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The future of motion artefacts correction in MRI

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Summary

Motion during the MR examination affects the degradation of the image in the form of „ghosting" or blurring and is one of the last major unresolved issues. The patient's movements can be of different scope, directions and orientation, making it difficult to find a solution from the early days of MR (Magnetic Resonance) development.

Today, there are three main strategies for combating motion artefacts: motion prevention, motion reduction and motion correction. Due to the lack of possibilities and exhausted ideas, most work is being done on the development of motion correction using MR navigators or prospective and retrospective methods.

Of the future techniques, the most promising are SMS (Simultaneous Multi-Slice) sequences and CS (Compressed Sensing) due to the resistance to movement and shorter recording time compared to conventional sequences.

There is currently no general solution for the artefact of motion, but the problem is approached according to a specific case by choosing the method with the least shortcomings and taking into account the type of movement and the sequence of selection. More research is needed in clinical settings given the complexity and diversity of the problem with the involvement of a multidisciplinary team.

Keywords: MRI, motion artefact, methods, future, sequences, correction

Introduction

Motion is the most common adverse event associated with magnetic resonance imaging (MRI) [1]. Movement during scanning leads to image degradation, unnecessary repetition of certain parts of the procedure (sequences), prolongs the imaging and negatively contributes to the diagnostic quality of the examination, and consequently to additional costs of the healthcare system [2].

This is primarily due to the longer time it takes to collect a sufficient amount of data to create an image in certain sequences compared to other clinical modalities. Imaging times have a greater range than most physiological movements, including involuntary movements, heart rate, respiration, intestinal peristalsis, pulse, blood flow and cerebrospinal fluid flow. As much as 17% of MR imaging contains some unwanted event, a quarter of which is motion, and this is even more common in children [1].

Currently, there are no indications that the problem of motion artefacts could be solved solely by progress in the configuration, construction and external equipment of the device itself. The potential for a significantly shorter time of scanning is limited by biological limitations: peripheral nerve stimulation limits gradient rate switching, specific absorption rate (SAR) limits the use of radiofrequency (RF) pulses for excitation and T1/T2 times limit TR (repetition time) or TE (time to echo) depending on the required contrast. The greatest value and probable solution in the future are provided by techniques and methods of prevention, reduction and correction of motion using prospective or retrospective methods [3].

Methods

There are many different types of movements and many possible mechanisms that can lead to data corruption. Therefore, various strategies have been developed to address these challenges. Below, we classify these methods into three different groups: motion prevention, artefact reduction, and motion correction (Table 1). Motion prevention is the most obvious way to suppress motion artefacts. Avoiding the move avoids the impact described at the beginning, making other more complex strategies unnecessary. This review defines artefact reduction as gain strategies to mitigate artefacts in the resulting image or replace them with artefacts that are less dramatic in

appearance compared to standard Cartesian acquisition methods. Motion correction, on the other hand, typically involves explicit estimation and compensation of movement. All three methods can be used in combination, and multiple tools can be used as complements.

Table 1. Motion artefact mitigation strategies

Prevencija micanja	Redukcija artefakta	Korekcija micanja
Training	Faster imaging	Navigators
Distraction	Insensitive sequences	Self-navigated trajectories
Feed and wrap (for babies)	Gradient moment nulling	Prospective correction
Foam restraints	Saturation bands	Retrospective correction
Sedation	Triggering and gating	
Head holders	Phase reordering	
Breathhold		

Source: https://www.ncbi.nlm.nih.gov/pmc/articles/PMC4517972/

Prospective Correction

Prospective motion correction is an intuitive approach that keeps the relative spatial position and orientation between scan coordinates and the object of interest constant. For head motion compensation, we assume a rigid body moving with 6 degrees of freedom. If we can measure the position and orientation of such objects in real time, we can adjust the receiver's magnetic gradient, RF pulse, frequency, and phase accordingly. Rolling an object requires rotating an encoded gradient; translation requires a change in transmit and receive frequency and phase [4].

For head motion, the head pose can be measured using MR navigators operating in image space (PROMO, PACE, etc.) [5-7] or k-space (cloverleaf, spherical, orbital navigators, etc.)[8-10]. Alternatively, external tracking devices such as stereo camera systems [11], miniature RF probes [12-14], in-bore camera systems [15-17], and ultrasound systems can be used [18].

The navigator should be compatible with the timing of the sequence, usually with a lower temporal sampling rate. External tracking, on the other hand, typically requires an MR-compatible design and accurate calibration of the scanner coordinate system. Additionally, most external approaches require a tracking device to be attached to the object, which is a concern for routine applications.

The topic of prospective motion correction has gained popularity in recent years, leading to numerous new applications such as fMRI [19,20], DWI [21,22], and spectroscopy [23-25]. Although it is a highly promising approach for neuroimaging, it has some limitations, such as practical considerations (e.g., marker fixation of external tracking systems) and uncorrectable effects (e.g., motion-induced B0 distortion) [26].

Retrospective Correction

The prospective modification technique aims to maintain data quality during acquisition, whereas the retrospective technique improves data consistency after the scanning by modifying the collected data or the reconstruction model. This can be done by incorporating precise knowledge of motion during scanning (e.g. using navigator data) or by iterative algorithms (e.g. optimising image entropy, gradient entropy, or other artefact measures) [27-29].

The basic idea of these methods is to undo motion-related changes to the MR data. Rigid-body motions are described according to Fourier theorems. Translation of the object leads to a phase ramp in the acquired k-space, and rotation of the object corresponds to rotation in k-space [30]. Translation can be corrected relatively easily by applying a phase change to the acquired data, whereas rotation correction requires the use of non-Cartesian reconstruction techniques with sophisticated algorithms [31-34] and is computationally intensive. As long as the acquired signal is corrupted but not lost (e.g., no signal dropout due to phase shift within voxels), rigid body motion can be corrected in 3D imaging. In 2D imaging, these correction methods are limited to in-plane motion, because in-plane motion introduces disparity that cannot be corrected during scanning.

Elastic movement remains a major challenge for both prospective and retrospective corrective approaches. In MRI of the abdomen and heart in particular, the complexity of the underlying motion limits current concepts to a combination of gating/triggering followed by correction of affine motion within the gating window [35,36].

Bidder et al. explained that by discarding motion-corrupted data and filling the resulting parts of k-space using parallel imaging techniques, the reconstructed image quality can be increased at the cost of SNR and possibly some residual blurring [37].

Future Techniques

Kinetic Sensing (Prospective Motion Correction)

With the help of a high-precision internal camera, head movements, especially translational and rotational changes, can be recorded with high precision (0.1 mm for translation and 0.1 degrees for rotation). Then, by continuously updating the field of view during scan acquisition, prospective motion correction in all axes for both translation and rotation can be achieved (both in-plane and through-plane motion). Such real-time prospective motion correction could offer significant advantages over previous approaches. The latter includes various k-space scanning schemes such as BLADE (aka PROPELLER and MultiVane) and radial imaging (a motion minimization and predictive correction technique), which have proven to improve the diagnostic quality of images and are currently in clinical use [38,39].

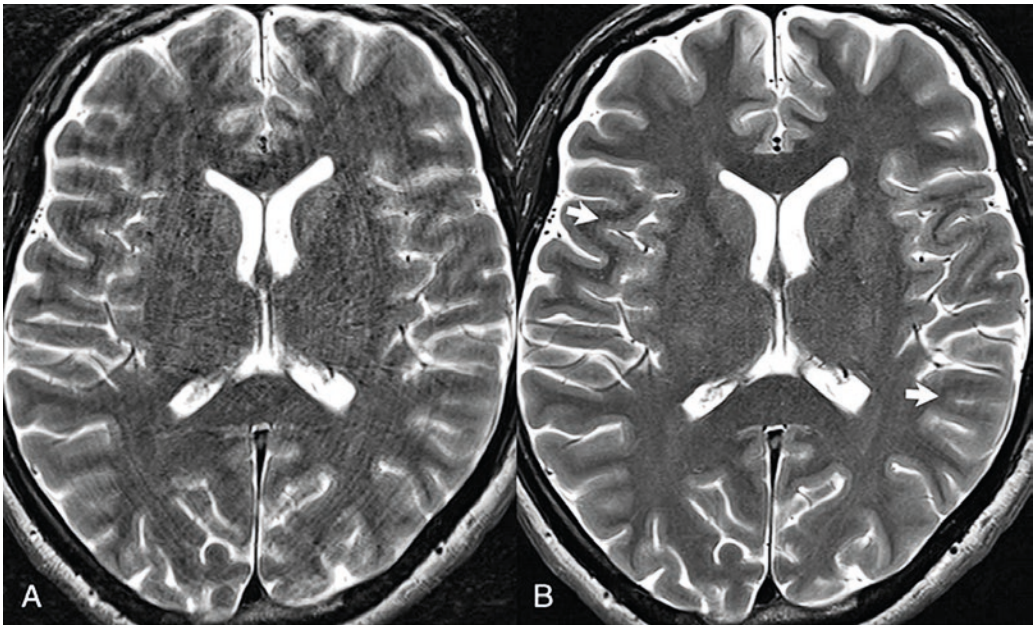

Figure 1. Image degradation (ghosting, blurring, loss of grey-white matter differentiation) due to mild motion in a normal volunteer, on (A) high-resolution T2-weighted fast spin echo imaging, with a marked improvement (B) using the kinetic sensor. Note the minimal residual ghosting (small white arrows) in the motion corrected image, superficial to the right insula and overlying the left parietal lobe.
Source: Runge VM, Richter JK, Heverhagen JT. Motion in Magnetic Resonance: New Paradigms for Improved Clinical Diagnosis. Invest Radiol. 2019 Jul;54(7):384.

Prospective motion correction using motion sensors currently requires the placement of markers on the bridge of the nose, which are then tracked by a camera system. This information forms the basis of motion compensation, where motion is detected and compensated in real time. Sensor tracking speed is important, and with current systems, he reaches speeds of 60 frames per second. Furthermore, unlike other techniques such as using navigator echoes, scan time and image contrast are not compromised. As expected from the theoretical basis, such future motion correction may have the greatest impact on high-resolution imaging. Such a system should work ideally for patients with mild movements (Figure 1). This is very common in clinical settings and patients with a wide range of motion, such as children and uncooperative patients. The 3 T (Tesla) should be the priority in regards to implementation-specific pulse sequences like 3D MP-RAGE (aka 3D-FGRE, 3D fast SPGR, and 3D-TFE) and two-dimensional (2D) Fast Spin Echo (FSE, including FLAIR). Other high-resolution head scans, such as 3D SPACE (also known as CUBE and VISTA) and time-of-flight MRA, are also expected to have clinical applications. 7 T is also likely to benefit from this approach, with early clinical evaluations still underway. However, due to the small number of clinical systems, the market impact of 7 T will be small in the beginning [40].

Respiratory Sensing

A major innovation in clinical MR is the ability to monitor the respiratory cycle using coils that are an integral part of the patient table, within the spine coil. Respiratory monitoring and the use of this signal for image acquisition and reconstruction has a long history in MRI and has been associated with long setup times and unreliable in clinical use [41]. Navigator echoes have their advantages and are now widely used, but they require operator setup (because they are prone to mispositioning) and time within the pulse sequence [42].

This new technological advance requires no set-up, is automated, and provides reliable and reproducible airway waveforms. An additional simple loop coil is incorporated into the spine coil located within the scanner's patient table. However, since this coil operates at a different frequency than the Larmor frequency (specifically a 30MHz transmit/receive coil), its changes can be detected separately from the image signal. Breathing causes a change in coil load, resulting in a change in current in the coil, which is detected and monitored (reflecting respiration). A small current is supplied to the coil as input and the resulting current output follows the shape of the respiratory cycle, with a slightly higher current during inspiration and a lower current during expiration [40].

In fact, in the current implementation of this technology, there are two such coils within the spine coil, as the coil for respiration detection must be placed on the patient's diaphragm. The coils are positioned so that they automatically align correctly for most people when they are head-first into the scanner, regardless of their height. Her second coil of this type is at the end of the spinal coil and can be used for feet-first examinations, including very large and small subjects and paediatric patients. Respiratory signal detection occurs whether the patient is inside or outside the bore and begins as soon as the patient lies over the spinal coil. In addition to being used in conjunction with scan acquisition (e.g., breath-triggered scans), respiration sensors are also important for patient monitoring and breath-hold scans to verify if a patient can hold their breath for a period of scan time .

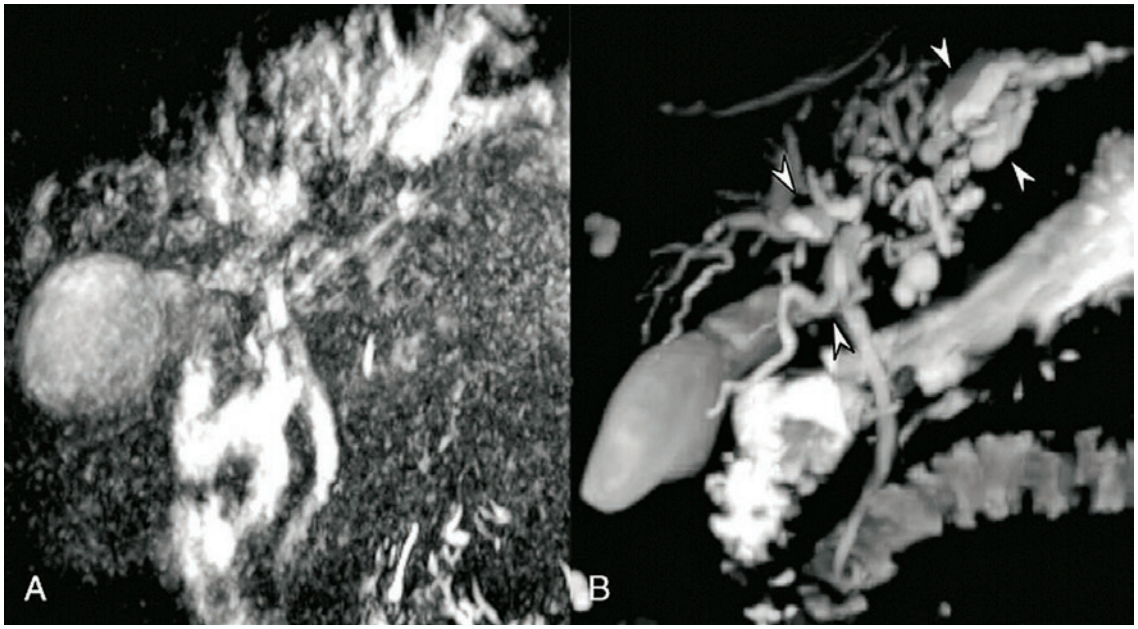

Figure 2. Three-dimensional MRCP scans acquired in a patient with a cholangiocarcinoma invading the portal hilum, comparison of (A) conventional and (B) breath-hold compressed sensing (CS) examinations. Scan times were 7 minutes on average (for the patient cohort evaluated) versus 16 seconds. The conventional examination is nondiagnostic due to motion. The breath-hold CS study well depicts the dilatation of the intrahepatic bile ducts (arrowheads)
Source: Runge VM, Richter JK, Heverhagen JT. Motion in Magnetic Resonance: New Paradigms for Improved Clinical Diagnosis. Invest Radiol. 2019 Jul;54(7):390.

Compressed Sensing

Although CS is still immature and has a wealth of undeveloped potential applications, it has already had a significant impact on motion management in MR. This was achieved primarily by reducing scan times and minimising the effects of involuntary movements, respiration and cardiac motion. For more demanding applications, scan reconstruction times are still long (on the order of minutes). Although convenient for clinical use, reconstruction times are less than 1 minute for most routine applications. Further improvements in this area mean that this problem will soon cease to be an issue as computing power continues to improve. Existing CS variants of VIBE, including GRASP-VIBE, XD-GRASP, CS-VIBE, and XD-VIBE serve mainly for liver imaging. It is also worth noting that post-contrast dynamic free-breathing scans can be used to assess liver perfusion without injecting additional contrast agents or increasing examination time. XD-GRASP provides motion-resolved free-breathing reconstructions, resulting in improved diagnostic quality of multiphase images after intravenous administration of gadolinium chelates [40].

Compressed sensing magnetic resonance cholangiopancreatography (CS MRCP) is available at both 1.5 T and 3 T, which can significantly reduce scan times [43-45]. Compressed sensing can be applied to conventional sequences (half to one-third of scan time). At 3 T, he is capable of an MRCP scan with one breath-hold (16 sec) with diagnostic quality (Fig. 2). Given the difficulty of obtaining MRCP scans without significant motion artefacts and the length of such examinations, it is likely that in the future his CS MRCP breath-hold examination will become the reference standard for this anatomical region.

CS is already having a major impact on cardiac imaging. Successful introduction of CS is demonstrated by two standard examinations, myocardial perfusion and CINE imaging. Single breath-hold 3D-CINE imaging of the left ventricle is possible with nearly isotropic resolution and can be reformatted in any orientation and without significant difference in ventricular function parameters compared to conventional 2D-CINE-based values (require multiple breath-holds) [46]. Acceleration has also been demonstrated in first-pass cardiac perfusion to quantify myocardial blood flow, allowing for greater spatial coverage and higher spatial resolution [47]. Whole-cardiac coronary MRA using CS is also advancing, with reduced scan time as a major goal [48].

A big problem in musculoskeletal imaging is artefacts caused by surgically placed metal. Slice Encoding for Metal Artefact Correction (SEMAC) has provided significant improvements, but long scan times have limited clinical implementation. Initial studies using the CS version showed a high-quality reduction of metallic artefacts, superior image quality to high-bandwidth FSE imaging, and diagnostic quality similar to conventional longer (non-accelerated) SEMAC. Over the past couple of years, as the software has evolved, clinically acceptable scan acquisition times of 2–6 min with a satisfactory time of reconstruction [40].

Beat Sensing

Next-generation cardiac motion sensors may enable automatic detection of the cardiac cycle, replacing the use of electrodes (electrocardiogram (ECG) monitoring), which are often difficult and time-consuming to place. ECG signals are also susceptible to artifacts caused by gradient activation and magnetohydrodynamic effects (especially at high magnetic field strengths). Coils with such sensors are already available, but sequences are still under

development. It is expected that such sensors will be fully implemented in the near future [40].

In the current concept, a local magnetic field generator is placed (embedded in the anterior coil) that produces a Pilot Tone (PT) which is modulated by changes in conductive geometry (particularly involving cardiac motion) [50]. This signal is picked up by a nearby MR local receive coil. The coil combination correlated well with the ECG and was shown to be stably detectable even during free breathing, allowing the acquisition of high-quality images of the heart. The frequencies used are just outside the MR signal band (64.4 MHz in the first work at 1.5 T). The PT design increases cardiac acquisition robustness concerning gradient and ECG lead placement. In the future, it may be possible to place flexible trigger time points. This single device, in full implementation, could enable both cardiac and respiratory synchronisation. This is because both movements affect the PT and can be observed. Although this implementation demonstrated that cardiac sequences (the acquired cine cardiac images from prospective cardiac triggering) in normal volunteers at 1.5 T, were indistinguishable from those acquired with ECG [49]. Pilot Tone prototypes have so far been implemented in both 1.5 and 3T systems.

Another potential advantage of PT navigation is that heart motion is measured directly (as opposed to observing the heart's electrical activity, which the ECG measures). Furthermore, unlike navigator echoes, PT navigation can be used in any scan sequence without the need to acquire additional navigator echoes [40].

Simultaneous Multislice (SMS)

In the SMS, multiple slices are excited simultaneously and reconstructed independently. The latter includes the use of blipping CAIPIRINHA applied during the echo train (to minimise SNR loss due to G-factor), followed by slice GRAPPA-based unaliasing, then GRAPPA-based in-plane unaliasing [50-52]. Implementations of SMS in 2D multislice diffusion-weighted (echo-planar) and fast spin-echo imaging are currently underway to mitigate motion artefacts (Fig. 3). Most clinical protocols currently use an acceleration factor of 2 (although an acceleration factor of 8 is easily reached in BOLD imaging protocols), resulting in ~40% reduction in scan time. Simultaneous excitation of multiple slices is achieved by using multiband pulses. Pulses in different bands (slices) are phase modulated and summed to form a multiband radio-frequency excitation pulse. Critical to this process is optimising the RF pulse in terms of energy deposition, the latter being a potential limitation of the SMS technique. For example, variable rate excitation (VERSE) can be applied to 180-degree refocusing pulses. Using blipped CAIPIRINHA is also very important as it can greatly reduce the disadvantage of the G-factor. The g-factor is position dependent and depends on the coil geometry along with many other variables. This happens because of the sensitivities of the individual coils that are very close. This can have a significant negative impact on SNR, for example in parallel imaging where the resulting SNR is g-factor dependent (inversely proportional).

In the clinical implementation of SMS, the use of blipped CAIPIRINHA essentially eliminates the drawbacks of the G factor. This is demonstrated, for example, by Setsompop from 32% to 1% for the 3-fold acceleration of diffusion imaging [53]. All-slice GRAPPA-based SMS implementations require the addition of a fast single-band reference scan for slice unfolding (during scan reconstruction), requiring additional time during the scan sequence. For SMS 2 this means that the scan time is not half that of a similar clinical scan without SMS. Another important factor is the potential for signal loss between non-aliased slices acquired at the same time, which can lead to artefacts in the final image. These are mitigated by using the leak block algorithm [54].

The difference between parallel imaging and SMS is important but not well understood. Parallel imaging reduces the number of acquired k-space lines by an acceleration factor R. Missing k-space lines are reconstructed using the coil sensitivity information. This causes an SNR loss proportional to the coil's g-factor and the square root of the acceleration factor. SMS imaging with blipped CAIPIRINHA results in minimal g-factor loss and no additional reduction of acquired k-space lines. Therefore, there is no SNR penalty proportional to the square root of the SMS factor. It is also important to note that SMS benefits from higher channel density coils but not always because higher channel density can cause artefacts of motion in this case. A typical 3 mm scan of the whole liver at 3 T took 4 min 46 s with standard ssSE-EPI but can be reduced to 2 min 54 s with SMS and an acceleration factor of 2. The impact of SMS on DWI of the prostate was recently demonstrated with a 5-minute screening protocol for prostate cancer in patients who have elevated levels of prostate-specific antigen, suggesting that the diagnostic performance of this test compared to a full multiparametric MR is similar but shortened [40].

Readout segmented EPI (rs-EPI, RESOLVE) was developed after ss-EPI and has the advantage of reducing geometric distortion and blurring that can be noticeable in 3T's ss-EPI. Improved image quality is achieved by acquisition of subsets of k-space over multiple TRs, rather than encoding the entire k-space in a single shot. Visualisation of areas prone to bulk-susceptible artefacts, such as the brain near air-filled sinuses, is also greatly improved [40]. An important limitation of rs-EPI is its long scan time compared to ss-EPI. Simultaneous multislice is now routinely used in combination with TR reduction for scan acceleration with rs-EPI. A significantly reduced scan time reduces the risk of poor scan quality due to unintended patient movement. Readout segmented EPI is also used for DWI optimised for acoustic noise (reduction of scan noise to increase patient acceptance). In this application, echo spacing is increased (thereby reducing gradient noise), resulting in higher image distortions and mass-sensitive artefacts (although much smaller than in SS-EPI). Areas of interest for the extracerebral application of rs-EPI are spinal cord, head and neck cancer, and breast.

Originally applied to DWI, the simultaneous multislice technique has now been extended to fast spin-echo imaging. Simultaneous multislice can apply all tissue contrasts including proton density, and T1 and T2 weighting among others. Anatomical areas where FSE SMS is applied include the brain, neck and especially the musculoskeletal system „knee, hip joint, foot/ankle, hand/wrist". Examples of proton density-weighted scans at 1.5 T in the knee and ankle without and with SMS are shown in Figure 16. Simultaneous multi-slice FSE is useful for protocols that require the use

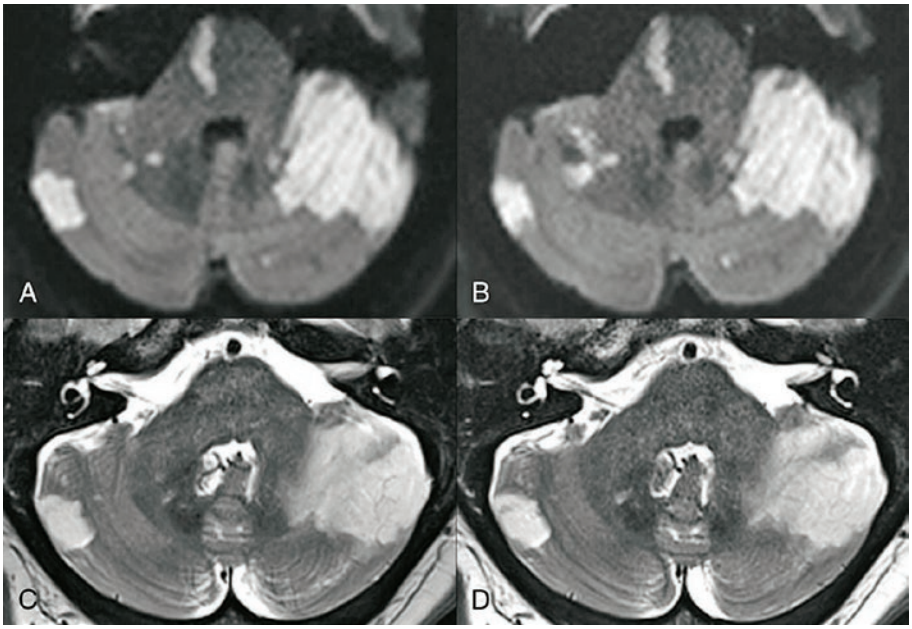

FIGURE 3. Multiple early subacute cerebellar infarcts, including one involving the right pons, visualized with high signal intensity on (A and B) DWI and (C and D) FSE T2-weighted scans—illustrating the utility of SMS. For the ss-echo-planar imaging (EPI) diffusion-weighted scans, the application of SMS (with 2 acceleration) reduced the scan time from(A) 1:23 to (B) 50 seconds. The decrease in scan time was achieved by reducing the TR from 6300 to 3500 milliseconds, with no loss in diagnostic quality. For the T2-weighted FSE comparison, use of a 3 acceleration combined with a reduction of the number of concatenations from (C) 3 to (D) 1 led to the scan time being reduced by a factor of 3
Source: Runge VM, Richter JK, Heverhagen JT. Motion in Magnetic Resonance: New Paradigms for Improved Clinical Diagnosis. Invest Radiol. 2019 Jul;54(7):391.

of multiple concatenations (to get the required number of slices), scan time reduction („managing" movement) or a large number of higher SNRs in the brain [40]. For diffusion-weighted (echo-planar) scans, SMS can be used to reduce TR while maintaining the number of slices compared to conventional scans. For example, in Figure 3 the TR is reduced from (A) 6300 ms to (B) 3500 ms. However, changing the TR can have implications. Depending on the clinical protocol, the minimum TR (determined by the number of slices) may not be achievable due to differences in SNR and CNR (contrast to noise ratio) [40].

Two-dimensional single-shot spin echo (ssSE) echo planar imaging (EPI) is widely used for diffusion-weighted imaging in both brain and body applications. In conventional acquisition, diffusion coding is performed over the entire imaging volume for each 2D slice excitation, making this scan very time inefficient [40].

Parallel imaging is typically not performed to speed up scanning, but rather to reduce EPI encoding time, thereby reducing image distortion and blurring. However, the reduced scanning time of ssSE-EPI to the brain is limited by SMS implementation. This is because scans are often very short, to begin with. For example, at 3T, the typical scan time for a 4-mm section of the whole brain is only 1 minute and 23 seconds, but SMS implementation reduces it to 50 seconds (Fig. 13A, B). Due to the number of slices required for liver imaging (and the lower SNR of the body DWI), SMS may have a greater impact in this body region to reduce scan time (e.g. axial long bone imaging). The possible potential for acceleration imaging of the knee is high, with similar quantitative and qualitative results as demonstrated in a recent small clinical study that evaluated the effect of a ~50% reduction in scan acquisition time using an acceleration factor of 2.41. It has also been found to help reduce the time patients spend in the magnet, even in trauma patients. SMS technology was initially introduced for 3 T, but has been extended to 1.5 T. There appears to be little difference between 1.5 T and 3 T in terms of effectiveness, especially in terms of reduced scan time [40].

Discussion

The strategy for managing motion artefacts on MR images is divided into three categories: prevention, reduction, and correction. Prevention is the simplest and often the most effective method because it acts on the very source of artefact creation. It is especially effective for preschool and school children if a lot of effort and time is devoted by the radiological staff, as well as the preparation of the parents before the examination itself. In people with claustrophobia and anxiety, there is no impact unless it is general anaesthesia, which is generally less often available due to the lack of specialised personnel, as well as special devices for working in a magnetic field.

The next method used is artefact reduction using external devices to monitor the patient's functions such as respiratory compensation, ECG and pulse monitoring. After recording phases of breathing or heartbeat, the device records the phase of the smallest movement (expiration or diastole). Methods such as shortening the duration of the sequence or the use of software triggering and gating are used in most MR departments, but in the case of an uncooperative patient with different breathing patterns, their possibilities are limited.

The most recently developed method is the correction of movement after the movement has occurred, and it is currently the most active area of development of the mentioned topic. There we have the use of navigators that are determined by software on reference images or prospective methods with the use of a navigator or a camera system with a marker on the coil. Both methods require additional time to assess the breathing cycle, heart rate and setup itself and are sensitive to poor positioning, which is impractical in a clinical system. There is also a retrospective method where artefact correction is performed after the recording itself using the MR navigator or iterative algorithms („deep learning"). The retrospective method has shown solid results, but it is currently still limited to certain types of movement (in the case of complex and different patterns, there is currently no solution).

Among the future techniques, the most significant is SMS, which is currently used significantly in studies, because, unlike parallel recording, there is no loss of SNR during reconstruction, and compressed observation (CS), which resulted in shortening the time of the sequence, enabled an additional reduction of the impact of motion on the images, but it currently has a disadvantage in the duration of the reconstruction, which should be corrected by the further development of hardware capabilities.

There is currently no general solution to the issue of motion artefacts, and there probably won't be in the future either. Each of the mentioned methods has its advantages and disadvantages, and their application is limited to specific cases. The reason for this is the large variety of sequences, methods of k-space sampling and image contrast, along with different types and patterns of motion during MR imaging, which makes them different from sequence to sequence. We can imagine the development of techniques as a set of tools that will be used depending on the purpose. ■

Budućnost tehnika korekcije artefakta micanja na MR uređaju

Sažetak

Micanje tijekom MR pregleda utječe na degradaciju slike u obliku „*ghosting*-a" ili zamućenja i jedan je od posljednjih velikih neriješenih problema. Pokreti pacijenta mogu biti različitog opsega, smjerova i orijentacija pritom otežavajući pronalazak rješenja od ranih dana razvoja MR-a. Cilj ovog rada je prikazati razvoj budućih strategija provedenih istraživanja o tehnikama i načinima smanjenja utjecaja artefakta micanja na kvalitetu dobivenih slika u svakodnevnim kliničkim MR pregledima.

Danas postoje tri glavne strategije borbe sa artefaktom pomicanja: prevencija micanja, redukcija micanja i korekcija micanja. Uslijed nedostatka mogućnosti i iscrpljenih ideja, najviše se radi na razvoju korekcije micanja pomoću MR navigatora ili prospektivnih i retrospektivnih metoda.

Od budućih tehnika trenutno najveću perspektivu imaju SMS (Simultaneous Multi-Slice) sekvence i komprimirano opažanje (CS) zbog otpornosti na micanje i kraćeg vremena snimanja u usporedbi sa konvencionalnim sekvencama.

Generalnog rješenja za artefakt micanja trenutno nema, već se problemu pristupa prema specifičnom slučaju birajući metodu s najmanje nedostataka, a uzimajući u obzir i vrstu pokreta kao i sekvencu odabira.

Ključne riječi: MR, artefakt micanja, metode, budućnost, sekvence, korekcija

Literature

1. Sadigh G, Applegate KE, Saindane AM. Prevalence of unanticipated events associated with MRI examinations: a benchmark for MRI quality, safety, and patientexperience. J Am Coll Radiol. 2017;14:765–772.
2. Gumus K, Keating B, White N, et al. Comparison of optical and MR-based tracking. Magn Reson Med. 2015;74:894–902.
3. Zaitsev M, Maclaren J, Herbst M. Motion artifacts in MRI: A complex problem with many partial solutions. J Magn Reson Imaging. 2015 Oct;42(4):887-901.
4. Brown TT, Kuperman JM, Erhart M, et al. Prospective motion correction of high-resolution magnetic resonance imaging data in children. Neuroimage 2010;53:139–145.
5. Thesen S, Heid O, Mueller E, Schad LR. Prospective acquisition correction for head motion with image-based tracking for real-time fMRI. Magn Reson Med 2000;44:457–463.
6. Zaitsev M, Dold C, Sakas G, Hennig J, Speck O. Magnetic resonance imaging of freely moving objects: prospective real-time motion correction using an external optical motion tracking system. Neuroimage 2006;31:1038–1050.
7. Ooi MB, Krueger S, Thomas WJ, Swaminathan SV, Brown TR. Prospective real-time correction for arbitrary head motion using active markers. Magn Reson Med 2009;62:943–954.
8. Fu ZW, Wang Y, Grimm RC, et al. Orbital navigator echoes for motion measurements in magnetic resonance imaging. Magn Reson Med 1995;34:746–753.
9. Welch EB, Manduca A, Grimm RC, Ward HA, Jack CR JR. Spherical navigator echoes for full 3D rigid body motion measurement in MRI. Magn Reson Med 2002;47:32–41.
10. Costa AF, Petrie DW, Yen YF, Drangova M. Using the axis of rotation of polar navigator echoes to rapidly measure 3D rigid-body motion. Magn Reson Med 2005;53:150–158.
11. Qin L, Gelderen P, Zwart J, Jin F, Tao Y, Duyn J. Head movement correction for MRI with a single camera. In: Proceedings of the 16th Annual Scientific Meeting of ISMRM, Toronto, Ontario, 2008. (abstract 1467).

12. Korin HW, Felmlee JP, Riederer SJ, Ehman RL. Spatial-frequencytuned markers and adaptive correction for rotational motion. Magn Reson Med 1995;33:663–669.
13. Sengupta S, Tadanki S, Gore JC, Welch EB. Prospective real-time head motion correction using inductively coupled wireless NMR probes. Magn Reson Med 2014;72:971–985.
14. Maclaren J, Armstrong BS, Barrows RT, et al. Measurement and correction of microscopic head motion during magnetic resonance imaging of the brain. PLoS One 2012;7:e48088.
15. Schulz J, Siegert T, Reimer E, et al. An embedded optical tracking system for motion-corrected magnetic resonance imaging at 7T. Magn Reson Mater Phys 2012;25:443–453.
16. Feinberg DA, Giese D, Bongers DA, et al. Hybrid ultrasound MRI for improved cardiac imaging and real time respiration control. Magn Reson Med 2010;63:290–296.
17. Speck O, Hennig J, Zaitsev M. Prospective real-time slice-by-slice motion correction for fMRI in freely moving subjects. Magn Reson Mater Phys 2006;19:55–61.
18. Schulz J, Siegert T, Bazin PL, et al. Prospective slice-by-slice motion correction reduces false positive activations in fMRI with taskcorrelated motion. Neuroimage 2014;84:124–132.
19. Gumus K, Keating B, Poser BA, et al. Prevention of motion-induced signal loss in diffusion-weighted echo-planar imaging by dynamic restoration of gradient moments. Magn Reson Med 2014;71:2006–2013.
20. Herbst M, Maclaren J, Weigel M, Korvink J, Hennig J, Zaitsev M. Prospective motion correction with continuous gradient updates in diffusion weighted imaging. Magn Reson Med 2012;67:326–338.
21. Keating B, Deng W, Roddey JC, et al. Prospective motion correction for single-voxel 1H MR spectroscopy. Magn Reson Med 2010;64:672–679.
22. Zaitsev M, Speck O, Hennig J, Buchert M. Single-voxel MRS with prospective motion correction and retrospective frequency correction. NMR Biomed 2010;23:325–332.
23. Lange T, Maclaren J, Buechert M, Zaitsev M. Spectroscopic imaging with prospective motion correction and retrospective phase correction. Magn Reson Med 2012;67:1506–1514.
24. Boegle R, Maclaren J, Zaitsev M. Combining prospective motion correction and distortion correction for EPI: towards a comprehensive correction of motion and susceptibility-induced artifacts. Magn Reson Mater Phys 2010;23:263–273.
25. Atkinson D, Hill DLG, Stoyle PNR, et al. Automatic compensation of motion artifacts in MRI. Magn Reson Med 1999;41:163–170.
26. McGee KP, Felmlee JP, Jack CR Jr, Manduca A, Riederer SJ, Ehman RL. Autocorrection of three-dimensional time-of-flight MR angiography of the Circle of Willis. AJR Am J Roentgenol 2001;176:513–518.
27. Loktyushin A, Nickisch H, Pohmann R, Scholkopf B. Blind retrospective motion correction of MR images. Magn Reson Med 2013;70:1608–1618.
28. Lauzon ML, Rutt BK. Generalized K-space analysis and correction of motion effects in MR imaging. Magn Reson Med 1993;30:438–446.
29. Beatty PJ, Nishimura DG, Pauly JM. Rapid gridding reconstruction with a minimal oversampling ratio. IEEE Trans Med Imaging 2005;24:799–808.
30. Pruessmann KP, Weiger M, Bornert P, Boesiger P. Advances in sensitivity encoding with arbitrary k-space trajectories. Magn Reson Med 2001;46:638–651.
31. Atkinson D, Hill DL. Reconstruction after rotational motion. Magn Reson Med 2003;49:183–187.
32. Marxen M, Marmurek J, Baker N, Graham SJ. Correcting magnetic resonance k-space data for in-plane motion using an optical position tracking system. Med Phys 2009;36:5580–5585.
33. Leung G, Plewes DB. Retrospective motion compensation using variable-density spiral trajectories. J Magn Reson Imaging 2005;22:373–380.
34. Aksoy M, Forman C, Straka M, Cukur T, Hornegger J, Bammer R. Hybrid prospective and retrospective head motion correction to mitigate cross-calibration errors. Magn Reson Med 2012;67:1237–1251.
35. Schmidt JF, Wissmann L, Manka R, Kozerke S. Iterative k-t principal component analysis with nonrigid motion correction for dynamic three-dimensional cardiac perfusion imaging. Magn Reson Med 2014;72:68–79.
36. Schmidt JF, Buehrer M, Boesiger P, Kozerke S. Nonrigid retrospective respiratory motion correction in whole-heart coronary MRA. Magn Reson Med 2011;66:1541–1549.
37. Bydder M, Larkman DJ, Hajnal JV. Detection and elimination of motion artifacts by regeneration of k-space. Magn Reson Med 2002;47:677–686.
38. Wintersperger BJ, Runge VM, Biswas J, et al. Brain magnetic resonance imaging at 3 Tesla using BLADE compared with standard rectilinear data sampling. Invest Radiol. 2006;41:586–592.
39. Shin HJ, Kim MJ, Lee MJ, et al. Comparison of image quality between conventional VIBE and radial VIBE in free-breathing paediatric abdominal MRI. Clin Radiol. 2016;71:1044–1049.
40. Runge VM, Richter JK, Heverhagen JT. Motion in Magnetic Resonance: New Paradigms for Improved Clinical Diagnosis. Invest Radiol. 2019 Jul;54(7):383-395.
41. Runge VM, Clanton JA, Partain CL, et al. Respiratory gating in magnetic resonance imaging at 0.5 Tesla. Radiology. 1984;151:521–523.
42. Liu YL, Riederer SJ, Rossman PJ, et al. A monitoring, feedback, and triggering system for reproducible breath-hold MR imaging. Magn Reson Med. 1993;30:507–511.
43. Feng L, Axel L, Chandarana H, et al. XD-GRASP: golden-angle radial MRI with reconstruction of extra motion-state dimensions using compressed sensing. Magn Reson Med. 2016;75:775–788.
44. Taron J, Weiss J, Notohamiprodjo M, et al. Acceleration of magnetic resonance cholangiopancreatography using compressed sensing at 1.5 and 3 T: a clinical feasibility study. Invest Radiol. 2018;53:681–688.
45. Yoon JH, Lee SM, Kang HJ, et al. Clinical feasibility of 3-dimensional magnetic resonance cholangiopancreatography using compressed sensing: comparison of image quality and diagnostic performance. Invest Radiol. 2017;52:612–619.
46. Zhu L, Wu X, Sun Z, et al. Compressed-sensing accelerated 3-dimensional magnetic resonance cholangiopancreatography: application in suspected pancreatic diseases. Invest Radiol. 2018;53:150–157.
47. Wetzl J, Schmidt M, Pontana F, et al. Single-breath-hold 3-D CINE imaging of the left ventricle using Cartesian sampling. MAGMA. 2018;31:19–31.
48. Naresh NK, Haji-Valizadeh H, Aouad PJ, et al. Accelerated, first-pass cardiac perfusion pulse sequence with radial k-space sampling, compressed sensing, and k-space weighted image contrast reconstruction tailored for visual analysis and quantification of myocardial blood flow. Magn Reson Med. 2019;81:2632–2643.
49. Bustin A, Ginami G, Cruz G, et al. Five-minute whole-heart coronary MRA with sub-millimeter isotropic resolution, 100% respiratory scan efficiency, and 3DPROST reconstruction. Magn Reson Med. 2019;81:102–115.
50. Bacher M, Speier P, Bollenbeck J, et al. Pilot tone navigation enables contactless prospective cardiac triggering: initial volunteer results for prospective cine. Proceedings of the 26th Annual Meeting of the ISMRM, #2960. 2018. [citirano 15.3.2022.]. Dostupno na: http://archive.ismrm.org/2018/2960.html.
51. Setsompop K, Gagoski BA, Polimeni JR, et al. Blipped-controlled aliasing in parallel imaging for simultaneous multislice echo planar imaging with reduced g-factor penalty. Magn Reson Med. 2012;67:1210–1224.
52. Barth M, Breuer F, Koopmans PJ, et al. Simultaneous multislice (SMS) imaging techniques. Magn Reson Med. 2016;75:63–81.
53. Runge VM, Richter JK, Heverhagen JT. Speed in clinical magnetic resonance. Invest Radiol. 2017;52:1–17.
54. Setsompop K, Cohen-Adad J, Gagoski BA, et al. Improving diffusion MRI using simultaneous multi-slice echo planar imaging. Neuroimage. 2012;63:569–580.
55. Cauley SF, Polimeni JR, Bhat H, et al. Interslice leakage artifact reduction technique for simultaneous multislice acquisitions. Magn Reson Med. 2014;72:93–102.