The Problem of Motion in Cardiovascular MRI

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Abstract

Over the past decade Magnetic Resonance Imaging (MRI) has become an increasingly important non-invasive tool in risk assessment and treatment monitoring of cardiovascular disease. However, despite ongoing progress and developments in MR acquisition and reconstruction technology, phisiological motion remains a major problem in many cardiovascular MRI applications. Since MR acquisition is slow compared to physiological motion, the extensive cardiac- and respiratory- induced motion of the heart during the acquisition period can degrade image quality by introducing ghosting and blurring like motion artifacts. Several cardiac and respiratorty motion compensation techniques have been proposed over the last two decades to overcome this problem. These techniques are based on minimizing or correcting the motion during the acquisition. This part of the Cardiovascular MR Imaging: Pushing the Limits Course at ISMRM 2013 will include an overview of some of these methods, discussing their strengths and limitations.

1 Cardiac-induced Motion

In order to minimize cardiac motion data have to be acquired within a time frame (acquisition window) which is short in comparison to the cardiac cycle length. In the majority of cardiovascular MR acquisitions data from the same cardiac phase but different cardiac cycles are acquired during a short acquisition window and then combined under the (strong) assumption that the cardiac motion is the same between heart-beats (segmented acquisition). Image data are acquired either in a single phase of the cardiac cycle using cardiac triggering or in multiple phases using cardiac gating. Cardiac triggering is mainly used for high-resolution anatomical images, as coronary MR angiography and late-gadolinium enhancement. With this approach data acquisition is carried out within a certain acquisition window during a relatively quiescent period in the cardiac cycle (i.e. end-systole or mid-diastole). Cardiac gating (prospectively or retrospectively) is mainly used for CINE MRI and with this approach data is acquired throughout the entire cardiac cycle. Both approaches require information about the cardiac cycles for cardiac synchronization.

Information about cardiac cycles is commonly obtained from external electrocardiogram (ECG) devices. The ECG signal represents the electrical activity of the heart muscle which correlates with cardiac contraction and relaxation. The R-peak of the ECG signal is detected and used as reference point during MRI acquisition. However, a reliable signal for cardiac synchronization is not always available with this approach since the quality of ECG can be strongly impaired by several sources of noise (gradient and RF fields) and the magneto-hydrodynamic

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effect (especially of high and ultrahigh fields). Furthermore, in some applications such as fetal imaging a non-invasive ECG device is not applicable [1].

Cardiac synchronization may be also achieved by using acoustic triggering techniques that exploit heart sounds created by the opening and closing of the heart valves [2] or pulse oximeters [3]. However, both approaches require dedicated devices and additional hardware and may suffer from limited accuracy.

Cardiac self-gating approaches have been proposed to overcome some of these problems by estimating the cardiac-induced motion from the acquired data itself. In these methods usually the central k-space line is acquired repeatedly during the imaging sequence to derive the ECG-like signal [4, 5, 6, 7, 8]. Most of these techniques rely on the change of the overall signal intensity of the entire field of view mainly due to increases and decreases of the blood volume during the cardiac cycle. A limitation of these approaches is that they cannot accurately distinguish between changes of the blood pool due to cardiac motion and changes due to other effects, which also lead to signal variations, such as inflow effects. Recently an image-based cardiac navigator signal derived from the motion of the left ventricle during the cardiac cycle has been proposed [9]. This approach uses 2D real-time images obtained with a golden angle radial sampling scheme to obtain a cardiac gating signal directly from physiological processes of the heart.

2 Respiratory-induced Motion

Breathing leads to a shift and deformation of the heart mainly in the foot-head (FH) direction [10] but also to additional 3D affine and non-rigid components which differ strongly between different subjects [11, 12, 13, 14].

A simple approach to minimize respiratory-induced motion artifacts in 2D cardiac MRI is to perform the acquisition during one or multiple breath-holds. Breath-holds are usually performed at end expiration which is the more stable and reproducible respiratory position and it has been shown to present similar blood flow than during free-breathing [15]. However breath-holding acquisitions may suffer from slice-misalignment due to different breath hold positions and requires patient cooperation, which may be challenging in elderly or severely ill patients (it has been shown that more than 30% of patients can have problems holding their breath in a realiable and reproducible way [16]). Moreover this approach is incompatible with the clinically preferred whole-heart approach due to the long scan time required to satisfy signal-to-noise and high-resolution requirements.

Similar to cardiac motion compensation, respiratory motion monitoring (or surrogates) can be used to combine data from multiple breathing cycles acquired at the same respiratory position (segmented acquisition). Surrogates are used to relate the chest wall or diaphragmatic motion directly with the respiarory motion of the heart. Systems to monitor the motion of the chest wall include external devices such as respiratory bellows [17] and optical compression devices [18]. Techniques to monitor the hemi-diaphragm position include navigators and ultrasound probes [19].

The most common navigator technique is the diaphragmatic one-dimensional (1D) navigator echo. A navigator echo usually consists of a two-dimensionally (2D) selective excitation pulse [20] (pencil beam) oriented, in general, along the FH direction to monitor the position of the lung-liver (right hemi-diaphragm) interface. One navigator echo is typically acquired prior, after, or on either side of the imaging sequence every cardiac cycle. The 1D Fourier transform of the navigator echo corresponds to the projection of the excited volume onto the

FH direction. Therefore, the repetitive acquisition of navigator echoes provides continuous or discrete monitoring of the diaphragm temporal FH translation. From this information, the respiratory signal can be obtained using zero moment, cross correlation or least square fitting techniques [21].

The respiratory signal provided by the navigator echoes can be used to minimize the respiratory motion by gating the acquisition to a breathing position (e.g. end-expiration). With this approach data is acquired only when the respiratory signal is within a predefined acceptance window (so called gating window) of the breathing cycle with all other data being rejected. This approach has shown to considerably reduce motion artifacts when small gating windows are employed (3-5mm), however it leads to prolonged scan times since only a fraction of the acquired data is accepted for reconstruction (referred to as scan efficiency). For example a high-resolution 3D scan with a nominal scan time of 5-6 minutes may take 10-20 minutes for a typical scan efficiency of 30-50%. Moreover, in subjects with highly irregular breathing patterns drift in respiratory motion can lead to scan abortions due to zero gating efficiency. Another drawback of the navigator-gated approach is the use of an oversimplified and patient-independent model for motion correction. The correction is typically performed only for FH translational motion (slice tracking, i.e. moving the acquired slice with the expected motion of the heart) and using a fixed factor that accounts for the relationship between diaphragmatic and cardiac motion (fixed scaling factor of 0.6 [10]). Several alternative methods have been proposed to improve the respiratory efficiency and/or image quality of the navigator echo gating approach. A description of these methods is given below.

K-space ordering techniques:

The MR signal is not measured in the image domain but in the corresponding frequency domain, known as k-space. K-space ordering techniques perform respiratory compensation by adapting k-space sampling depending on the current respiratory position. The assumption of these techniques is that motion occurring during the acquisition of the outer k-space samples is less harmful to image quality than motion during the acquisition of the central part of k-space. These methods include respiratory-ordered phase encoding (ROPE) [18], centrally ordered phase encoding (COPE) [22] and hybrid-ordered phase encoding (HOPE) [23].

Techniques not requiring a gating window:

These methods attempt to deal with irregularities of the breathing pattern and respiratory drift by acquiring data in several navigator windows simultaneously instead of setting an acceptance window prior to the scan. Examples of these methods include the diminishing variance algorithm [24], the motion-adaptive gating technique [25] and phase ordering with automatic window selection (PAWS) [26, 27, 28].

Subject-specific motion modeling techniques:

These techniques have been proposed to improve the tracking factor used to relate diaphragmatic to cardiac motion. These methods use a low-resolution calibration scan prior to image acquisition to generate translational or affine (translation, rotation, scaling and shearing) transformation parameters to subsequently correct the respiratory-induced cardiac motion [29, 30].

Self-navigation techniques:

Self-navigation methods have been proposed to derive the respiratory motion from the acquired data itself without the need of either a navigator echo or a heart-diaphragm tracking factor. Respiratory-induced displacements can be estimated from the repetitive acquisition of: a) the central k-space point (0D self-navigation) [31], b) the central k-space line (1D self-navigation) [32, 33, 34, 35, 36], or c) segmented k-space lines (2D self-navigation,) [37, 38]. A drawback of the 0D and 1D approaches is the inclusion of static structures, such as the chest wall, which can degrade motion compensation to some extent.

Image-based navigators for high-respiratory scan efficiency:

To overcome these problems and account for more complex motion several 2D image-based navigator (so called here iNAV) approaches have been recently proposed for cardiovascular MRI [40, 41, 42]. In these approaches a low-resolution 2D image is acquired before (or after) the actual image acquisition and used to estimate directly the respiratory induced-cardiac motion via (rigid, affine or non-rigid) image-registration of iNAVs at different respiratory positions. iNAVs are binned at different respiratory positions usually using a 1D diaphragmatic surrogate (e.g. navigator echo or self-gating). The estimated motion is then used to prospectively or retrospectively correct for the respiratory-induced cardiac motion in a high-resolution 2D or 3D cardiovascular MRI acquisition. These methods eliminate the need for respiratory motion models and can use all acquired data for reconstruction (100% scan efficiency). 3D image-based navigators have been recently proposed for estimating the affine 3D cardiac motion, instead of just 2D. A Cartesian approach that uses the startup-echos of a balanced-SSFP acquisition to acquire low-resolution 3D images [43], and a radial sampling approach that reconstructs low-resolution 3D images from the actual acquired data [44] have been recently proposed for whole-heart coronary MR angiography. However, in all these approaches the estimated motion has not been used to correct the acquired data itself but to transform respiratory-resolved undersampled images to a reference position aiming to improve signal-to-noise ratio through image averaging.

Image-based navigators with motion correction incorporated in the reconstruction:

Methods that estimate complex motion (affine or non-rigid) from image-registration of image-based navigators and correct for such motion directly in the reconstruction have been recently introduced. These methods are based on a generalized motion correction framework [45] introduced by Batchelor et al. in 2005. This framework model the transformation from the motion free image to the acquired motion corrupted k-space samples at different motion states via a matrix-vector equation, and standard numerical matrix inversion algorithms can be used to reconstruct a motion corrected image. Using information from multiple coils in parallel MRI, the application of this framework has been recently demonstrated in non-rigid motion compensated 3D coronary MRI [46] and 2D cardiac CINE [47, 48, 49]. Recently this generalized motion correction formulation has been incorporated directly into the Compressed Sensing reconstruction [50, 51, 52] of free-breathing 2D cardiac CINE MRI [53]. This last approach has the advantage of simultaneously combining an undersampling reconstruction technique to accelerate the acquisition with a general motion correction approach to achieve 100% respiratory scan efficiency. Although these techniques are currently limited to research studies they may become a preferable clinical alternative in the future.

This talk will include examples of the above methods for cardiac and respiratory motion compensation, discussing their strengths and limitations. An extensive review in motion in cardiovascular MR imaging can be found in [54].

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