REAL-TIME REDUCTION OF MOTION ARTEFACTS USING K-SPACE WEIGHTING IN MAGNETIC RESONANCE IMAGING

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Abstract

An improved version of motion-adapted gating useful in MR imaging is presented. Compared to other gating approaches, this advances method shows an improved ability to suppress motion artefacts at a reduced scan time. It is based on a k-space dependent weighting function and a real-time system feedback. During MR data acquisition patient motion is monitored and only those profiles are accepted whose respiratory motion induced displacements are below a pre-defined threshold (gating) function. The displacements are measured relative to a reference position, which is automatically determined during the initial phase of the MR scan. While MR data are acquired, the object motion is statistically analysed in parallel by calculating the histogram of the displacement distribution for consecutive time intervals. This information can be used to interfere with the measurement process in the case that motion statistics changes during the scan. Initial in-vivo results are presented and compared to conventional techniques.

Keywords: abdominal MRI, artefact reduction, diminishing variance algorithm, k-space weighting, motion adapted gating, respiratory motion.

1. Introduction

Respiratory motion that takes place during magnetic resonance (MR) data acquisition gives rise to artefacts in the final image. Especially in abdominal imaging, image quality is severely degraded due to blurring and ghosting [1]. Ultra-fast MR imaging techniques [2,3], whose total acquisition time is short compared to the time scale of the respiration, could be used to overcome this problem using the single breath-hold method. However, a variety of clinical MR protocols employ sequences that cannot be performed within that short
time interval. Thus, many techniques for motion compensation have been developed [4–10].

One approach is based on the measurement of the respiratory motion simultaneously to image acquisition. For motion detection, external devices or MR signals, such as navigator echoes [8,9] have been proposed and successfully used for gating purposes. Here, only those profiles are used for the final image, whose motion state deviates by a fixed threshold from a pre-defined reference position. Thus, the motion is frozen relative to the reference state. The selected threshold determines the compromise between the improved image quality and the prolonged scan time. In addition, the chosen reference position should have a high probability of occurrence in order to minimize the overall scan time. There are various approaches to obtain this reference position. Either one starts the scan in an exhaled position and uses the first position measurement as the reference, or one determines the reference during a preparation phase.

A more elaborated approach to circumvent the difficulties associated with the proper choice of the reference state has been proposed recently by the so-called Diminishing Variance Algorithm (DVA) [10]. This approach can be divided into two parts. During the first, all profiles that are necessary to reconstruct an image are acquired, regardless of the corresponding motion states which are monitored in parallel. These data are used to generate a histogram that shows the probability of occurrence for the various motion states. The most frequent one is chosen as the reference position (seed). The second part of the DVA data acquisition is dedicated to the re-acquisition of those profiles which deviate most from the seed. Thus, subsequently the width of the displacement distribution narrows and different criteria can now be applied to terminate the scan [10]. For example, a fixed number of rescans can be chosen or the scan can be terminated if the final displacement distribution becomes narrower than a given threshold. However, the algorithm does not consider that the central regions of k-space are more sensitive to motion-induced errors than the outer ones [11]. This is the essential idea of the recently suggested motion-adapted gating (MAG) approach [12] that is based on a k-space dependent gating threshold function. Here, only those profiles are accepted, whose corresponding displacements $\Delta y$ (measured relative to the seed) are below a pre-defined gating threshold function, which varies as a function of $k_y$. Thus, during data acquisition, the system analyses the motion and decides, in real-time, which profiles in k-space are compatible with the current motion state. Still, in the previous version of MAG, there was a need for a preparation phase to determine an appropriate seed ($y_{ref}$) which thus prolonged the total measuring time. In addition, macroscopic patient-motion during scanning
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Fig. 1. The weighting functions of different gating procedures are schematically depicted. The functions indicate the maximum allowed displacement $\Delta y$ relative to the seed for all $k_y$ values of the image. The simplest approach is shown in (a), only a fixed threshold is applied regardless of the $k_y$ value of the corresponding profiles (conventional gating). In fractional gating (b) only the central region of k-space is strongly gated while large values of $\Delta y$ are allowed for those profiles whose $k_y$ values are large. The so-called weighted gating approach (c) allows a smooth transition from the central to the outer k-space.

may have invalidated $y_{ref}$ as a good choice at a later stage of the scan leading to an unacceptable increase in scan time. The aim of the present work is to merge the benefits of the DVA concept with the MAG ideas. We present an improved MAG approach [13], which is capable of reducing the total scan time, omits the preparation phase and minimizes the dependence on patient co-operation.

2. Method

The idea of k-space weighting is based on the assumption that there are regions in k-space which can be measured with less accuracy than others without significant loss of image quality. This assumption has been justified for spin warp imaging by analysing the artefact contribution of different k-space regions during in-vivo motion [11]. It is shown that the central regions in k-space are more sensitive to motion-induced errors than the outer ones. This result has to be interpreted carefully, since the outer regions of k-space carry the information of the fine structure of the imaged object, while the central k-space determines the coarse contrast of the final image. The choice of a particular weighting function represents the trade-off between the final image quality and the scan time. For example, in Fig. 1 different weighting functions are shown.

In MAG the current given motion state is used to select a phase-encoding step, i.e. a profile, which is compatible with the weighting function that is measured next. To improve the robustness of this approach against a chosen seed which has a low probability of occurrence, an initial data acquisition phase was added to the MR imaging experiment. Here, since there is no a priori
reference, profiles are acquired at random order to increase the probability of matching by chance the weighting function, once \( y_{\text{ref}} \) has been determined later. During this initial phase, local minima in the respiration curve are registered. In real-time, all measured displacements \( y \) of one or more respiratory cycles are filled into a histogram, until the number of entries exceeds a predefined value \( (N = 100) \). The choice of this parameter depends on the temporal resolution of navigator measurement to ensure reasonable statistics. This histogram thus represents the respiratory motion statistics during a number \( M \) (an integer multiplicative) of breathing cycles. This yields the most probable position that can be used as the reference in the subsequent MAG process. During the ongoing process of data acquisition, two kinds of histograms are generated in steps of \( M \) breathing cycles: at first a local distribution reflecting the current motion statistics is calculated and secondly an accumulative one that takes the entire history of the scan into account. From these distributions, the most abundant states can be extracted. A wide difference between both values over several cycles indicates macroscopic patient movement and thus may trigger a restart of the scan or a message to the operator in future implementations.

The entire data acquisition process of the improved MAG procedure is illustrated in Fig. 2 using data obtained from a measurement of a volunteer together with the respiratory motion curve. A parabola-shaped \( k \)-space weighting function (c.f. Fig. 1c) was used, which assigns to each \( k_y \) value a maximum allowed displacement \( \Delta y \). The initial phase, where the phase encoding steps are applied at random order, clearly separates from the motion-adapted gating phase. This phase is considerably shorter than that used in the DVA approach. At the end of the initial phase the seed is determined, which serves as a reference state for the subsequent motion-adapted gating process, during which yet missing profiles are measured according to the weighting function, where priority is always given to the profile with the lowest \( k_y \) value compatible with the current \( \Delta y \). If for a given \( \Delta y \) all compatible profiles have already been acquired, those profiles are re-measured whose corresponding values of \( \Delta y \) are above the required weighting function. In case that all the \( \Delta y \) values of these profiles are below the weighting function, the profile with maximal \( \Delta y \) is improved. This entire chain of evaluations and decisions must be performed in real-time. It is reasonable to continue data acquisition until all values of \( \Delta y \) are below the required weighting function.

The very difference between MAG and DVA is most obvious at the stage of data acquisition when the seed has been determined. While DVA re-measures those profiles deviating most from the seed regardless of their \( k_y \)-value, MAG re-acquires only those profiles incompatible with the weighting function. Thus,
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Fig. 2. Schematic representation of the MAG data acquisition process illustrated using in-vivo data. In (a) the respiratory motion curve as a function of the shot number together with the corresponding histograms of the motion statistics are shown. Local minima in the motion curve are used to trigger the evaluation of the histograms in integer multiplicatives M of the breathing cycle. The so-called *sliding* histogram contains only entries from the M current respiratory cycles, thus it is a local distribution. The so-called *accumulative* histogram contains all entries of the past and it is therefore less sensitive to fluctuations of the motion curve. The most probable state of the first sliding histogram becomes the reference position (seed) for the subsequent data acquisition. Thus, the initial random phase is terminated and the MAG phase begins. This is presented in (b) where the acquired profile number versus the shot number is plotted. The various marker types represented the reasons why the MAG algorithm has chosen a particular profile to be measured. Vertical dotted lines indicate instances where no profiles have been acquired.
for a reasonable function, the central \( k \)-space data will always be more accurately measured than the outer \( k \)-space data.

3. Experimental

Experiments were performed on a research system operating at 1.5T (based on Gyroscan NT15, Philips). The MR scanner is equipped with a self-shielded gradient system that delivers 21 mT/m at a slew rate of 100 mT/m/ms. The standard built-in body coil was used for signal reception. The MR system architecture allows a change in imaging parameters during scanning in real-time based on a feedback control [14]. This real-time capability of the MR system is essential to perform motion-adapted gating.

A fast-field echo imaging sequence was used to obtain abdominal images of healthy volunteers in the coronal orientation. For motion detection, a MR navigator signal was used. This is based on a slice-selective gradient echo which is read out in the motion sensitive direction. The navigator echo acquisition was interlaced between each acquisition of a \( k_y \)-line. The sequence is schematically depicted in Fig. 3. Timing parameters and net gradient induced phases were kept identical for the navigator and the image profile. From each navigator echo the displacement \( y \) was determined relative to a pre-stored navigator profile obtained in an initial dummy cycle phase. The displacement was calculated in real-time using a least-squares algorithm [13] implemented on the receiver of the MR scanner. This information was passed over to the MAG/DVA routine (c.f. Fig. 3). The necessary system response time measured from the end of the navigator echo acquisition until the experiment could be varied from 4 ms to 6 ms. The total duration of these experiments was artificially increased to cover a reasonable number of respiratory cycles during the scan. Therefore, only one data profile was measured after an preceding navigator. In addition, a long profile repetition time (TR) of 45 ms was chosen while the effective echo time (TE) was set to 6 ms. The final data matrix was set to \( 256 \times 256 \) in a rectangular field of view (FOV) of 450 mm. Signal accumulation was performed by acquiring 512 profiles in the phase encoding direction while oversampling that direction by an factor of two. This virtually increases the numerical resolution in the phase encoding direction. The final image was obtained by chopping the reconstructed data set. The slice thickness was set to 10 mm. The slice selective navigator was oriented along the AP direction to be most sensitive in that direction. This is the same direction as used for phase encoding in the imaging experiment. For all experiments performed, the navigator information was measured in a slice parallel to the slice of interest in the imaging experiment. The navigator resolution was higher compared
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Fig. 3. Schematic representation of the gradient echo pulse sequence used. The acquisition of a line in the image raw data set is preceded by a slice selective navigator measurement. The navigator echo is Fourier transformed and correlated with a reference projection to extract the actual motion-induced displacement $y$. According to the motion state and the scan history, the profile to be measured next is selected by the MAG/DVA procedure. All these calculations were performed in real-time. Simultaneously, motion statistics are performed, i.e. histograms are filled and processed.

to the imaging pixel size and was set to 1 mm to resolve the diaphragm motion with sufficient accuracy.

The comparison between MAG/DVA was drawn with respect to image quality for approximately constant total scan time. Therefore, a fixed number of re-scan was used as a DVA termination criteria. This fixed scan time was chosen to be the average scan time used in a number of improved MAG scans.

4. Results and discussion

Measurements using different gating parameters were performed to compare the improved MAG and the DVA approach. Selected representative
results are given in Figs 4 to 6 and will be discussed now. To illustrate the data acquisition process the respiratory motion curve and the acquired profile numbers are plotted as a function of the shot number. The final image quality is dependent on the motion-induced errors present in the measured data used for image reconstruction. Thus, in order to judge the gating efficiency, additionally the final relative displacement $\Delta y$ (relative to the seed) is plotted as a function of the profile number in $k$-space. In Fig. 4 results obtained without any gating are shown. The respiratory motion is clearly imposed as motion-induced error in $k$-space data since during data acquisition the $k$-space is covered sequentially from $-k_{\text{max}}$ to $+k_{\text{max}}$. Thus, the final image suffers severely from ghosting and blurring (Fig. 4c). This example is given to demonstrate the general need for respiratory gating. In Fig. 5 the data acquisition process of the DVA gating technique is shown. The initial phase extents up to shot number 512, which is equal to the total number of profiles required to reconstruct an image. In this example, profiles are acquired in random order with respect to $k$-space. After the initial phase the seed is determined via the histogram analysis that describes the motion statistics. In the following DVA-phase, those profiles are re-measured which deviate most from the seed regardless of their position in $k$-space. This is reflected in the histogram of the final displacements versus the profile number. The distribution of $\Delta y$ is almost flat and no care is taken to measure the central $k$-space data with higher accuracy than the outer one. The resulting image (Fig. 5c) thus exhibits motion artefacts. However, further experiments showed that image quality can be considerably improved if the number of allowed re-scan for DVA techniques is increased. In Fig. 6 the data acquisition process of the MAG approach is presented. After an initial random phase of about 200 shots the seed is determined and the algorithm starts to acquire profiles according to the earlier described procedure. The histogram of final relative displacements $\Delta y$ satisfies the imposed weighting function. Thus the final image quality (Fig. 6c) is satisfactory and superior to the one of Fig. 5c. No motion artefacts are visible any more. It is important to emphasize that the MAG approach performs, for equal total acquisition time (900 shots), significantly better when compared to DVA.

The compilation of all acquired image data sets allows the drawing of several conclusions:

(1) The motion artefact suppression and thus the image quality obtained by the MAG approach is always superior or at least comparable to the DVA method for equal total scan times. This is a consequence of the additional feature of a $k$-space dependent weighting function in the MAG approach.
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Fig. 4. The process of image data acquisition as obtained without any gating is shown. In (a) the respiratory motion of the patient and the acquired profile number in k-space as a function of the shot number are depicted. A linear acquisition scheme is applied. Thus the motion-induced errors are clearly visible in the final distribution of $\Delta y$, plotted versus the profile number (b). The resulting image suffers from severe ghosting and blurring (c).
Fig. 5. The process of image data acquisition as obtained with the DVA approach is shown. In (a) the respiratory motion of the patient and the acquired profile number in $k$-space as a function of the shot number are depicted. The initial random phase (up to shot number 512) clearly separates from the subsequent DVA-phase. Since the profiles are re-measured regardless of their position in $k$-space, the resulting errors $\Delta y$ are distributed at random in $k$-space (b). The resulting image (c) already shows an improvement, but the quality is not yet satisfactory. A scan time prolongation would reduce the present motion artefacts.
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Fig. 6. The process of image data acquisition as obtained with the MAG approach is shown. In (a) the respiratory motion of the patient and the acquired profile number in k-space as a function of the shot number are depicted. Again, there is a clear separation from the initial random phase (up to shot number 220) and the following MAG-phase. The various marker types indicate the different criteria why the MAG routine has decided to acquire a certain profile. The final distribution of motion induced errors $\Delta y$ (b) satisfies the imposed weighting function (filled line). Thus the final image is expected to be free from severe motion artefacts because the central k-space has been measured accurately relative to the seed. The resulting image in given in (c).
(2) The image quality of DVA varies from one image acquisition process to another, because by chance the central k-space might be measured more or less accurately relative to the seed. This is a consequence of the chosen termination criteria (fixed scan time). Thus MAG, with its inherent feature to measure the central k-space more accurately, always produces, for a given weighting function, comparable results.

(3) The weighting function determines the compromise between image quality and prolongation of the scan time. Its very shape has its counterpart for example in the minimal width of the residual displacement histogram that can be used to terminate the scan in DVA. However, even if the scan time varies due to distortions in the respiratory motion of the patient, MAG produces the requested image quality, while for DVA it becomes a matter of luck because no attention is paid to the central k-space profiles. In the limit of very long scan times both methods converge to similar image qualities as all profiles are measured with a zero displacement \(\Delta y\).

(4) The effect of the initial phase on the type of profile order acquisition, (i.e. randomly or linearly) was studied. In the case of MAG the final image quality does not depend on how profiles are initially acquired while this is not the case for DVA. Here, a linear acquisition order significantly deteriorates the final image quality. This is because initially induced motion errors are subsequently imposed into k-space and, later in the DVA-phase, not removed or scrambled any more. The data acquisition of MAG performs, in that respect, much better.

(5) The introduction of the initial random phase improves the performance of the algorithm when compared to the old version of MAG. Profiles acquired at this stage happen to match the weighting function, by chance, once the seed has been determined at a later stage of the data acquisition process. This is the case for about 15 to 20% of all profiles and depends on the total scan time and the duration of the initial phases. This is a significant improvement with respect to scan time reduction because those profiles do not need to be re-measured.

5. Conclusion

An improved real-time motion-adapted gating procedure was presented, which uses k-space weighting that allows different displacement \(\Delta y\) for different \(k_y\) values. Preparation phases or patient co-operation are minimized since the reference position is statistically determined during an initial acquisition phase and monitored during the entire scan. Thus, possible macroscopic
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patient motion can be detected, followed by appropriate system responses. Depending on the weighting function motion, artefacts are reduced at the cost of an increased scan time. Compared to the DVA algorithm the MAG approach leads to an improved image quality at a comparable acquisition time.

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REFERENCES


Authors Biographies

Ralph Sinkus was born in Hannover, Germany, in 1967. He received his Master’s degree in Physics from Hamburg University in 1993. He joined the ZEUS collaboration at the particle collider HERA/ZEUS (Hamburg, Germany) in 1993, where he worked as a PhD student on the application of artificial intelligence for particle identification and the measurement of the structure function F2 of the proton. In 1996 he received his PhD degree from Hamburg University and worked as a post-doc at Tel Aviv University, Israel. Since 1996 he has worked as a research scientist in the Magnetic Resonance group at the Philips Research Lab in Hamburg.
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