Motion Correction

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Introduction

Patient motion has long been recognised as a problem in MR imaging. A 1997 study of 297 first-time adult MRI patients showed 6.4% of scans were of impaired diagnostic quality due to motion [1]. These numbers are described by the authors as being inline with an earlier German study of 21,000 cases (8-17% scans of impaired diagnostic quality) [2]. Despite modern hardware and reconstruction, motion remains a significant problem within images. Furthermore, the expectation of motion results in the use of rapid sequences that may compromise resolution, contrast or signal to noise.

This article describes the physics of motion artefacts and their correction in image reconstruction, strategies for avoiding motion and methods to infer motion prior to correction. The article and the accompanying lecture are intended as an overview for the interested reader and not a comprehensive review. The remainder of the article will be technology focused, but in practice, good patient preparation is also effective in reducing motion artefacts, see for example [3].

The Effect of Motion

An MR image is the Fourier Transform (FT) of the acquired k-space data. The summation operations in the discrete FT mean that each image pixel is composed of a weighted sum of every point in k-space. Thus any damage to k-space at any time during the acquisition can potentially affect every pixel in the image - typical artefacts being image blurring or ghosting [4].

The effect of *affine* motion on k-space is known [5], for example image translations cause a phase ramp across k-space and rotations give a rotation of k-space by the same angle. K-space is said to be 'inconsistent' if the object position changes during the acquisition. For example a nod half-way through a scan could result in one half of k-space being rotated with respect to the other half. Potentially these effects can be corrected if motion parameters are known for each 'shot' of k-space. Here a 'shot' is defined as a set of k-space points acquired in a duration of time assumed short enough for motion within the shot to be neglected. In most schemes, only the motion between the shots is corrected. Some within-shot motion problems can be reduced using flow compensation or gradient moment nulling techniques [6, 7].

In diffusion weighted imaging, the sequence is deliberately sensitised to (diffusional) motion making it very sensitive to bulk tissue motion. In this case, sub-pixel scale motion can cause large phase changes in the image [8]. For single-shot diffusion imaging this does not matter as only the modulus image is used in the diffusion analysis, however, for multi-shot sequences the phase errors differ from shot to shot and need to be corrected to prevent gross image artefacts.

Acquisition Strategies to Reduce Motion Effects

There are a number of exclusively acquisition based strategies to minimize the effects of motion. Rapid acquisitions such as EPI aim to freeze out motion. Breatholding can reduce respiration artefacts but obviously can only be applied for a short duration, especially in sick patients. Multiple averaging does not correct for motion but weights coherent (stationary) tissue signal greater than incoherent motion artefact [9]. The following strategies modify the sequence prospectively at run-time to reduce artefacts. Information from an ECG or respiratory bellows can be used to determine the respiratory or cardiac phase and gate or trigger the acquisition. Additionally, a decision on data quality can be made and if necessary data rejected and reacquired. The decision can be based on the k-space data [10] or analysis of the ECG. K-space ordering techniques such as respiratory ordered phase encoding [11] can be used to remove the periodicity between motion and phase encode order. Removing the periodicity reduces ghosts but can still leave blurring.

The remaining prospective techniques require an absolute estimate of motion parameters, rather than just phase of a cardiac or respiratory cycle. Examples include shifting the excited volume in cardiac imaging [12] and modification of the RF and gradients to apply a prospective correction for 3D affine motion [13, 14]. With prospective correction techniques, the type of motion that can be compensated is limited to affine, at least for standard imaging using linear gradients, Fourier encoding and conventional RF excitation. In order to correct for motion more complex than affine, post processing methods are currently used in combination with image reconstruction from the raw data.

Reconstruction in the Presence of Motion Artefacts

As a general principle, it is preferable to compensate prospectively for motion especially as this reduces the chances of motion sending tissue out of the excited region. However, prospective control may not be possible because the sequence modifications required to infer motion could adversely affect scan time or contrast, initial motion measures may lack sufficient precision, or, the data processing and feedback may be too slow to interface to a running scan. For these reasons, and the requirement to cope with non-rigid motion, post-processing and reconstruction techniques have been developed. Outlined here is a matrix formalism that encapsulates the physics of the acquisition and the cause of the artefact. This is based on the paper by Batchelor et al [15] and related work [16, 17, 18]. Whilst at first sight this may look complicated, a matrix formalism brings a number of significant advantages: the well-developed tools of linear algebra are available to solve the system; the incorporation of the coil sensitivities means that we are effectively doing a parallel imaging reconstruction, and the ability of the coils to 'bridge' gaps in k-space is useful, especially if motion leads to local undersampling (pie slice gaps following rotation [19] or gaps following data rejection); the system can represent artefacts caused by motion, phase changes and potentially contrast changes; when the matrix equation is solved iteratively we do not have to explicitly compute any k-space sampling density changes that may be caused by motion.

The measured k-space for shot t and coil γ, written as column vector $\mathbf{m}_{\gamma,t}$ can be expressed as,

$$\mathbf{m}_{\gamma,t} = \mathbf{\Theta}_t \mathbf{F}_{k \leftarrow i} \mathbf{C}_\gamma \mathbf{M}_t \ \boldsymbol{\rho_0}. \tag{1}$$

Here the columns of the (unknown) artefact free image are stacked into one long column vector $\boldsymbol{\rho_0}$. The effect of the artefact at time t is expressed as a matrix $\mathbf{M}_t$, $\mathbf{C}_\gamma$ represents the spatial sensitivity of coil γ, $\mathbf{F}_{k \leftarrow i}$ is the Fourier transform from image to k-space and $\mathbf{\Theta}_t$ is the sampling of k-space for shot t. Here the coils are assumed stationary and time invariant though the formalism can be readily adapted to moving coils. As mentioned above, this approach is very flexible because $\mathbf{M}_t$ can represent a range of artefact causes such as the phase errors in multi-shot

diffusion imaging due to pulsatile brain motion, or image intensity and phase changes due to flow and contrast agent, or non-rigid motion.

At first glance, one might expect that only affine or 'linear' motion could be represented by a matrix $\mathbf{M}_t$. It is important to note that here the matrix multiplies the image pixel intensities and not their coordinates. (In the situation where matrices multiply a vector of coordinates then only affine transforms can be applied.) In an extreme case, $\mathbf{M}_t$ could be a permutation matrix that reorders the pixels in the image and this operation clearly corresponds to non-linear motion. To represent more conventional deformations, the rows of $\mathbf{M}_t$ act as interpolation kernels on the intensities in $\boldsymbol{\rho}_0$. For small kernels, e.g. linear interpolation, this leads to a sparse $\mathbf{M}_t$.

To solve the system all the terms except $\boldsymbol{\rho_0}$ are stacked to account for measurements from all the coils and all shots. The system can then be written as $\mathbf{A}\boldsymbol{\rho_0} = \mathbf{m}$ and solved for $\boldsymbol{\rho_0}$ using a conjugate gradient solver such as LSQR [20]. The matrices in equation 1 that form $\mathbf{A}$ are very large but the conjugate gradient solver requires only the matrix-vector products $\mathbf{Av}$ and $\mathbf{A}^H\mathbf{v}$ where $\mathbf{v}$ is a vector supplied by the solver. These products can be computed efficiently in functions using image pixel-wise operations and the Fast Fourier Transform (FFT). In the forward direction ($\mathbf{Av}$), the algorithm supplied vector $\mathbf{v}$ is reshaped to an image, the artefact cause is applied (e.g. a transformation), the new image is multiplied by the coil sensitivities, an FFT to k-space is applied and the sampled points returned as a vector.

In the reverse direction ($\mathbf{A}^H\mathbf{v}$) we need to evaluate,

$$\mathbf{A}^H\mathbf{v} = \mathbf{M}^H\mathbf{C}^H\mathbf{F}^H_{k\leftarrow i}\boldsymbol{\Theta}^H\mathbf{v}. \tag{2}$$

Here the matrices have similar meanings to before, except the coil and shot subscripts have been removed to indicate that this expression must account for all coils and shots. In practice this means the input vector is assembled into a zero-filled array using knowledge of the sampling pattern, the matrix representing the Fourier Transform is unitary so $\mathbf{F}^H_{k\leftarrow i} = \mathbf{F}_{i\leftarrow k}$ and we just perform the opposite FFT (possibly taking into account the non-symmetric $1/N$ scaling applied by many FFT algorithms), the Hermitian transpose of the coil sensitivity matrix is just the complex conjugate of the values (note when written as a matrix, $\mathbf{C}_\gamma$ is diagonal thus its Hermitian transpose is just the complex conjugate of the entries). This leaves us with the problem of how to evaluate $\mathbf{M}^H\mathbf{w}$ where $\mathbf{w}$ is the result of the previous matrix-vector products.

When the artefact cause is a pixel-wise multiplication by the image, the Hermitian transpose operation is just a complex conjugation (similar to the coil sensitivities). This situation occurs in multi-shot diffusion imaging where the errors are just phases [17, 21]. Also, it might be expected in flow correction or contrast enhanced MRI when the pixel intensities change magnitude or phase. For more general motion, we need to be able to evaluate $\mathbf{M}^H\mathbf{w}$. One approach is to apply an image deformation using the negative displacements of those used in the forward transformation, another approximation is to use the inverse transformation. The matrix has a sparse structure and in some situations it can be stored and the conjugate transpose explicitly formed. Finally, it is possible to correctly evaluate $\mathbf{M}^H\mathbf{w}$ using for-loops and accumulating the result in an array. (This computation is possible because we know where the entries in $\mathbf{M}$ and hence $\mathbf{M}^H$ should be, based on the motion deformation and the interpolation kernel applied.)

An issue with conjugate gradient solvers is determining the stopping criteria. It is known that terminating on iteration number acts like a regularization term and can be used to prevent noise amplification [22]. As yet, there has been little published on choosing the stopping criteria, or alternative regularisation strategies, for this particular application. Typically authors let conjugate gradient algorithms run for the order of 10 iterations.

The above techniques, and more complex nonlinear solvers (e.g. [23]), provide a powerful mechanism for motion correction but they require a measure or estimate of the motion.

Measuring or Estimating Motion

In order to apply a correction (rather than just gate or trigger) in either retrospective or prospective schemes, the motion needs to be known. Some of the techniques for determining motion are listed below.

Navigator echoes (non phase encoded k-space lines) provide a projection through the object from which motion can be inferred [24]. Recent applications of the central profile approach include their use in balanced cardiac sequences with signals recorded from multiple coils [25]. To isolate a specific region, usually a dome of the diaphragm, pencil-beam excitations are often used. These methods provide motion information in 1D and multiple beams can be used to gain more motion information [26]. Navigators with more complex trajectories such as orbital [27], spherical [28] and cloverleaf [29] have been developed to provide more motion information, as well as floating navigators with a small phase encoding [30]. As a means of trying to detect specifically cardiac motion, navigators from the fat signal near the coronaries have been developed [31]. It is also possible to capture the FID following an RF pulse and obtain motion information [32].

The central region of k-space can be scanned rapidly and provides a low resolution image from which motion can be determined. Radial and spiral sequences inherently have a high sampling density near k-space centre (as well as some natural robustness to motion) and can be used to infer motion. The PROPELLER [33] method obtains low resolution images from rotating blades of k-space.

External sensors can provide additional information with no compromise to the scan sequence or imaging. Optically tracked markers or active markers cannot directly measure the motion of internal tissues but have been the subject of recent demonstrations for neurological imaging [34, 35, 36].

A range of algorithms have been developed that aim to detect motion from the data without the need of other motion measures. These include autofocus [37, 38, 39] and autocorrection [40] where entropy or other metrics [41] are used to assess image quality as part of an optimisation scheme that aims to find the motion during the scan (once motion is found and corrected, the metric should show a minimum). Here the prior knowledge is in the interaction of typical motion artefacts with the quality metric. It is also possible to use data acquired from multiple coils [42, 43] to optimise data consistency, or to predict data and then compare and correct the measured data [44]. A recent approach extended the FOV to enable correlation of adjacent k-space lines [45]. If the patient location is known from other data acquired in the same session then POCS methods may be used [46]. Finite support and edge enhancement have also been used to generate k-space for comparison to determine motion [47]. With no prior knowledge of motion, we are often limited to determining simple motion with a small number of unknown parameters per shot.

In order to correct for more complex motion (i.e. non-rigid, non-affine), arbitrary displacement fields at each shot need to be described. These can be estimated by incorporating a motion model derived from training sessions [48, 18]. The estimated fields are then used to correct the data using the reconstruction described by equation 1 (with $\mathbf{M}_t$ representing non-rigid motion). In order to avoid errors between calibration and prediction of the displacement fields, it is possible to combine image reconstruction by equation 1 and motion determination into a coupled optimisation problem called 'GRICS' [49], leading to an autocalibrated motion model. Within GRICS the residual reconstruction error (arising from initial motion estimation errors) is linearized using optic flow and minimised in order to detect non-rigid motion.

Outlook

In the future we might expect to see a more coherent integration of prospective scanner control, motion measures, learned models of motion and reconstructions that include not just inter-shot

motion but effects such as motion during a shot, T2 signal decay and field inhomogeneities. Advances in computing power through parallel processing enable faster data processing and the area of parallel RF transmit offers the possibility of new ways to monitor and perhaps compensate prospectively for motion during acquisition. We may also be able to use motion more to our advantage, either to modify the k-space acquisition, or to provide additional information about tissue properties.

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