Published in final edited form as:

Magn Reson Med. 2011 November ; 66(5): 1286–1292. doi:10.1002/mrm.22909.

CINE Turbo Spin Echo Imaging

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Abstract

While high resolution TSE images have demonstrated important details of carotid artery morphology, it is evident that pulsatile blood and wall motion related to the cardiac cycle are still significant sources of image degradation. Although ECG gating can reduce artifacts due to cardiac induced pulsations, gating is rarely used because it lengthens the acquisition time and can cause image degradation due to non-constant TR. This work introduces a relatively simple method of converting a conventional TSE acquisition into a retrospectively ECG correlated cineTSE sequence. The cineTSE sequence generates a full sequence of ECG correlated images at each slice location throughout the cardiac cycle in the same scan time that is conventionally used by standard TSE sequences to produce a single image at each slice location. The cineTSE images exhibit reduced pulsatile artifacts associated with a gated sequence but without the increased scan time or associated non-constant TR effects.

Keywords

TSE sequences; motion artifact suppression; carotid imaging

Introduction

Accurate identification of carotid atherosclerotic plaque composition can better identify those individuals at higher risk for subsequent cerebral embolization and neurologic symptoms (1-6). MRI presents an advance over x-ray techniques, including Computed Tomography, because it produces images with a variety of different tissue contrasts. Because MRI can better discriminate soft tissue types, it offers improved discrimination of atherosclerotic plaque components such as hemorrhage, disrupted fibrous cap, lipid/necrotic core, calcification and ulceration. TSE (Turbo Spin Echo or Fast Spin Echo) pulse sequences, with efficient acquisition of T1, T2 and proton density weighted images, have become useful in the evaluation of cervical carotid artery disease (7-9). High resolution 3D TSE techniques would be valuable in carotid MRI, but motion artifacts that arise due to the relatively long data-acquisition time result in an unacceptable fraction of non-interpretable studies. As a result, nearly all clinical cervical carotid MRI studies are performed with 2D TSE techniques, with double inversion blood suppression, where motion artifacts corrupt a smaller fraction of studies and motion in one slice does not cause artifacts in an adjacent slice. However, in many subjects, even 2D TSE techniques result in obvious blurring or ghosting artifacts that greatly reduce the quality of the images.

Motion artifacts due to swallowing and other neck muscle movements can be reduced by applying parallel imaging techniques to reduce the acquisition time (10-12). Reduced field of view sequences with more efficient double inversion blood suppression have also shown reduced susceptibility to these types of motion artifacts (13,14). Single shot sequences, such as Half Fourier Acquisition Single Shot Turbo Spin Echo, can also be used to reduce acquisition time and hence the susceptibility to certain motion artifacts (15). However, single shot techniques tend to have image blurring due to T2 decay. Navigators and dedicated motion detection RF coils have been shown to be successful in detecting and rejecting data corrupted by swallowing (16,17). Using average specific phase encoding ordering or comparing adjacent sets of k-space data lines are alternative methods for detecting data corrupted by patient motion (18-20).

Motion artifacts that are correlated to the cardiac cycle can be greatly reduced by synchronizing data acquisition with the heart cycle using electrocardiogram (ECG) gating. Cine sequences based on gradient echo or true fast imaging with steady state procession have shown the dynamic nature of the carotid artery and cyclic blood flow (21,22). Gradient echo based pulse sequences typically have a very short repetition time ($TR < 30\text{ms}$) and can therefore be efficiently designed as cine sequences to obtain complete image measurements at multiple cardiac phases in the same acquisition. However, because of the relatively long TR of TSE techniques ($> 500\text{ms}$), ECG gated acquisitions typically only obtain one image at a single cardiac phase. For example, ECG gated acquisitions have been used reduce the effect of cardiac motion by acquiring data during diastole (23-25). Reduced field of view GRASE with cardiac gating has been shown to generate reasonable images of carotid plaque (26). However, ECG gating can result in image degradation due to non-contrast TR times (27). This problem can be improved by limiting gating to just the central portion of k-space, however, the effect of a variable TR is still noticeable (28).

This work introduces a novel and relatively simple method of converting a conventional TSE acquisition into a retrospectively ECG correlated cineTSE sequence. Because the cineTSE images are ECG correlated they exhibit the reduced pulsatile artifacts obtained with gated sequences. Additionally, cineTSE generates a full sequence of ECG correlated images at each slice location throughout the cardiac cycle in the same scan time that is conventionally used by standard TSE sequences to produce a single image at each slice location. Given that the data is acquired with the same timing as a standard TSE sequence, there are no non-constant TR effects such as are typically associated with gated sequences. In this work, the basic theory of the cineTSE method is presented and supported with data from several patients with and without carotid disease.

Theory

Information about the patient's cardiac cycle is recorded during the MR scan utilizing ECG information or a pulse oximeter. Because the pulse oximeter is less susceptible to gradient induced noise and requires less patient preparation, it has been utilized exclusively in this work. Hence the term ECG correlation is applied generally to correlating MRI data to the patient's cardiac cycle, regardless of the method used to monitor the cardiac cycle. The acquired k-space data lines are sorted into N_t temporal bins according to the time elapsed since the last systolic trigger. The result is N_t undersampled data bins, each representing a different phase of the cardiac cycle. The undersampled images are reconstructed by simultaneously considering information encoded by the coil sensitivities as well as applying a temporal constraint. This is accomplished by minimizing the following objective function:

$$G = \|W(k_x, k_y, t) \cdot \mathfrak{F}_{xy} [s_n(x, y) \cdot m(x, y, t)] - d_n(k_x, k_y, t)\|_2^2 + \|\lambda(x, y) \cdot \nabla_t [m(x, y, t)]\|_2^2 \quad [1]$$

where $m(x, y, t)$ is the current image estimate, t is the time (bin number) in the cardiac cycle, $W(k_x, k_y, t)$ is a weight function specifying which lines in k -space have been acquired in each bin, $\mathfrak{F}_{xy}()$ is the 2D Fourier transform along the spatial coordinates, $s_n(x, y)$ is the coil sensitivity of the n^{th} coil, $d_n(k_x, k_y, t)$ is the actual data acquired from the n^{th} coil in a given bin, $\lambda(x, y)$ specifies the level of temporal constraint and $\nabla_t()$ is a temporal gradient.

Since $m(x, y, t)$ is periodic in time, we have found the following definition of $\nabla_t()$ to be optimal compared to methods that estimate the temporal gradient using only a few neighboring points in time:

$$\nabla_t [m(x, y, t)] = \mathfrak{F}_t^{-1} \left\{ \mathfrak{F}_t [m(x, y, t)] \cdot \frac{2\pi i f}{N_t} \right\}. \quad [2]$$

where $M(x, y, f)$ is the Fourier transform in time of $m(x, y, t)$. With the temporal gradient definition in Eq. [2], the objective function in Eq. [1] is minimized using the following gradient term:

$$\nabla_m G = -2\lambda^2 \nabla_t^2(m) + 2 \sum_n s_n^* \cdot \mathfrak{F}_{xy}^{-1} [W \cdot \mathfrak{F}_{xy}(s_n m) - d_n]. \quad [3]$$

The images are then reconstructed using a non-linear conjugate gradient descent algorithm (29,30).

Method

Data Acquisition

All studies were performed on a Siemens Trio 3T MRI scanner with custom designed 4 or 16 element phased array surface coils and a modified TSE sequence (31,32). All human studies were approved by the institutional review board and informed consent was obtained from all subjects.

Two modifications have been made to the standard TSE pulse sequence. First, the standard sequence was modified to record the patient's cardiac cycle utilizing a pulse oximeter. Since the heart rate of most patients is not constant over several minutes of scanning, the filling of temporal bins is random in nature (as desired). For patients with very little variability in the heart rate, the randomness of bin filling can be increased by selecting a TR that it is not a multiple of the average heart rate. The second change to the standard TSE pulse sequence was to utilize variable density sampling to help ensure that each temporal data bin contains a sufficient number of lines near the center of k -space (where most of the image energy is found). The sampling density (phase encode direction) of the standard sequence was modified to more frequently sample the center of k -space. To prevent an increase in scan time, this is accomplished at the cost of missing varying lines near the edge of k -space in each average. Every line of k -space is sampled at least once during the entire acquisition. Figure 1 shows the sampling pattern and density resulting from the modified TSE sequence where the equivalent of two averages is acquired. Because any repeated k -space lines are averaged before reconstruction, this variable density sampling does not affect image

blurring, but it does introduce a slight increase in image noise correlation that is not readily observed in the final reconstructed images. Figures 2b and 2c compare a standard TSE image (2 averages) and a variable density TSE image (equivalent of 2 averages).

Data used to generate images and plots in Figs. 2, 4 & 5 was acquired with a T2 weighted cineTSE sequence with the 4 element coil, 2 averages, a resolution of $0.6\text{mm} \times 0.6\text{mm} \times 2\text{mm}$, 12 echoes per train and a TR/TE of 2.5s/61ms. The data used in Fig. 3 was acquired with a T1 weighted cineTSE sequence with the 16 element coil, 2 averages, a resolution of $0.5\text{mm} \times 0.5\text{mm} \times 2\text{mm}$, 12 echoes per train and a TR/TE of 650ms/8ms. 12 slice locations were imaged for both T1 (4 concatenations of 3 slices) and T2 weighted cineTSE acquisitions. The echo train spacing is 7.6 ms with a readout bandwidth of 64kHz. Typical scan time for the T1 or T2 weighted cineTSE sequences is about 2 minutes. Including the data presented in this work, a total of 19 carotid patients, 4 patients with brain aneurysms and 4 volunteers have been imaged with the proposed cineTSE sequence.

Image Reconstruction

In each experiment, only one data set is acquired and used to reconstruct both the variable density TSE and the cineTSE images. In this sense, a direct comparison can be made between the two methods. The variable density TSE images are reconstructed by simply averaging the oversampled data in k-space and utilizing a Fourier transform. For the cineTSE images, we are only reconstructing magnitude images in this study and it is sufficient to estimate the coil sensitivities, $s_n(x,y)$, as:

$$s_n(x,y) = \frac{m_n(x,y)}{\sqrt{\varepsilon + \sum_n m_n(x,y) m_n^*(x,y)}} \quad [4]$$

where ε is a small constant and $m_n(x,y)$ is the image estimate from the n^{th} coil. In cases where phase information from the final images is required, a more specific measure of the coil sensitivities is required.

The strength of the temporal constraint, $\lambda(x,y)$, is a tradeoff between temporal blurring, artifact reduction and image noise. If we highly constrain all the pixels, each image from the cineTSE reconstruction will converge to the standard TSE image. The result will be images with little artifact reduction, a SNR similar to the averaged standard TSE image and limited dynamic information. This would circumvent the main purpose of the cineTSE sequence which is to obtain dynamic information. If no temporal constraint is applied, we regain the dynamic information but increase the amount of undersampling artifact in the cineTSE images. The SNR of the unconstrained cineTSE images would be similar to a standard TSE image acquired with a single average. Optimizing temporal blurring, artifact reduction and image noise is best accomplished with a spatially varying temporal constraint. Since we are correlating our data to the cardiac cycle, it is consistent to constrain pixels varying with frequencies unrelated to the heart rate more strongly than those varying at frequencies near the heart rate. The spatially varying temporal constraint, $\lambda(x,y)$, is constructed by observing the temporal spectrum of the initial image estimates as:

$$\lambda(x,y) = \sum_f \Im_t [m(x,y,t)] \cdot \alpha(f) \quad [5]$$

where $\alpha(f)$ is a bandpass filter selecting the frequency components occurring near the patients average heart rate and $m(x,y,t)$ is the initial image estimate created using bilinear interpolation in the temporal dimension. The acquired data, $d_n(k_x, k_y, t)$, is scaled so that our final image, $m(x,y,t)$, has a magnitude range of approximately 0~1.0. In this study, $\lambda(x,y)$ ranges from zero (no temporal constraint for pixels with variations strongly correlated to the patients heart rate) to 0.3 (high temporal constraint for pixels with variations that are uncorrelated to the cardiac cycle). In addition, we utilize a temporal constraint that varies smoothly with spatial position by convolving the result of Eq. [5] with a Gaussian (FWHM of ~8 pixels). We note that while a more optimum spatially varying temporal constraint may be found, the proposed version has shown success to date.

Since missing data lines are calculated using coil sensitivities and a temporal constraint, there is a practical limit on the number of temporal data bins the data should be sorted into. Two few data bins might not allow enough temporal resolution to observe important events in the cardiac cycle. Too many bins will impose a computational burden on the iterative reconstruction without yielding a significant increase in dynamic information. We have found that with the current patient data (2 data averages with heart rates ranging from 51 to 74 beats per minute), sorting data into more than 12 temporal data bins does not yield significant improvements to the results. Therefore, unless specified, all data is sorted into 12 temporal data bins.

The cineTSE sequence acquires data independent of the cardiac cycle and as a result some of the acquired data lines may occur on or near the dividing boundary of data bins in the cardiac cycle. This data may reasonably be sorted into more than one temporal data bin. Such an effect can be taken into account by allowing our data bins to overlap slightly. We have found that an overlap of 30% is enough to help ensure each data bin contains an adequate amount of data but not too much to blur results in the temporal direction.

Finally, due to the random filling of temporal bins, it is possible that a particular line of k-space in a bin is sampled multiple times. Although this does not occur very often, it is handled by averaging the multiply sampled lines and retaining the binary definition of the weight function ($W(k_x, k_y) = 0$ or 1).

Results

Figure 2a displays a standard T2 weighted TSE image reconstructed from a volunteer (4 element coil and 2 data averages). A zoomed version of the carotid is shown in Figs. 2b and 2c for a standard and variable density TSE acquisition respectively. Figure 2d shows six images, among twelve total images, obtained from the same data set as Fig. 2c but reconstructed using the cineTSE technique. The arrows in Figs. 2b through 2d show a flow artifact in the carotid artery just below the bifurcation. The flow artifact is seen to come and go throughout the cardiac cycle and can easily be identified in the cineTSE images (arrows in Fig. 2d). SNR measurements of the lumen (area shown at the bottom right of Fig. 2d) were 8.4 for the standard TSE (Fig. 2b), and 8.6 for the variable density TSE (Fig. 2c) and 5.9 for the cineTSE (Fig. 2d with time $t=0.36s$). CNR measurements (for the vessel wall and lumen areas shown at the bottom right of Fig. 2d) were 8.1 for the standard TSE (Fig. 2b), 7.8 for the variable density TSE (Fig. 2c) and 10.5 for the cineTSE (Fig. 2d with time $t=0.36s$). SNR was defined as the ratio between mean lumen signal (in the area specified) and the standard deviation of the signal measured outside the object. CNR was defined as the difference in mean signals of lumen and vessel wall (in the areas specified) divided by the standard deviation of the signal measured outside the object.

T1 weighted cineTSE images from a patient with an ulceration of plaque in their carotid artery are shown in Fig. 3. Each column contains images from the same three consecutive slices. Figure 3a contains the averaged TSE images while minimum and maximum intensity projections of the cineTSE images are shown in Figs. 3b and 3c respectively. SNR and CNR measurements were made on each slice (regions shown for 3 of 12 slices in Fig. 3d) using the methods described previously and then averaged across all 12 slices. For cineTSE, measurements were made on a single time frame selected during systole. The vessel wall had an average SNR of 18 ± 3 for the averaged TSE images (Fig. 3a) and 17 ± 2 for the cineTSE images (Fig. 3b). The lumen had an average SNR of 6 ± 1 for the averaged TSE images (Fig. 3a) and 3 ± 1 for the cineTSE images (Fig. 3b). These correspond to an average CNR of 12 ± 3 for the averaged TSE images and 14 ± 2 for the cineTSE images.

The motion of a valve in a facial vein is captured with a T2 weighted cineTSE reconstruction as shown in Fig. 4. Figure 4a shows the averaged TSE image with twelve cineTSE images of the valve shown in Fig. 4b (obtained from the same data set as Fig. 4a but reconstructed using a cineTSE technique). The time of each cineTSE image in the cardiac cycle is indicated in Fig. 4b. Where a typical TSE gated sequence returns images at one or two time points, the cineTSE sequence delivers a whole set of images without an increase in scan time.

While the cineTSE technique is designed to eliminate cardiac related motion artifacts, other motion such as swallowing, will still pose a problem. Figure 5a shows that two averages in an averaged TSE image can help to reduce a swallowing artifact. However, this method eliminates any temporal cardiac information we hope to receive since only a single time frame is reconstructed. When the corrupted data is sorted into multiple temporal bins (Fig. 5b) the swallowing artifact returns. The corrupted lines can be detected using data from adjacent data lines or from the same set of lines acquired in different averages (18, 19). These lines can then be removed prior to the cineTSE reconstruction to reduce artifacts due to swallowing (Fig. 5c).

Discussion

Although ECG gating can reduce artifacts due to cardiac induced pulsations, gating is rarely used because it also lengthens the acquisition time, resulting in increased susceptibility to swallowing and respiratory motion artifacts. In addition, many TSE protocols acquire two or more data averages to increase image signal to noise ratio and to help reduce the amount of motion artifact. One of the main advantages of the cineTSE sequence is that it requires no additional scan time over a standard two average TSE sequence. All the examples shown in this work utilize the equivalent of 2 data averages as described previously in the Methods section. Since the cineTSE utilizes a variable density sampling scheme, an averaged TSE image with variable density sampling is used for comparison purposes. Comparing the variable density cineTSE with a standard TSE sequence (uniform sampling in k-space) would require two separate data sets to be acquired. Although this was done for some preliminary work, it was determined that the averaged variable density TSE image was similar enough to the standard TSE image to forego acquiring the additional data (compare Fig. 2b and 2c). While a degree of temporal information could be obtained from just a single average of cineTSE data, we have found that in most cases at least two averages are required to yield consistent and accurate temporal information. In addition, any preparation pulses (such as inversion, suppression or saturation pulses) used with a standard TSE sequence can be employed with the cineTSE sequence.

The volunteer with data shown in Fig. 3 had an average heart rate of 74 ± 4 BPM (beats per minute $\pm$ the standard deviation) which resulted in 12 temporal bins each being filled from

13% to 21%. A total of 15 carotid patients, 4 patients with brain aneurysms and 4 volunteers have been imaged with the proposed cineTSE sequence. The average heart rate ranged from 51 to 74 beats per minute with the standard deviation of heart rate from 2 to 18 beats per minute. Of these data sets, the most uneven case of bin filling was the patient with data shown in Fig. 4 (57 ± 18 BPM with 12 temporal bins filled in the range of 11% to 27%). In this particular case the patient had an arrhythmia, which prevented bins near the end of the cardiac cycle from getting as much data as those near the start of the cycle. In cases like this, utilizing a variable density encoding scheme increases the probability that the central portion of k-space will be adequately sampled in all the temporal data bins.

Both carotid coils used in this study are custom built coils. The 16 channel coil was still being tested and not routinely used clinically at the time of the initial studies (T2w images presented in this paper). All future studies are proposed on the new 16 channel coil. There is no reason to choose one coil over the other based on image contrast (i.e. T1w vs T2w). Since many sites will not have access to a 16 channel carotid coil, showing success with a carotid coil with fewer channels is appropriate. In addition, the cineTSE reconstruction is unchanged between the two coils and either coil could be used for T2w and T1w images.

A major advantage of the cineTSE sequence is the ability to resolve certain flow artifacts and to discriminate between image signal from tissue and recirculation of blood. This is evident in Figs. 2 and 3 where residual blood signal may be misinterpreted for tissue wall. One method of analyzing flow artifacts is to create a set of minimum and maximum intensity projections of the cineTSE images. Pixels with flow artifact will have a large difference between these two projections since the artifact tends to appear and disappear periodically during the cardiac cycle. In the middle slice of Fig. 3, the minimum and maximum intensity projections of the cineTSE images clearly show the ulceration whereas the conventional image shows a lighter signal in the ulceration that might be interpreted as loose necrotic core. Additionally, it has been shown that in standard sequences pulsatile motion of vessels on the order of 1mm peak to peak can occur, causing major blurring of wall morphology (28). Because the cineTSE sequence produces cardiac cycle correlated images, it is possible to scroll through a dozen images at different time frames in the cardiac cycle and select the best image(s) with the least residual blood signal to best depict the vessel wall and plaque morphology.

T1 weighted images provide image contrast that is complementary to that of T2 weighted images and can therefore help discriminate plaque components. However, T1 weighted images have a short TE such that signal from moving blood is not eliminated completely by the spin echo sequence. To eliminate the signal from moving blood, T1 weighted sequences usually employ a double inversion preparation to eliminate the signal from flowing blood. Double inversion techniques are very time inefficient since they prevent the slice interleaving that is used in standard TSE sequences. Even with an efficient variant of double inversion recovery, our 2 minute scan (for the protocols described in the Method section) jumped to just over 6 minutes (14). Since cineTSE is able to distinguish flow artifacts, it appears that T1 weighted imaging could be performed without double inversion or other flow suppression techniques. If blood signal can be detected and eliminated from analysis of T1 weighted cineTSE images, then the cineTSE technique will substantially decrease the scan time typically required for standard T1 weighted TSE sequences. In this case, substantially more information will be obtained in less time. It is also possible that cineTSE coupled with systolic and diastolic pressures will provide an improved mechanism for measuring arterial wall compliance. CineTSE also captures the dynamic motion of cerebrospinal fluid during the cardiac cycle.

An additional highlight of the cineTSE sequence is that the data acquisition is almost identical to a standard TSE sequence. As a result, many of the motion correction techniques (for non-cardiac related motion) developed for standard TSE sequences can be easily applied to the cineTSE sequence (33-39). Because many k-space lines have multiple averages, this data can be used to help detect corrupted data. Additionally, data from adjacent data lines can also be used to detect corruption from patient motion (18,19). Since we are already dealing with undersampled data, it is not a significant problem if a few lines of data need to be discarded due to non-cardiac related motion corruption. As with most sequences, significant amounts of non-cardiac related motion will pose a problem. In Fig. 5, even though 20% of the data was discarded due to corruption, the cineTSE sequence produced images relatively free of cardiac motion artifact.

Conclusion

In this work a method of converting a conventional TSE acquisition into a retrospectively ECG correlated image sequence was demonstrated. Because cineTSE acquisition is asynchronous with the cardiac cycle, it does not have the problems caused by non-constant TR periods of gated sequences. CineTSE generates a full sequence of ECG correlated images at each slice location throughout the cardiac cycle in the same scan time that is conventionally used by a standard (2 average) TSE sequences to produce a single image at each slice location.

Acknowledgments

This work has been supported by NIH grants HL48223 and HL57990, as well as grants from the Cumming Foundation, the Ben B. and Iris M. Margolis Foundation, and the Mark H. Huntsman Endowed Chair.

Grant Support: NIH R01 HL48223, NIH R01 HL57990, Siemens Medical Solutions, the Mark H. Huntsman Endowed Chair, and the Ben B. and Iris M. Margolis Foundation.

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Figure 1.

Typical sampling trajectory and density for the cineTSE sequence used in this study. The trajectory of k-space sampling with each echo train is shown in (a) with the overall sampling density shown in (b). In this case 528 lines were sampled from a k-space spanning 264 lines. This corresponds to the same number of lines that would have been sampled with 2 full averages and a uniform sampling density.

Figure 2.

Flow artifact identification using a T2 weighted cineTSE reconstruction (4 channel receive coil and 2 data averages). The standard TSE image with uniform k-space sampling (2 averages) is shown in (a) with a zoomed version of the carotid shown in (b). A zoomed version of the same location acquired with a variable density TSE (see Fig. 1 for sampling density) is shown in (c). Six equally spaced images in the cardiac cycle (of 12 total images) from the cineTSE reconstruction are shown in (d). The time (in seconds) of each of the 6 images relative to the systolic trigger is indicated by the text on each image. The regions of interest for SNR/CNR measurements of lumen and vessel wall are represented on the bottom right image in (d).

Figure 3.

T1 weighted cineTSE and variable density TSE images of a patient with carotid disease (16 channel receive coil, 2 averages). Each column contains the same three consecutive slices with variable density TSE images shown in (a) and regions of interest for SNR/CNR measurements of lumen and vessel wall shown in (d). To better visualize flow artifacts, minimum and maximum intensity projections of the cineTSE images are shown in (b) and (c) respectively.

(a)

(b)

Figure 4.

Depiction of a valve captured with a T2 weighted cineTSE reconstruction (4 channel receive coil and 2 data averages). The location of the valve is indicated by the arrow on the variable density TSE image in (a). Close up view of the valve for all 12 equally spaced images across the cardiac cycle is shown in (c). The time (in seconds) of each of the 12 images relative to the systolic trigger is indicated by the text on each image.

Figure 5.

Correction of non-cardiac related motion in a T2 weighted cineTSE reconstruction. A standard TSE image with two data averages and volunteer swallowing is shown in (a). A single time frame (of 12 cineTSE cardiac phases) from the motion corrupted data is shown in (b). The same time frame after corrupted data has been detected and removed is shown in (c).