volunteer acquisitions was 3.9 mm. Despite being on average small there was a statistically significant difference with the Average Errors of the No-Motion studies. However, the Average Errors are small compared to the corresponding motion estimates from the VTS and MRI. Thus one could expect the VTS estimate would correct the majority of the motion of the heart for this type of motion. Also note that the Average Errors are less than the typical pixel dimension of 6.4 mm recommended by the American Society of Nuclear Cardiology Guidelines for cardiac SPECT.⁶² The parameters for the 6-DOF motion as estimated by VTS and MRI for an example axial translation are listed as the first motion type in Table IV, and a comparison of sagittal slices from this translation studies is shown in Fig. 7. As can be seen from these, the motion is indeed mostly axial, and the agreement between VTS and MRI estimated motions is fairly good.

With the bending or twisting of the volunteer to the right or left, Table III shows the amount of volunteer motion was on average less than that seen with axial or lateral translation. Also the agreement between the VTS and MRI estimated motions was not as good as with translational motion, and was statistically significantly different from the No-Motion studies. Still the VTS reduced the alignment error on average by more than a factor of 2. A detailed analysis of the differences in the 6-DOF motion parameters for one of the studies where the volunteer was requested to perform a twist is presented as the second study in Table IV. A comparison of sagittal slices from this study is shown in Fig. 8. A difference particularly in the angles determined by the two motion estimation methods

TABLE IV. VTS versus MRI 6-DOF estimates for a human volunteer asked to perform the indicated motion between the acquisitions of the two sets of MRI slices.

Motion type	Estimated by	AP translation (mm)	Axial translation (mm)	Lateral translation (mm)	AP rotation (deg)	Axial rotation (deg)	Lateral rotation (deg)	Distance moved (mm)	Average error (mm)
Translation	VTS	0.5	14.1	-0.4	-0.8	0.3	-0.3	14.1	
	MRI	-0.2	15.7	-0.7	-0.1	1.00	-1.0	15.7	
	Difference	0.7	-1.6	0.4	-0.7	-0.7	0.7	-1.6	1.9
Twist	VTS	-7.9	-0.5	15.0	3.2	-3.6	3.4	17.0	
	MRI	-6.8	-5.0	12.2	5.2	-8.9	5.7	14.8	
	Difference	-1.1	4.4	2.8	-2.0	5.5	-2.3	2.1	6.4
Shoulder	VTS	-4.5	-4.6	0.3	-0.7	-1.3	-0.7	6.5	
	MRI	-2.0	-2.5	-1.5	0.5	-3.0	3.9	3.6	
	Difference	-2.5	-2.1	1.7	-1.3	1.6	-4.6	2.9	5.0

FIG. 7. Sagittal MRI slices through the heart region of a volunteer asked to perform an axial translation towards their feet between the pre- and postmotion acquisitions. The ordering of the slices within the two rows is as given in Fig. 5. The change in the location of the heart when the volunteer performed the axial translation can be noted by the location of the heart in the pre- and postmotion slices in comparison to the dashed line which in each row is drawn at the location of the inferior level of the heart in the Navigator slice second from the left.

can be seen such that there is only about a 2 mm difference in the magnitudes of the estimated motions, but approximately a 6.4 mm average difference in the motion of the voxels of the heart volume. This is typical of the other volunteer studies with these types of motion. Part of this difference may be due to the limitations in estimating combined angles and translations even for a truly RBM as illustrated by our study with the anthropomorphic phantom in Table II. It likely also is due to nonrigidity between the chest markers and heart within the chest. Thus, we hypothesize that combining a data-driven method such as that of Kyme *et al.*²⁸ or Schumacher *et al.*³¹ with our VTS when there is the indication of other than mostly translational motion, may further improve the robustness of correction. The VTS could be used to divide the projections into groups with similar motion states, provide an initial guess as to the motion, and perhaps constrain motion estimation.

From discussions with our imaging technologists, the most common types of arm motion during Cardiac SPECT imaging are those which do not involve the shoulder significantly. As summarized in Table III, when volunteers performed this type of motion the average Average Error and maximum Average Error were 1.5 and 3.4 mm, respectively. These are close to

and not statistically different from the No-Motion studies in agreement with the observation that little motion was detected by either the VTS or MRI.

With the less common case of arm motion involving significant motion of the shoulders the errors increased. It can be seen from Table III that the Average Errors increased to 4.3 mm when averaged over all such studies, and 5.6 mm was the largest Average Error. These are definitely larger than the No-Motion Average Errors and were statistically different from them. In looking at the extents of motion, it can be noted that marker motion shows an indication of being larger than the motion of the heart, and in five of the seven studies this was true. The motion parameters and differences for one of these studies are listed as the third motion example in Table IV. In this case it can be seen that even though the magnitude of estimated motion is less for MRI than the VTS, the largest error is in the extent of lateral rotation between the two. From these results there is the suggestion that the VTS may have an issue accurately correcting for the small amount of heart motion when a patient moves their shoulders. On the basis of this result we plan to look for a pattern of movement in the chest markers such as axial movement towards or away from

FIG. 8. Sagittal MRI slices through the heart region of a volunteer asked to perform a twist of their upper body between the pre- and postmotion acquisitions. The ordering of the slices within the two rows is as given in Fig. 5. Note the marker seen anterior (left) to mid-level of the heart in the postmotion slice. The presence of some axial translation during the twisting motion can be appreciated by the location of the apex of the heart in the postmotion slice extending below the dashed line.

the head which is not duplicated on the other side to see if we can isolate this type of motion from impacting our correction of cardiac motion.

Patient leg motion is also quite common during the protracted imaging time of SPECT. By the results given in Table III only a small amount of motion, similar to that of when solely the arms were moved, was observed by the VTS or in MRI with leg motion. The Average Errors were also small, but larger than the No-Motion studies. The observed Average Errors were not statistically different between these two motion types. The impact of leg motion is however much larger on the abdominal markers during the period of motion, as can be seen in Fig. 6. The lower plot in this figure shows a change of >1 cm in vertical position while the volunteer moved their leg by likely involving usage of their abdominal muscles. The corresponding change was minimal for the chest markers. Thus leg motion is seen to have only a small impact on the estimation of body-motion from the chest markers, but could corrupt the estimates of the amplitude of respiratory motion when the vertical motion of abdominal markers is employed in the correction of respiratory motion.^{51,63–67}

4. DISCUSSION

In past work we have determined that the tracking of patient motion during clinical imaging can be robust with little impact on workflow, and can make considerable improvements in the appearance of reconstructed slices when employed in motion correction.⁴⁸ The goal of these investigations was to determine how well the 6-DOF rigid-body motion (RBM) estimated from the markers on the surface of the chest predicts internally the RBM component of heart motion, so that it can be used to correct such motion in SPECT and PET cardiac perfusion imaging. Thus we investigated solely this motion model when registering the heart in the pre- and postmotion MRI slices. This allowed us to compare the RBM which best accounted for the heart motion between the two motion states internally with that estimated externally. That the estimated RBM accounted for the major portion of the heart motion in MRI was visually validated by comparing the difference images between the sagittal MRI slices of each motion state with the baseline “Navigator” slices postregistration. Example comparisons are given in Figs. 7 and 8 where the difference images have only minor differences in the heart region with the “Navigator” slices once the pre- and postmotion slices have been registered to them. Note that since our MRI registration was of the heart regions, it is not surprising that significant differences are seen outside of the heart region especially in Fig. 8 where the volunteer was asked to perform an upper body twist.

The error in tracking the motion of an individual marker is quite small. We regularly observe the SD in marker location to be <0.2 mm. Note for example the low noise level in the plot of the vertical position of an abdominal marker shown at the top in Fig. 6 allows one to see the ~ 0.2 mm pulsatile motion of the blood under the marker. In a previous publication⁴⁸ we found excellent agreement between changes in position of a phantom with seven markers with Tc-99 m inside as estimated

by Vicon tracking and the center-of-mass of counts in SPECT slices. With respiratory motion we have investigated the pairwise correlation of the five individual markers on the chest with each other and the two markers tracking the abdomen with each other for nine volunteers.⁶⁸ The average (SD) in Spearman ranked correlation coefficients in these values were 0.97 (0.02) for the chest and 0.98 (0.01) for the abdomen. This demonstrates the consistency in marker tracking for that form of motion. Thus although likely a part of the disagreements between VTS and MRI estimates of motion we do not believe VTS accuracy played the dominate role in the disagreements, when large.

To control other sources of error we put considerable effort over a course of years to perfect our MRI acquisition and processing. This enabled us to achieve the excellent agreement between VTS and MRI for the rigid motions of the phantom reported in Table I, and the baseline No-Motion studies in human volunteers reported in Table III. Still there are limitations to our methodology which may have impacted the agreement between the VTS and MRI in human volunteers beyond the possibly nonrigid nature of motion between the markers and heart. The first is that the volunteers may not have held their breath throughout the two sequential MRI acquisitions. To minimize the possibility of this, we trained the volunteers in the motions to be performed while holding their breath prior to MRI acquisition for each type of motion. Further, by adopting the motion estimation strategy of King *et al.*⁵⁷ we significantly reduced the period of breath-holding from our previous studies.^{59,60} We also discarded any study in which the volunteer self reported that they suspected that they were unable to hold their breath. Finally we checked the vertical motion of abdominal markers to look for evidence of the volunteer breathing, and discarded those where we saw such evidence. The normal pattern of motion we looked for with successful breath-holding was a gradual change in vertical position with time, except when the volunteer moved in between the pre- and postmotion MRI acquisitions. An example of this is shown in the lower plot of Fig. 6. Any significant deviation from this resulted in the study being discarded as suspicious. Still it is possible that the volunteer breathed in or out at the time of motion. When this happened we would not be able to separate this from the motion of the abdominal marker during the requested motion of the volunteer.

The second limitation is that it is possible for the stretchy band which was wrapped tightly around the volunteer’s body could have slipped when the volunteer moved in response to our instructions. Slippage of markers is always a possible cause of error in motion estimation employing them; however, it is exacerbated in this study by our requesting the volunteers to move rapidly between the pre- and postmotion MRI acquisitions while holding their breath. Thus this possible cause of differences between chest marker and heart motion should be born in mind when assessing the agreements shown in Table III, and when correcting clinical studies for motion. Figure 9 shows an example of one of the three studies in which we found definite evidence of marker slippage and excluded from the motions summarized in Table III. In the example shown in this figure the VTS reported an 8.7 mm axial

FIG. 9. Illustration of marker slippage on the body the between (left) premotion and (right) postmotion MRI acquisitions of a volunteer asked to perform an axial translation during breath-hold.

translation towards the feet; whereas, the MRI estimate was a 21.3 mm translation. The upper horizontal dashed line in this figure is drawn at the caudal most level of the heart in the premotion MRI slice. The lower horizontal dashed line was drawn at the caudal most level of the heart in the postmotion MRI slice. Marker slippage can be seen by noting the marker position in the two slices relative to the horizontal lines. In the premotion slice the marker is mostly below the upper dashed line, and it is above the dashed line in the postmotion slice. Based on this observation we are considering investigating ways to decrease the likelihood of marker slippage when employed to track motion clinically.^{43,69} In some cases one can even forego the markers and estimate the motion directly by stereoisimaging of a projected structured light pattern on the external surface of the patient.⁷⁰ We removed these three obvious cases; however, it is possible that some nonobvious slippage was not found by us and is in the motion results summarized in Table III.

A third limitation is that neither estimating RBM from a group of markers which are not attached to a rigid surface nor our method of estimating cardiac motion within the chest of human volunteers through slice registration is without the possibility of inaccuracies. Still by looking at the results in tables for the phantoms and volunteers, it can be seen that we have reached fairly good agreement, especially in the cases where agreement might be expected.

5. CONCLUSIONS

In this study we investigated the hypothesis that in the absence of respiration the RBM model could be used to approximately estimate the gross body-motion of the heart from that of external markers on the patient's chest. We did this by comparing the motion of the heart as estimated in MRI slices with VTS estimated motion of chest markers for a variety of motions emulating those seen clinically in cardiac MPI SPECT imaging. We determined that with motion of the arms and legs the agreement between the externally and internally estimated movements of the heart (Average Error) is not significantly different than when the volunteers tried to remain stationary. With volunteer translational gross body motion there was a small (average 2.6 mm) but statistically significant difference

between the estimated motions (Average Errors). For motions that could be expected to be non-RBM in nature such as torso bends and twists, the VTS estimate used with the RBM model still decreased the error compared to no correction but to a lesser extent. This is due to the differences between the estimates (Average Errors) being larger (average 4.9 mm) as well as statistically significant. When movement of the shoulders became involved with arm movement there also was a reduction of the agreement between exterior prediction and interior motion, with the difference again being statistically different. Thus overall the use of VTS based motion correction is anticipated to reduce the presence of motion artifacts in cardiac SPECT imaging. However, further work is needed to improve agreement in the case of shoulder movement. Also needed are improved methods of affixing the markers to the volunteers which further diminish issues with slippage while maintaining the benefits of ease of use, being inexpensive, and being disposable so as to avoid issues with contamination. Investigation of motion models beyond that of simple 6-DOF RBM such as finite-element modeling could also improve results, especially when combined with patient specific anatomical information.⁷¹

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