IMAGING METHODOLOGY - Review

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Prospective Motion Correction in Brain Imaging: A Review

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Motion correction in magnetic resonance imaging by real-time adjustment of the imaging pulse sequence was first proposed more than 20 years ago. Recent advances have resulted from combining real-time correction with new navigator and external tracking mechanisms capable of quantifying rigid-body motion in all 6 degrees of freedom. The technique is now often referred to as "prospective motion correction." This article describes the fundamentals of prospective motion correction and reviews the latest developments in its application to brain imaging and spectroscopy. Although emphasis is placed on the brain as the organ of interest, the same principles apply whenever the imaged object can be approximated as a rigid body. Prospective motion correction can be used with most MR sequences, so it has potential to make a large impact in clinical routine. To maximize the benefits obtained from the technique, there are, however, several challenges still to be met. These include practical implementation issues, such as obtaining tracking data with minimal delay, and more fundamental problems, such as the magnetic field distortions caused by a moving object. This review discusses these challenges and summarizes the state of the art. We hope that this work will motivate further developments in prospective motion correction and help the technique to reach its full potential. **Magn Reson Med 69:621–636, 2013. © 2012 Wiley Periodicals, Inc.**

Key words: review; prospective motion correction; MRI; motion tracking

INTRODUCTION

The first clinical application of MRI was imaging of the brain and head. Now, over 30 years later, artifacts caused by head motion during MR imaging of the brain are still an unsolved problem in many clinical imaging situations. This incurs a significant financial cost, due to repeated scans and the need for anesthesia, and can adversely affect patient outcome if images of diagnostic quality cannot be obtained. These problems are particularly severe in pediatrics (1,2), stroke patients (3), and in the elderly.

Head motion is a major problem not only in clinical imaging, as described above, but also in research. In high-resolution imaging with cooperative subjects, scan time, and hence spatial resolution, is largely limited by subject motion. In diffusion-weighted imaging (DWI), motion can introduce a bias in diffusion values such as fractional anisotropy and mean diffusivity (4). In functional MRI (fMRI), motion correlated to the stimulus can produce false activations (5); this is particularly problematic when motion is part of the study, such as movement tasks in fMRI. In spectroscopy, the effects of motion are particularly insidious, as motion reduces spectral quality and diagnostic value, but often does not cause distinctive artifacts and so can go unnoticed. This is exacerbated by the long scan times required for spectroscopy, which increase the probability that major head motion will occur.

Although numerous motion prevention and correction techniques exist, current solutions are limited in their applicability. Restraints are only partially effective and tend to be uncomfortable. Anesthesia has the obvious disadvantages of patient safety and cost. Self-navigation techniques, such as PROPELLER (6), and its variants, are of immense value in clinical MRI but are often limited to two-dimensional imaging or particular imaging sequences and trade scan time for robustness to motion. Cardiac and respiratory gating are widely used but also sacrifice scan time and are only suitable for the correction of periodic motion. They are not effective for involuntary patient head motion.

This review discusses prospective motion correction, which is a general technique that involves updating the pulse sequence in real time. To our knowledge, this was first proposed by Haacke and Patrick in 1986 (7), who corrected for object "scaling" caused by respiratory motion in abdominal imaging by updating the phase encode gradient in real time. An early use of the term "prospective motion correction" in MRI was in a 1996 paper by Lee et al. (8) and in a second paper published two years later (9). The work in Refs. 8 and 9 built on the seminal paper of Ehman and Felmlee (10), which described their "navigator echo" technique for imaging moving structures by applying phase corrections in post-processing (see also Ref. 11). In 1997, both Danias et al. and McConnell et al. showed that prospective correction during free-breathing coronary angiography is an effective means to compensate for respiratory motion (12,13). Eviatar et al. (14,15) then suggested six-degree-of-freedom correction for brain imaging, and, in Ref. 15, to use an external tracking system completely independent of the MR pulse sequence to obtain the head pose information. Unfortunately, Refs. 14 and 15 are single-page abstracts only and lack in vivo data. The next leap forward came in 1998, when Derbyshire et al. presented their dynamic scan-plane tracking method, using small coils to measure position and orientation (16). Prospective correction with image-based tracking was not far behind: in 2000, Thesen et al. published their method

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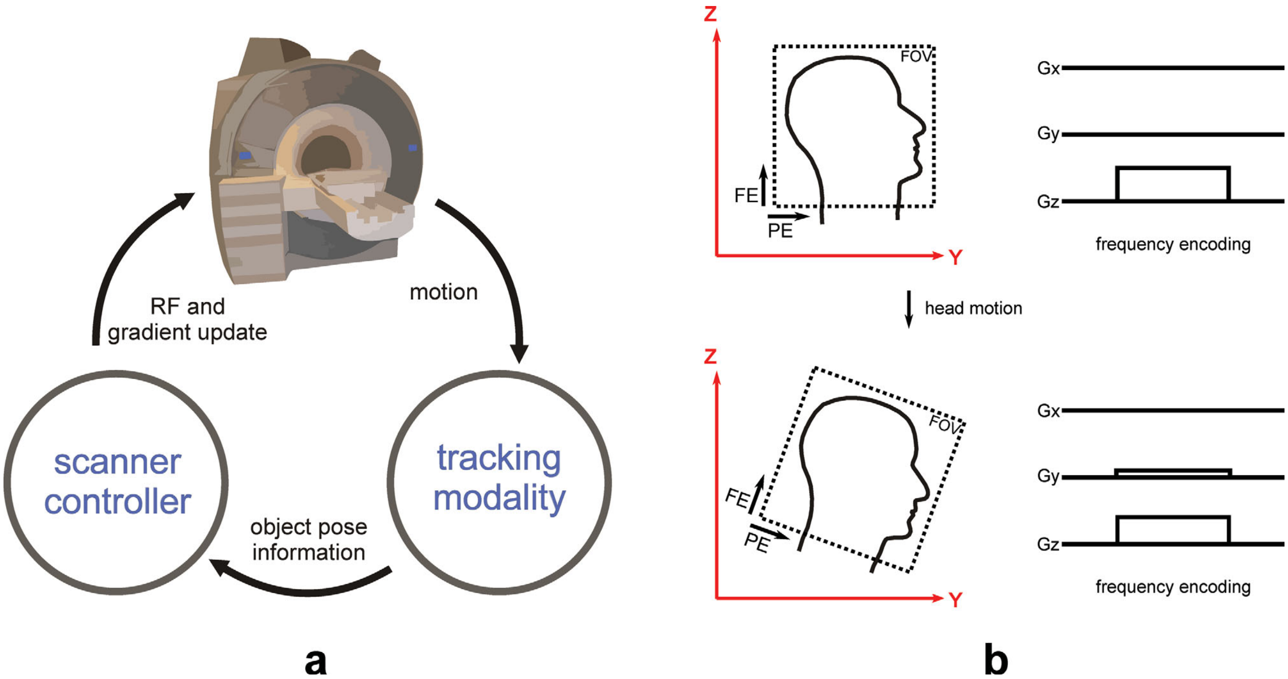

FIG. 1. **a**: Prospective motion correction uses tracking data describing the current pose of the imaged object to update the pulse sequence in real time. **b**: After a rotation, gradient directions are modified so that any given voxel in the sample experiences the same field as it would have if the rotation had not occurred. This process necessitates the recalculation of the physical gradient waveforms that are needed to generate the desired logical gradient. In the example shown, the frequency encode gradient initially requires only one physical gradient, Gz; after head rotation, the frequency encode gradient requires both Gz and Gy. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

for correction of motion during fMRI (17). The first successful implementation of prospective rigid-body motion correction in all 6 degrees of freedom using a camera system was then reported in 2006 (18).

Over 15 years after the publication of Ref. 8, referring to "prospective motion correction," much remains to be done, and there are still many unanswered questions concerning the limitations of the technique. However, due to recent progress, prospective motion correction has become more practical and now holds great promise for a number of imaging situations. As it is such a general approach (in principle, prospective correction can be applied to all imaging sequences if an external tracking system is used), it could have major implications for clinical practice. The combination of this high potential impact with the many unsolved questions that remain makes prospective motion correction a growing field of research.

In this work, we explain the basics behind prospective motion correction, review recent implementations of the technique, and summarize its advantages and problems. We focus on rigid objects (in particular, the head) but also include notes on other potential applications. The goal of this review is to give those new to the field insight into the state of the art and motivate the development of solutions to the present challenges.

THEORY

Prospective motion correction involves the application of a single, intuitive strategy: to maintain a constant relationship between the imaged object and the imaging volume, even under motion. In the majority of implementations discussed in this review, the strategy is achieved by obtaining tracking data representing the pose (position and orientation) of the object during MR imaging, passing these data to the scanner with minimal delay, and adjusting the MR pulse sequences so that the imaging volume "follows" the object (Fig. 1).

Several theoretical descriptions of prospective motion correction exist (e.g., Refs. 19–22) and will not be repeated here. It is useful, however, to summarize the main result. Assume that a point in the imaged object undergoes a transformation with 12 degrees of freedom representing rotation, scaling, shearing, and translation. This transformation, which is generally referred to as an affine transformation, can be described by

$$\mathbf{x}'(t) = \mathbf{A}(t)\mathbf{x} + \mathbf{t}(t), \qquad [1]$$

where $\mathbf{A}(t)$ is a time-varying linear transformation (representing rotation, scaling, and shearing) and $\mathbf{t}(t)$ is a time-varying translation vector. The translation of the object, $\mathbf{t}(t)$, can be then compensated for by changing the radiofrequency (RF) transmit frequency and receive phase. However, to compensate for $\mathbf{A}(t)$, the original gradient waveform, $\mathbf{g}(t)$, must be transformed to $\mathbf{g}'(t)$ by

$$\mathbf{g}'(t) = \mathbf{A}(t)\mathbf{g}(t). \qquad [2]$$

Equation [2] states that to compensate for an affine transformation of the object, the gradient waveforms must be transformed by $\mathbf{A}(t)$, meaning that they must undergo a rotation, scaling, or shearing (all linear operations). This is possible with conventional linear gradients (nonlinear warping, on the other hand, would not be correctable). This review focuses on rigid objects such as the brain. In this case, it is assumed that only rigid-body motion occurs. This includes translations and rotations but excludes scaling or shearing operations. Thus, Eq. [2] can be rewritten as:

$$\mathbf{g}'(t) = \mathbf{R}(t)\mathbf{g}(t), \qquad [3]$$

where the rotation matrix $\mathbf{R}(t)$ represents the rotation of the image object over time. Thus, to correct for rotations, the read, phase, and slice-encoding gradients (the logical gradients) are represented by different combinations of the x, y, and z gradients (the physical gradients) as the object rotates (Fig. 1b). Practical implementation details relating to the above are discussed in the following section.

METHODS

The Theory section described how prospective motion correction can maintain data consistency during scanning. Although the technique is conceptually simple, the difficulty often lies in the implementation details. Unlike many other MR methods, prospective motion correction involves real-time changes to the scanning process, which complicates the development and testing of the technique. This section describes successful implementations reported to date. The focus is placed on correction of head motion, but several other examples are mentioned.

Obtaining Tracking Data

For brain imaging and spectroscopy, various methods have been used to obtain the necessary head pose information. In this review, we classify these as optical methods, field detection methods, and navigator methods. This classification relates to three fundamentally different ways to obtain information describing the pose of the imaged object. Optical methods are completely independent from the MR sequence timing. Often, they use technology developed for a completely different purpose, such as motion capture for the movie industry. Field detection methods are based on a similar principle to spatial encoding, namely that different points in the scanner bore experience different magnetic fields. This principle can be used to encode the position of a marker. Navigator methods are a more traditional way of obtaining object pose, where the scanner is used for imaging, in one or more dimensions, and the resulting data are compared over time. The three methods defined above are described in more detail in the following paragraphs.

Optical methods include laser systems (14,15), bend-sensitive optical fibers (23), and camera systems. Camera systems have recently become popular, due to technology improvements in both cameras and computing. Methods successfully used for motion correction include out-of-bore stereo camera systems (18,24), out-of-bore single camera systems (25), in-bore single camera systems (26,27), and in-bore systems with multiple cameras (28,29). Generally, the optical approach began with out-of-bore camera systems [e.g., Zaitsev et al. (18)], but it has now moved to in-bore solutions. Cameras situated out of the MR scanner bore were useful for proof-of-concept studies, due to the lower requirements for MR compatibility, compared to in-bore cameras. However, they require optical line-of-sight and extremely high mechanical stability when the tracking marker is located several meters away from the camera. These practical considerations mean that in-bore tracking options are likely to be the better long-term solution.

All currently used optical systems require a marker. Although "markerless" head tracking would be ideal from a patient handling perspective, sufficient accuracy and speed have not yet been demonstrated. Examples of markers include reflective spheres (18), variations on traditional computer vision approaches [e.g., the "self-encoded" marker of Forman et al. (30)], or new technology such as moiré phase tracking (31), which generates moiré patterns allowing accurate determination of through-plane rotations [previously known as the retrograte reflector (25,32)]. The last example allows the use of particularly small targets (diameter 1.2 cm or less) with a single camera and has been shown to be a suitable alternative for conventional three-dimensional (3D) motion capture (33). Of course, in all of these examples, a marker must be rigidly attached to the head. This issue is discussed later in this review in greater detail.

Field detection methods are a completely different approach with a long history in MRI. The scanner gradient fields are measured to localize the object. The method requires the use of a short sequences of pulses to obtain position information from a small sample of MR-visible material fixed inside a miniature receive coil. This approach was first conceived in 1986, by Ackerman et al. (34) for catheter tracking. Dumoulin et al. (35) also pioneered developments in this area. A proof-of-principle study for slice-by-slice prospective motion correction using such a system was published by Derbyshire et al. in 1998 (16). More recent implementations, such as that of Ooi et al. (36,37), refer to these as "active markers." Active markers have been used for prospective motion correction in structural brain scans (36) and in echo-planar imaging (EPI) (37). A similar technique has been recently applied to measure gradient waveforms by Barmet et al. (38,39), who decouple tracking from MR imaging by using RF-shielded probes and separate transmit/receive chains. Recently, they have also demonstrated the possibility of computing the probe position during simultaneous MR imaging by applying "tones" [10 and 13 kHz in Brunner et al. (40)] simultaneously with the conventional gradient waveforms. However, this approach notably perturbs the k-space trajectory, which has to be accounted for in image reconstruction.

Field detection methods require several probes or active markers to be attached to the subject (a minimum of three markers are required, positioned noncollinearly and connected in a rigid arrangement). In Refs. 36 and 37, marker fixation is achieved by attaching the coils to a headband worn by the subject. There is a slight disadvantage over optical methods here, as the active

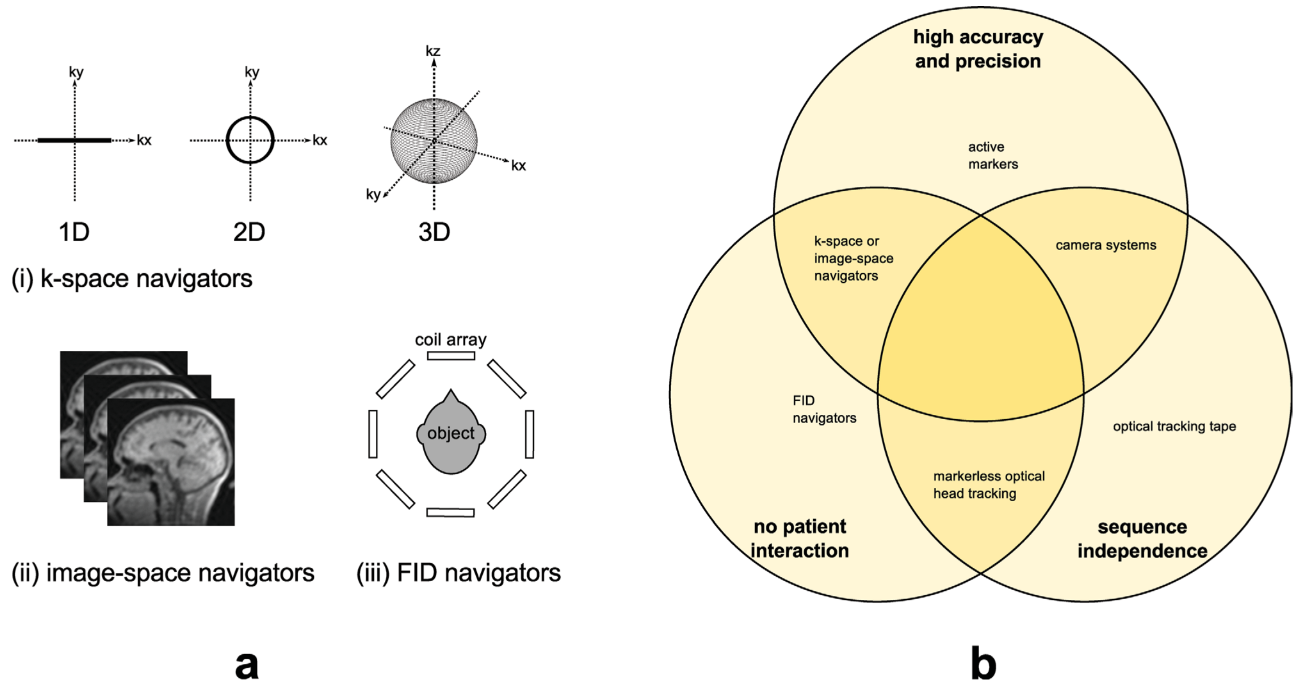

FIG. 2. Methods to obtain head pose information for prospective motion correction. **a**: Navigator methods use data from the MR scanner, rather than from an external source. Navigators can operate in both k-space (i) and image-space (ii). In both cases, data are repeatedly sampled and then compared between different time points to compute motion parameters. Alternatively, the relative change in signal intensity from multiple coils can be used to detect motion with a simple free induction decay (iii). **b**: Ideally, a method for use with prospective motion correction would meet all three requirements illustrated by the circles in the Venn diagram. Such a technique has not yet been devised, so the best choice currently depends on the imaging situation and the compromises that can be made in each case. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

markers (and hence the subject) are connected to the scanner by wires, which makes patient handling more difficult and could perhaps increase patient anxiety levels. The presence of cables also enhances the difficulties with the rigid marker fixation.

MR navigators are the traditional means of obtaining position information during MRI (Fig. 2a). Recent examples used for motion correction include navigators operating in k-space, such as cloverleaf navigators (41), orbital navigators (42,43), and spherical navigators (44) as well as image-based navigators, such as PROMO (45) or EPI navigators (46,47). k-Space navigators repeatedly sample parts of k-space and quantify rotations and translations of the object by measuring rotations and phase shifts in the k-space data. Depending on the trajectory used, this can allow motion quantification in all 6 degrees of freedom. Image-based navigators use low-resolution images or volumes. These generally require longer to acquire than k-space navigators but allow the user to define the region of interest for motion quantification, thus avoiding nonrigid regions (e.g., the neck). Alternatively, it is possible to detect, but not quantify, motion by comparing the relative intensity of a free induction decay signal between multiple receive coils (48). Navigator methods with sufficient accuracy for prospective motion correction all require unused time in the sequence to obtain accurate motion information [e.g., about 48 ms for PROMO (45)], which makes them incompatible with some sequences. This spoils one of the main advantages of prospective correction, namely that the technique can be applied to most MR sequences. Nevertheless, if time in the sequence is available, as is often the case in spectroscopy, this method is very practical. Navigator methods have an advantage over optical tracking and field detection methods, in that they require no additional hardware and that there is no need for a marker to be attached to the subject. This is particularly important in terms of patient handling in clinical MRI.

Figure 2b provides a rough guide to the strength and weaknesses of different systems. No current approach is perfect, and the best choice of method will depend on the relative importance of the three criteria indicated in the Venn diagram for each imaging situation. A quantitative comparison between the different systems discussed above is not provided here, due to the many different ways of measuring and quoting the parameters that describe their performance (such as accuracy).

Finally, it should be noted that the above description applies mainly to brain imaging. Other tracking systems, not mentioned above, can be applied to track other objects of interest. One example is ultrasound imaging to track the position of organs inside the abdomen (49,50).

Data Transfer and Transformation

Regardless of the tracking modality used (optical, field detection, or navigators), a key component of prospective correction involves the transfer of the pose estimation data to the imaging sequence. However, changing the tracking modality makes a significant difference to how this is performed. Navigator techniques, for example, often use the feedback facility made available by the scanner manufacturer; as this is vendor specific, it is not discussed further here.

In the case of external tracking systems, pose data are often computed on an external computer and sent to the scanner computer using a network connection. For this purpose, data are sent via user datagram protocol (UDP) [e.g., (18,25)] or transmission control protocol (TCP) [e.g., (26,49–51)]. UDP is perhaps better suited for real-time applications, as it puts more emphasis on timeliness than on reliability. Indeed, there is little purpose in resending missing packets, as a late packet will become redundant, due to new tracking information. The importance of minimizing latency is described later in this article.

For external tracking systems, the coordinates of the tracked object must be transformed into the coordinate system of the MR scanner. This process is trivial once the correct transform is known. Following the terminology introduced in Zaitsev et al. (18), we refer to the process of determining the transform as cross-calibration.

There are a number of ways in which this cross-calibration procedure can be performed. Aksoy et al. (27,30,52) use a 60 s cross-calibration procedure based on a precisely manufactured marker (30) that is visible to both the scanner and the camera. Two other common approaches involve recording motion of a phantom using both the tracking system and the MR scanner (using image registration). Depending on the exact implementation details, we call these approaches iterative or noniterative. The noniterative approach involves collecting numerous datasets and solving for the transform that best fits the data [e.g., as described by Kadashevich et al. (53)]. The iterative approach, as described in Ref. 18, applies prospective correction using the latest version of the transform. If the transform is accurate, the resulting images will be perfectly aligned, due to motion correction. If the transform is inaccurate, then errors in the image alignment will result; these are used to fine-tune the transform. In our experience, calibration based on image registration can produce very good results, but there are several confounding effects to be aware of. These include field distortions (caused by rotating the phantom during calibration), gradient nonlinearities, and imperfect fixation of the tracking marker to the phantom. These issues are similar to general limitations of prospective motion correction, which are discussed later in this review.

In Ref. 52, Aksoy et al. describe a hybrid prospective and retrospective correction method to mitigate the effect of cross-calibration errors. This involves retrospectively finding a transform by minimizing image entropy in a similar way to previous work by Atkinson et al. (54,55). As k-space lines are rotated off the Cartesian grid, a gridding reconstruction (56) is used to resample the data. Results show that application of the retrospective stage significantly improves image quality by reducing artifacts caused by poor cross-calibration.

Imaging Volume Update

To perform prospective motion correction of a moving object, the gradient and RF fields are adjusted so that the imaging volume follows the observed motion. This process is nothing more than the position update that is already applied at the start of imaging to set the position of the field of view (FOV). Details concerning adjusting this "on the fly" are specific to the scanner used. For the three main manufacturers, more information can be found in the literature, for example, Siemens in Zaitsev et al. (18), Philips in Manke et al. (20) and Ooi et al. (36), and GE in Qin et al. (26). These examples all describe "inter-view" correction, where adjustments are made between spin excitations. For MR sequences where the time between excitation and signal readout is short relative to the motion expected, a coordinate update prior to each excitation pulse is usually sufficient. However, when additional signal encoding such as diffusion weighing is used, a more sophisticated correction scheme can make sense, as motion during the strong and enduring diffusion gradients leads to severe motion artifacts and signal dropouts. This "intraview correction," was suggested, but not implemented, by Nehrke and Börnert in Ref. 22. Recently, however, Herbst et al. (57) have demonstrated a practical implementation of intraview correction on a Siemens system and have shown it to prevent signal dropouts in DWI.

Peripheral nerve stimulation and technical limits of the gradients mean that clinical MR scanners typically impose "hard limits" on the gradient strength and slew rate. This is relevant to the real-time imaging volume update, as these limits might be violated after transformation of the gradient waveform. Normally, such a violation would lead to an abortion of the pulse sequence, so having a mechanism in place to prevent this is essential. One approach is by incorporating an extra safety margin when specifying the initial maximum gradient strength (22).

An alternative to updating the imaging volume is data reacquisition when motion is detected. This approach has been actively pursued by Kober et al., who use their "free induction decay navigators" (48) to detect head motion exceeding a predefined threshold. Motion correction by data reacquisition increases scan time and cannot be considered to be "prospective motion correction." It also requires that the imaged object returns to its original position after motion, which is unlikely to be the case in brain imaging. However, a hybrid approach where detected motion triggers the acquisition of an extra EPI volume to quantify the motion parameters and apply prospective correction has also been developed (58). This appears to be a promising compromise.

Applications

Figure 3 shows a typical application of prospective motion correction. Data were collected at 1.5 T with a 3D gradient echo sequence, modified by the authors to allow prospective motion correction. Initially, the subject was asked to remain as still as possible. Then the subject was instructed move between two predefined positions, whenever prompted to by the scanner operator. In both cases, the subject was imaged twice: once without and once with prospective motion correction. Prospective motion correction improves image quality under motion

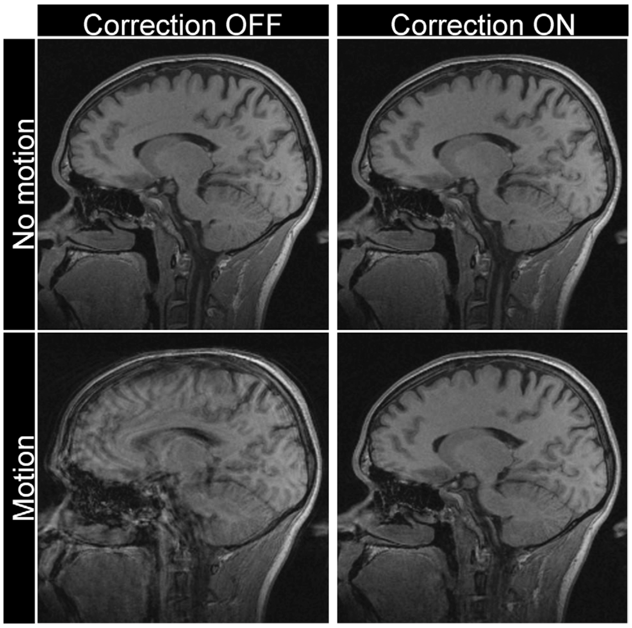

FIG. 3. Prospective correction results obtained at 1.5 T as a demonstration for this article. Head motion was quantified in real time using the single-camera moiré phase tracking system reported in Ref. 31. In (**a**) and (**b**), the subject did not perform any deliberate motion; in (**c**) and (**d**), the subject performed a series of repeatable head movements (predominantly left-right rotations). Prospective correction is off in (a) and (c) and on in (b) and (d). The motion occurring during each of the four scans in (a)–(d) is shown in Fig. 4.

and maintains it in the "no motion" case. Figure 4 shows the motion that occurred in each of the corresponding experiments in Fig. 3. The careful examination of tracking data is important in the evaluation of prospective motion correction techniques, as it can be difficult to ensure that motion is consistent between imaging experiments.

Figure 5 demonstrates the application of intraview (or "continuous") prospective correction in DWI, where additional correction updates are applied during the diffusion-encoding gradients [Herbst et al. (57)]. This technique adapts all gradients to motion occurring during the encoding process (Fig. 5a). Figure 5b shows results obtained using the method. The continuous correction of the diffusion gradients and the refocusing pulses significantly reduces motion artifacts, and signal dropouts are prevented even when strong motion occurs. Intraview correction might also have application in sequences with long echo trains, such as fast spin echo [RARE (59)] and related techniques, such as SPACE(Siemens)/CUBE(GE) or HASTE, as described in Ref. 60.

Many promising in vivo results involving brain imaging and spectroscopy of volunteers have been published in the last three years. In proton magnetic resonance spectroscopy, prospective motion correction has been shown to reduce spectral artifacts and quantitation errors in choline/creatine ratios (25) and reduce lipid contamination and increase spectral reproducibility (61). It has also been combined with interleaved reference scans (Ref. 62) to correct for both motion and motion-induced B_0 offsets (63), and, in the case of Hess et al. (64,65), combined with first-order shim correction. A similar methodology to that used in Ref. 63 has been applied to spectroscopic imaging (chemical shift imaging), where it can prevent data degradation under motion (66). Prospective motion correction has also been shown to be potentially useful in fMRI (24,37) and diffusion tensor imaging (DTI) (27).

Volunteer studies with prospective motion correction are becoming increasingly common. One implementation of prospective motion correction, PROMO (45), has been evaluated in healthy children (2,67), where it was shown to improve the quality of 3D anatomical imaging. However, little research has been performed in patient populations. Even less validation has been performed with prospective correction using an external tracking system. This translational step is an obvious goal for future work.

ADVANTAGES

Prospective motion correction has several advantages over methods that first collect k-space data and then retrospectively correct for the effects of motion.

Flexibility

Assuming an external tracking system is used, prospective motion correction is a general approach that can be applied to most imaging and spectroscopy sequences. Minimal changes to the pulse sequence are required, which is important for reducing development effort and for maintaining full pulse-sequence design flexibility. Field detection and navigator-based methods are not so flexible but are still applicable to many more imaging situations than specific retrospective methods.

Prospective motion correction is valid at all current field strengths and on clinical scanners from most major vendors (a prerequisite is that it must be possible to update gradients and RF during sequence execution in response to data from an external source).

Ensure Data Consistency and Adequate Sampling

In the presence of large rotations, retrospective correction does not guarantee sufficiently dense sampling of k-space. Prospective correction, on the other hand, avoids Nyquist violations if rotations occur (Fig. 6). A further example is when the object moves out of the imaging volume in the slice-encode direction: in conventional imaging, unrecoverable data loss occurs; in prospective correction, the FOV follows the object, avoiding such loss. This allows the FOV to be prescribed tightly around the imaged object. In both of these cases, prospective correction ensures the quality of the collected data; this is a fundamentally different approach to retrospective correction, which attempts to "reduce the damage" after it has already occurred. It is also important to note that prospective and retrospective correction approaches are not incompatible and in many cases could complement each other (52,68,69).

In the case of DWI, motion can cause signal dropouts that cannot be corrected retrospectively. These can be

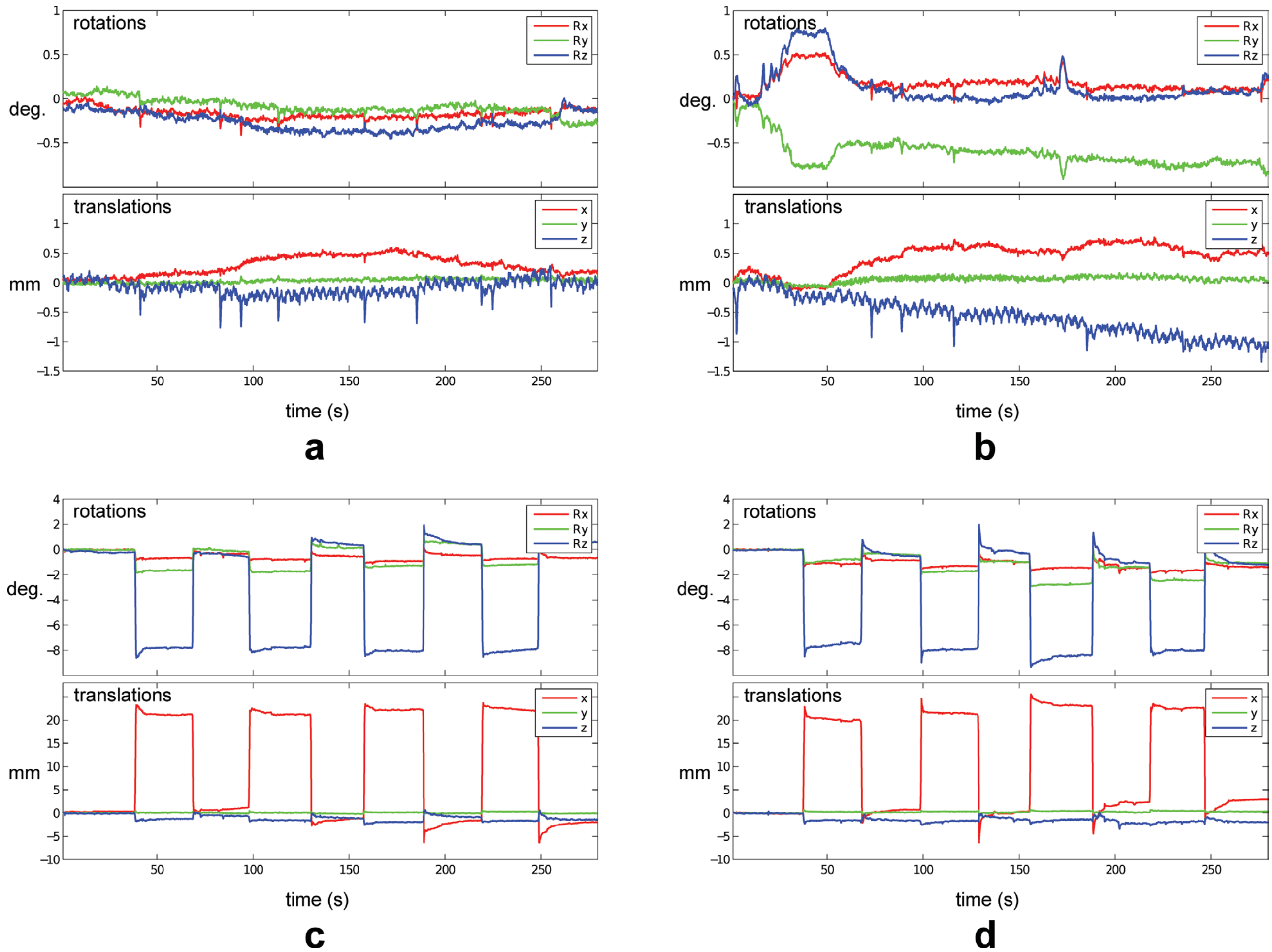

FIG. 4. The motion that occurred during the acquisition of the corresponding images is shown in Fig. 3. In (**c**) and (**d**), the scanner operator instructed the subject to rotate his head at particular points in the scan; this process ensured repeatable motion. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

avoided altogether using prospective correction (as shown in Fig. 5b). In the case of DTI (70), it is important to account for the diffusion-encoding direction when correcting for motion (71). This is done automatically with prospective correction, as the encoding direction is kept constant relative to the object. In flow imaging, a similar principle applies, as prospective correction prevents a change in flow direction relative to the gradients. In EPI, the direction of distortions stays constant relative to the object, which reduces misregistration and simplifies distortion correction.

In spectroscopy, it is important to note that motion artifacts are not necessarily easy to spot in the spectra. Thus, prospective correction is important, as the operator may otherwise not realize that the data have been corrupted and that the results are erroneous. Another benefit for spectroscopy, as well as imaging sequences, is the ability to apply an interscan motion correction [so-called position lock (66)], to align consecutive scans with possibly different contrasts, imaging volumes, and coverage. This allows the scanner operator to specify spatial saturation bands, or tightly define the imaging FOV, without fear that the subject has moved in the time since the scout data were acquired. This is similar to the "auto-align" functionality now provided by several major scanner vendors.

Spin-History Effects

Prospective correction compensates automatically for spin-history effects, which occur when tissue moves in and out of the slice being imaged (72). Tissue magnetization within a slice normally reaches a steady state after a few pulse repetition time periods but through-plane motion disrupts this. If spins with a different excitation history, and therefore a different magnetization, end up in the imaging plane, this can cause signal fluctuations. This affects mainly two-dimensional imaging, which is prevalent in routine clinical applications. It is reasonably clear that the spin-history effect will occur, and that it can be prevented by prospective motion correction, so most authors have simply referred to it as an advantage of prospective motion correction with little quantitative analysis.

However, recently Yancey et al. analyzed the spin-history effect in fMRI, with and without prospective motion correction (28). They performed Bloch equation simulations and in vivo experiments, where motion was largely

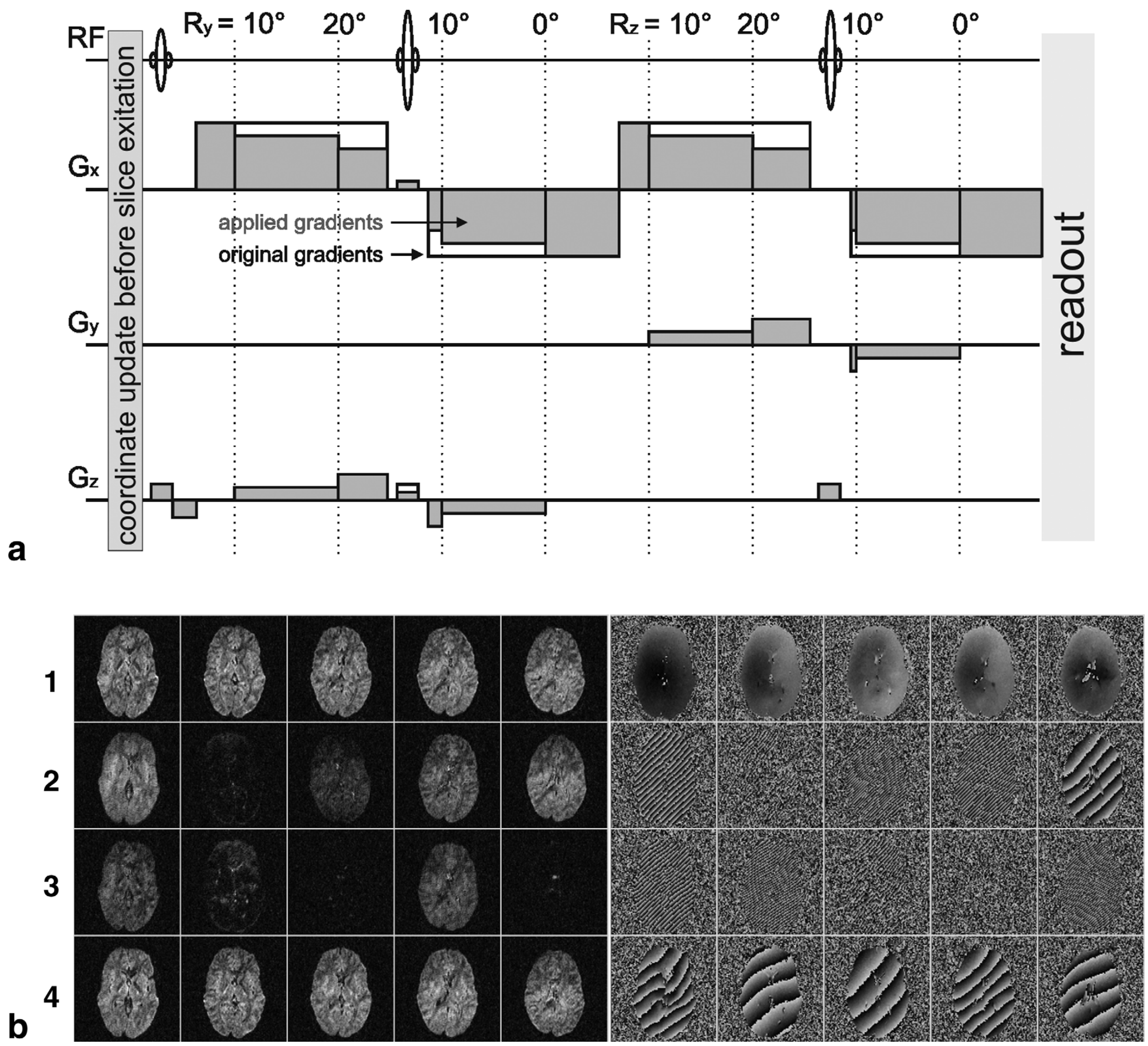

FIG. 5. Prospective motion correction can prevent signal dropouts in DWI. **a:** The "continuous correction approach" described by Herbst et al. (57), where correction updates are applied during the diffusion-encoding gradients. **b:** Results from four in vivo experiments showing magnitude images on the left and the respective phase images on the right. Row 1: no motion and no correction. Row 2: strong motion and no correction. Row 3: strong motion and slice-by-slice prospective motion correction. Row 4: continuous correction. These results are reproduced from Ref. 57.

through-slice (achieved through a specially designed task involving ankle dorsiflexion (moving the ankle joint so that the toes move toward the head) and was correlated to the fMRI paradigm. Although they restricted their analysis and experiments to single-slice imaging, where spin-history artifacts are the worst, their work showed that prospective correction was an effective means of eliminating the spin-history artifact and pushed the artifacts to below the noise level. Spin-history effects are likely to become more important at high fields, as T_1 increases with field strength.

Images Available Instantly

Real-time fMRI is a new application of fMRI that requires processing of activation maps to be performed online (73). Prospective motion correction would be advantageous here, as retrospective methods may cause an unwanted delay. For structural imaging, instant image availability is also an advantage of prospective correction, compared to computationally intensive retrospective correction approaches. For example, spiral projection imaging is a recently proposed 3D acquisition scheme that allows estimation and correction of motion in six degrees of freedom (74). In some imaging situations, this could be regarded as a competing technique to prospective correction; however, estimation of motion parameters currently requires around 30 min, which is a disadvantage.

CHALLENGES

Many practical challenges need to be addressed before general-purpose prospective motion correction will be

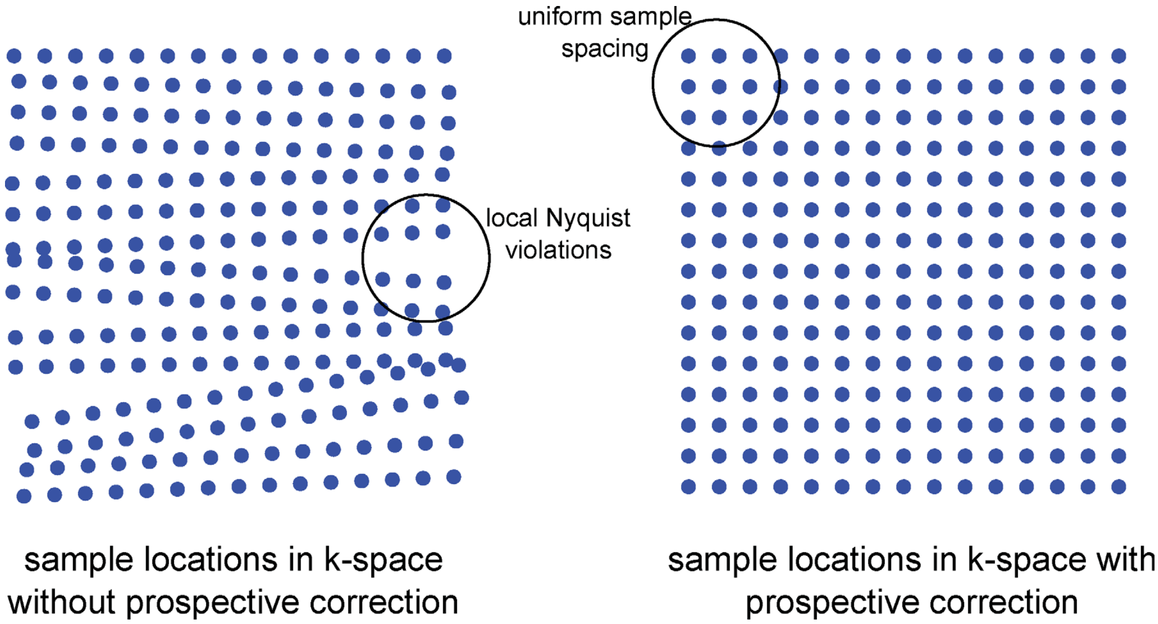

FIG. 6. Unlike retrospective techniques, prospective motion correction maintains uniform sampling in k-space even if rotations occur. This avoids "gaps" in k-space, which are not easily recoverable retrospectively. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

suitable for regular clinical use. The development of reliable and convenient solutions to these issues is a current area of research. Prospective motion correction also has a number of limitations. For many of these examples, it is unclear exactly how severely they will affect the limits of the technique; this remains an area for future work. Note that many of the challenges listed here also apply to retrospective correction techniques.

Quality of Tracking Data

In assessing the quality of tracking data, several parameters are important. In the MR literature, the terminology for these parameters is used somewhat loosely. Here, we use "precision" to refer to the level of jitter, or noise, in the data; "accuracy" to refer to the discrepancy between the measured and true position; and "latency" to describe the time delay (or lag) between the motion in question and the arrival of the corresponding tracking data on the scanner (to enable a position update to be performed).

For high-resolution imaging, the accuracy and precision of tracking are critical (75). If a tracking system is too noisy, the resulting "pseudo motion" causes artifacts (Fig. 7a). Although these can be retrospectively corrected to some extent (68) (Fig. 7b,c), the problem is best avoided by using a tracking system of sufficient precision. Reference 75 quantifies the precision required for translations if artifacts are to remain below the noise floor. The result is dependent on the imaged object and

FIG. 7. The accuracy and precision of tracking data are critical for effective prospective motion correction. **a**: Prospective motion correction using a tracking system that is noisy in 1 degree of freedom. The image contains motion-like artifacts, as predicted in Ref. 75. **b**: After retrospective correction using the method described in Ref. 68, the ghosting artifacts are reduced. **c**: The difference between (a) and (b).

the signal-to-noise ratio in the image, but, in general, tracking precision must be several times better than the image resolution. For very high-resolution imaging at high field, this could mean tracking precision needs to be better than 50 μm, depending on the level of artifacts than can be tolerated. For more conventional imaging at lower resolution, tracking precision can be relaxed to perhaps 200 μm. In both cases, achieving such precision is a challenge with commercial optical systems. Scanner vibrations are a related problem to be aware of, as these can reduce the precision of in-bore tracking systems, particularly when using gradient-intensive sequences (e.g., DWI).

Low latency is important, as any delay in obtaining pose data means that the imaging volume will "lag behind" the true head position. The importance of this depends on the object velocity. Quantifying the acceptable lag for clinical imaging has not yet been done, although Yancey et al. (28) have performed some theoretical work for fMRI. In Ref. 52, Aksoy et al. state that the latency of their system is between 60 and 150 ms. For this reason, they reacquire k-space lines when the detected motion exceeds 1 or 1 mm during one pulse repetition time. A lower threshold of 0.3 and 0.3 mm was used in Ref. 18. This is a reasonable strategy, although data rejection and reacquisition are not always acceptable, for example, in fMRI experiments. It is likely that with faster tracking systems, higher sample rates, motion prediction (76), and more frequent sequence updates (57), data rejection will soon be unnecessary in all but the most extreme cases.

Marker Fixation

Any external tracking marker must be securely attached to the subject and must move rigidly with the object of interest such as the skull or brain. If the marker moves while the skull remains stationary, then the erroneous imaging volume updates applied can irreparably damage the image quality. A number of attachment methods have been applied to try and achieve a rigid coupling between the marker and the skull; however, there is currently a trade-off between the degree of rigid coupling and the level of convenience and comfort for the subject (Fig. 8).

Figure 8 inspires two suggestions for future work: the need to accurately quantify the performance of different marker attachment methods and the need for a method of marker attachment that could fill the place in the top right corner of the plot.

MR Compatibility

Safety is the most obvious MR-compatibility issue when using external hardware near an MR scanner. The hazards posed by ferromagnetic objects or wire loops are well known and are not discussed further here. However, there are many other effects that should also be considered. Any external device used to obtain head pose information should not interfere with the MR scanner. Particular care must be taken to ensure that magnetic field inhomogeneity and RF noise levels are not affected (for suggestions concerning how to test for such effects see, e.g., Refs. 77 and 78). Similarly, the device must function in strong static and RF magnetic fields.

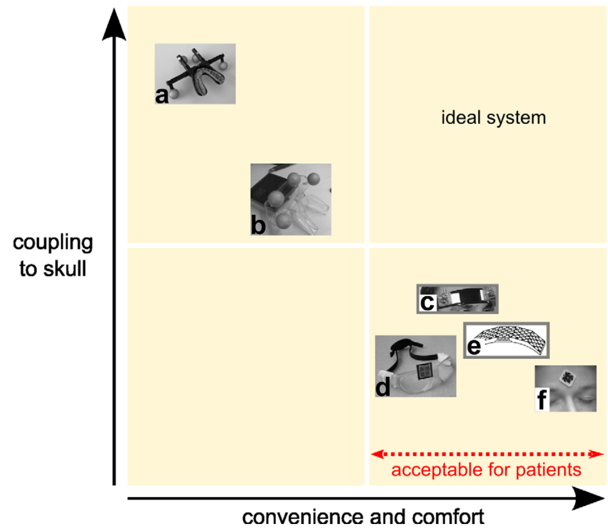

FIG. 8. A depiction of the relative performance of different marker attachment methods. The position of each method on the plot is purely subjective and is based on the authors' experience with a similar range of marker attachment approaches. The systems shown are (**a**) bite bar with dental impression (18), (**b**) sports mouth guard (photo courtesy of Brian Andrews-Shigaki), (**c**) head band (36), (**d**) plastic safety glasses (25), (**e**) the Stanford self-encoded marker, which is normally attached to the forehead (27), and (**f**) a 15 mm moiré phase tracking marker (31) attached to the forehead with double-side tape. All of the methods shown are capable of producing good results in specific imaging situations; however, the empty space at the top right of the plot indicates that there is still a need for a marker attachment solution that meets both criteria well. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

This requirement makes the use of most off-the-shelf equipment impractical. There are many other compatibility challenges to be met, including eddy currents, camera mounting, vibrations, and data transport. Space confinements in MRI make achieving line of sight to a target difficult for optical systems. For this reason, there is a trend toward in-bore systems; however, this tightens the requirements for MR compatibility.

Higher Order Motion

Higher order motion (e.g., velocity, rather than displacement) is typically not accounted for in traditional prospective motion correction implementations. An exception is the continuous update approach described by Herbst et al. (57) and illustrated in Fig. 5. The problem with higher order motion (and the reason for the signal dropouts in Fig. 5b, rows 2–3) is that motion during the application of a gradient will cause errors in the signal phase. This strongly depends on the sequence and has not been investigated thoroughly yet.

B_0 Inhomogeneities

Although the magnetic susceptibility of human tissue covers a small range and is relatively close to the susceptibility of air, the differences are sufficient to cause problematic inhomogeneities in the B_0 field. For a general discussion of magnetic susceptibility effects in MRI, we

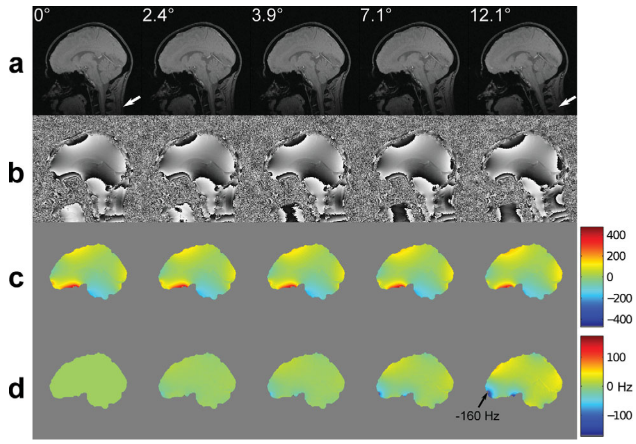

FIG. 9. A demonstration of magnetic susceptibility effects on the B_0 field in the brain during motion. In this example, the subject rotated his head backward in four steps. Prospective motion correction was applied while imaging with a modified field mapping sequence: (**a**) gradient echo magnitude images showing that the brain and skull remain aligned, due to prospective motion correction, while the neck deforms slightly, due to nonrigid effects (see white arrows); (**b**) the corresponding field maps, obtained directly from the scanner; (**c**) field maps after masking and unwrapping; (**d**) difference images between each field map in (c) and the field map obtained in the 0° reference position.

recommend Schenck (79). Here, we briefly describe the relevance of such effects for prospective motion correction.

Magnetic susceptibility boundaries in the body produce distortions in the B_0 field. This is particularly the case near tissue–air interfaces, where the change in susceptibility is the strongest. In prospective motion correction, the resulting field distortions mean that although the imaging volume follows the imaged object closely, the B_0 field at a particular point in the body will change as the object moves. It is also important to note that a related issue applies to the shim fields used to partially compensate the B_0 distortions when the object is in its original position. As these shim fields are not typically updated in prospective motion correction, the field "correction" will appear to move in the opposite direction to the object motion, compounding the problem.

Figure 9 shows results from an experiment performed to illustrate this problem. We modified a vendor-supplied field mapping sequence to allow prospective correction of motion during imaging. Field mapping was then performed on a volunteer, who moved his head four times during imaging, leading to data from five distinct poses. The imaging protocol included a pause of 2 s between each of five acquisitions, allowing the subject time to move; thus, each acquisition was acquired in a clearly defined position and not in a mixture of positions. Figure 9a shows magnitude images for the identical slice from each of the five acquisitions. As prospective motion correction was applied, all images are well aligned (at least within the brain: the neck region is distorted, as indicated by the white arrows). Figure 9b shows raw field maps, as reconstructed on the scanner. Figure 9c and d shows reconstructed field maps after unwrapping and masking and the difference to the reference. The maximum difference is 160 Hz, which is equivalent to 1.3 ppm (the B_0 field strength was 2.89 T on the Siemens Tim Trio used for this experiment).

It is worth noting that the "nodding" rotations applied were chosen specifically to emphasis the effect, as the orientation of susceptibility boundaries relative to the B_0 direction is the critical factor. Translations and in-plane rotations do not change the orientation of the object relative to the B_0 field direction. Unfortunately, it appears that such nodding motions are common (3). A similar experiment, although without prospective motion correction, has also been performed by Hess et al., who have investigated this effect for both single voxel spectroscopy (64) and spectroscopic imaging (MRSI) (65).

How important is this issue in practice? In an EPI time series, images may be distorted relative to each other, due to such changes (37,80). In both single voxel spectroscopy and spectroscopic imaging, results show that motion correction alone, without shim correction, is not enough to recover the spectral quality (64,65). In structural imaging, susceptibility induced distortions are a potential source of image artifacts, as k-space data acquired at different times might be inconsistent; however, the significance of this is unclear and is worthy of future study.

Several sequence-specific correction methods have been proposed. For EPI, Ooi et al. (37) present an unwarping algorithm that removes the effect of motion-induced B_0 distortions on EPI data. The method relies on the fact that the relative orientation of the head and the phase-encode direction stays constant under prospective motion correction; hence, the distortion correction problem simplifies to one dimension. This appears to be effective, but it is limited to EPI. For spectroscopy, Hess et al. use EPI volume navigators with resolution of $8 \times 8 \times 8$ mm^3 (64), which allow correction of both motion and field distortions. This method, however, is only applicable when the sequence timing allows the use of such navigators.

There is some hope that a general solution might be found. Rapid methods to predict (69) or to dynamically measure and/or correct (41,81,82) low-order B_0 distortions have been proposed. Combining methods like these with the prospective motion correction approaches described in this review could solve the issue for most sequences. However, much is still unclear, including the severity of the problem and the shim order that needs to be corrected.

Gradient Imperfections

Although gradient hardware has improved greatly since the early days of MRI, the actual gradient shapes played out on a standard clinical scanner deviate to some extent from ideal linear gradients, particular near the edges of the FOV. This effect is generally referred to as gradient nonlinearities. The result is that apparent deformations of the object may occur. These are often observed as warping near the edge of the FOV. As these distortions change with the position of the object in the FOV, it is likely that residual artifacts will occur after prospective

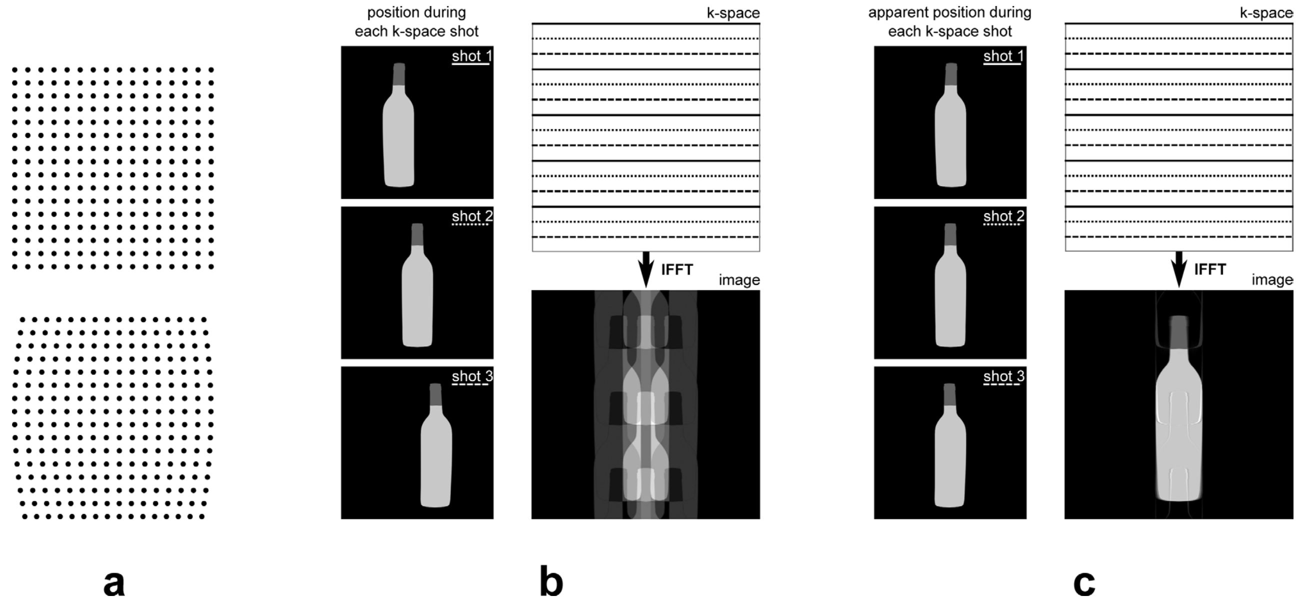

FIG. 10. **a:** Gradient nonlinearities result in a deviation of points from their ideal location in the image to a new location. Imaging of an object moving in the distorted field shown is simulated for a three-shot interleaved sequence: **b:** results without prospective motion correction, indicating the three poses of the bottle and the final reconstructed image; **c:** results with prospective motion correction. The image in (b) is totally corrupted, due mainly to the large amount of motion simulated; the image in (c) is mostly corrected, but residual ghosting remains, due to the remaining inconsistencies in k-space. These inconsistencies are caused by changes in the apparent shape of the object caused by the gradient nonlinearities.

correction, due to inconsistencies in the apparent shape of the object.

The exact effect of gradient nonlinearities on prospective motion correction has not been quantified, theoretically or experimentally. This is an important area for future work. Here, however, we present a simple simulation to demonstrate the effect.

Figure 10a shows the effect of simulated nonlinearities on the location of points in a square FOV. In our simulation, we assume that the object shown in Fig. 10b and c moves between three positions during the scan, and that k-space is filled in three shots. Each shot corresponds to one of the three object positions shown. Figure 10b demonstrates this process without prospective motion correction: the image reconstructed from the k-space data is completely corrupted by motion artifacts. Figure 10c shows results of the same simulation performed with prospective correction. Although the apparent position of the bottle is maintained during imaging, its shape changes with time, due to the change in its true position in the gradient fields. This results in residual errors in the reconstructed image.

Concomitant gradients (also known as Maxwell terms) may also be a relevant confounding factor for prospective correction. They have a spatial dependence, so a data consistency problem similar to that described for gradient nonlinearities might occur. Concomitant gradients also depend on the mixture of physical gradients being played out, so if a logical gradient is first represented by a single physical gradient, but then later by two, or all three, due to a rotation, then data inconsistencies could result. The importance of this effect will increase on high amplitude gradient systems or on low field scanners where the gradient magnitude is greater relative to the B_0 magnitude.

Nyquist Ghost Correction in EPI

Correcting for Nyquist ghosting in EPI when using a mixture of gradients for the readout direction may be problematic. This is due to a change in gradients (and gradient delays) used for the readout direction compared to when the scan adjustments were performed (see Fig. 1b). Speck et al. (24) demonstrated that this effect occurs, but noted that it is relevant for large motion amplitudes only. Thus, it is only likely to become an issue if large amounts of motion are part of the experimental paradigm (in fMRI) or for extremely uncooperative subjects in clinical imaging. A careful analysis of this problem is still needed.

B_1 Sensitivity Profiles

There is a trend toward using receive coils with higher numbers of channels together with parallel imaging to accelerate the image acquisition. This is potentially an issue for prospective motion correction, as with prospective correction, the coil sensitivity profiles will move relative to the apparently stationary object. Although we have not yet observed artifacts caused by this phenomenon in our experiments, the effect will likely become more pronounced when imaging with a high number of small coils, because each coil has a more localized sensitivity profile. It is possible that a retrospective correction step, such as that described by Bammer et al. (83), may need to be applied after prospective correction. At high fields, spatial variation of the transmit (B_1^+) field will further complicate matters (84).

The Rigid-Body Assumption

Given solutions to the other issues mentioned, it is possible that the assumption of rigid-body motion will become a limiting factor for prospective motion correction during brain imaging. As yet, the importance of this limitation is unclear, and this is an area that would benefit from future work. Work using displacement-encoded MRI (DENSE) in the brain has demonstrated that nonrigid pulsation of about 100 μm occurs in central brain regions, such as the thalamus (85).

If residual nonrigid motion in the brain proves to be large enough to cause visible artifacts (only likely at extremely high resolutions or in motion-sensitive sequences such as multishot DWI), then applying a nonrigid prospective correction strategy could be a good approach. A calibration scan could be used to construct a 3D affine motion model, and affine prospective correction could be applied, as was done by Manke et al. for coronary MR angiography (20). It would need to be determined whether the model needs to be created for each subject, as was done in Ref. 20, or whether a subject-independent calibration would be sufficient. Similarly, it is unclear to what extent an affine model (rotation, translation, shearing, and scaling) would approximate the true underlying motion. In addition, nonrigid prospective correction would help remove artifacts locally but may cause problems in other parts of the imaging volume. This is related to the "global correction problem" described below.

The Global Correction Problem

Another issue related to the rigid-body assumption is global versus local correction. In prospective motion correction, a gradient update applies a global correction to the entire imaging volume. In the case of a rigid phantom, this correction is ideal, but in all in vivo situations, this means that "apparent motion" will occur outside of the region of interest. The effect is visible in Fig. 9a, where the neck of the volunteer appears to move between the acquisitions, whereas it is the head that has moved, while the base of the neck was stationary for the duration of the experiment. If this occurs during a single acquisition, it may cause artifacts. Nehrke and Börnert (22) show an example where the spine becomes blurred after correcting for kidney motion.

Validation of Results

Validation of retrospective motion correction methods is relatively straightforward, as a direct comparison between corrected and uncorrected images can be made. However, after a scan using prospective motion correction, no uncorrected reference image exists. This is problematic in terms of the adoption and acceptance of the technique, because it will be difficult to show in patient studies (where controlled, repeatable motion cannot be used) that the technique is truly helpful. It is likely that statistical validation methods must be used, with large numbers of experiments. An alternative is the approach described in Ref. 57, where motion data from a prospectively corrected scan is saved and "replayed" in a second experiment on a phantom or stationary volunteer. This can provide a rough indication of what the first scan may have looked like without motion correction.

A related, and ongoing, problem in MRI is the assessment of image quality. Several groups have proposed methods to automatically quantify the level of motion artifacts in images (86–88), but this needs further work. The mismatch between what computer algorithms, MR scientists, and clinicians consider to be a good image makes automated image assessment difficult.

DISCUSSION

In its current state, prospective motion correction is immediately applicable to research at ultrahigh field, where high-resolution imaging is done on normal volunteers and issues such as the convenience of marker attachment are not as important. We can expect to see prospective motion correction used as tool to allow improvement in effective image resolution. It is likely that field strengths will soon move beyond 7 T, at least for research (89). For imaging at resolutions of a few hundred micrometers or better, effects of tiny, involuntary movements must be corrected.

How can we move forward with clinical applications? The value of prospective motion correction has now been well demonstrated in volunteer studies. The next step is validation in patient populations. For methods involving external hardware (e.g., optical systems and field probes), this has been problematic until now, as current marker fixation methods are generally not "patient compatible." Solving this final issue and performing patient studies must be done without delay. For MR-navigator-based techniques, such as PROMO (45), the transition to regular clinical imaging is likely to be much faster, although this will only apply to sequences compatible with MR navigators.

Should the method become generally accepted, there is a need for manufacturer support, particularly with regards to the integration of external tracking systems. External systems have the advantage of operating independently from the MR system, which should also mean that such systems could be used on all scanners. The standardization of communication protocols and systems for measuring the performance of different systems would greatly assist in quickly bringing the advantages of prospective motion correction to those patients who need it.

There are many possibilities for future research in addition to the application of prospective motion correction in patient populations mentioned above. Investigations into the relative importance of the various confounding effects identified in this article would provide a better indication where future developments are most needed. There is also plenty of work remaining on nonrigid motion, although we see this more as a niche application in brain imaging, given that most of the problem can be solved by a rigid correction.

Tracking system development for prospective motion correction has several potential spin-offs. One is retrospective motion correction (e.g., Refs. 90 and 91), which has disadvantages over prospective correction, as mentioned earlier, but avoids pulse sequence modifications. Another potential spinoff is multimodality imaging. If markers are reproducibly attached and systems are calibrated, data combination between modalities should be trivial. In addition, the field of therapy planning could benefit dramatically (e.g., surgery planning and radiation therapy). Finally, MR-compatible tracking systems could find application in interventional MRI, where the need to localize medical instruments relative to the patient is well established (92).

In conclusion, the recent developments in prospective motion correction have demonstrated that the technique has the potential to help solve what is still a very real problem in MRI and magnetic resonance spectroscopy. Although numerous challenges still remain, the advantages of prospective correction over existing techniques suggest that the benefits justify the effort required to develop solutions and that prospective motion correction may become an essential tool for next-generation MRI.

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