Controlled Aliasing in Parallel Imaging Results in Higher Acceleration (CAIPIRINHA)

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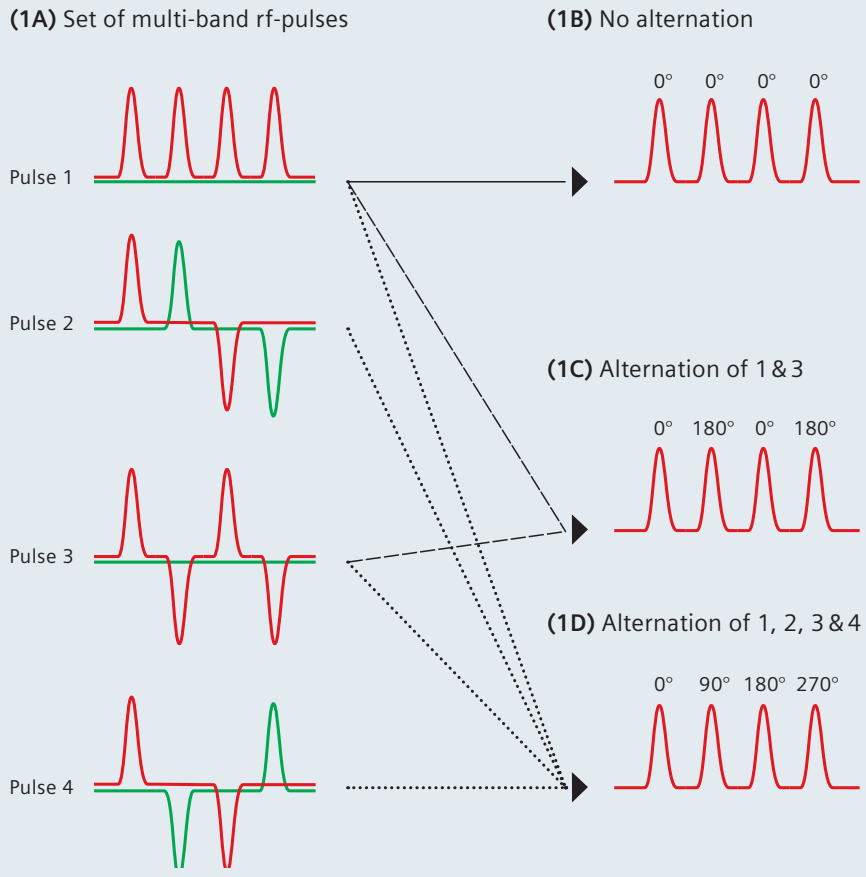


1 Multi-slice excitation with alternating rf-pulses taken from **(1A)** a set of 4 rf-pulses with different phase modulations (pulse 1 to 4) allows one to provide the individual slices with well defined phase-cycles along the phase encoding direction. The real part (red) and imaginary part (green) of the pulses are plotted. **(1B)** Using only one pulse (e.g. pulse 1) no phase-cycle is provided (0°, 0°, 0°, 0°). **(1C)** Alternation between pulses 1 and 3 yield no phase-cycle for slice 1 and 3 and an 180° phase cycle for slices 2 and 4 (0, 180°, 0, 180°). **(1D)** Alternation of all 4 pulses allows one to provide all the individual slices with an individual phase-cycle (0, 90°, 180°, 270°).

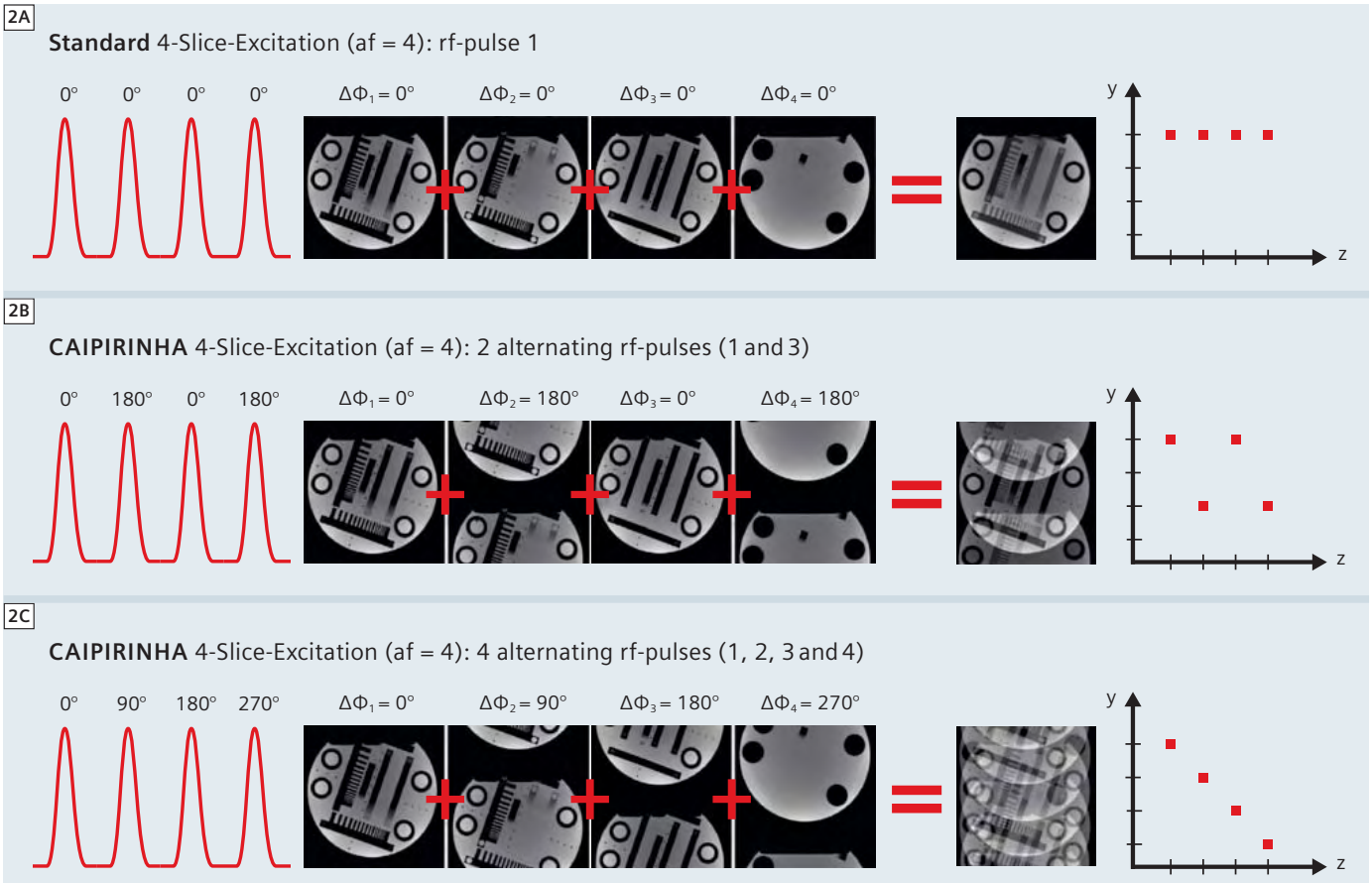
Introduction

Image acquisition time is one of the most important considerations for clinical magnetic resonance imaging (MRI). The development of multi-coil receiver hardware as well as dedicated parallel MRI (pMRI) reconstruction methods such as SENSE [1] and GRAPPA [2] allowed for significant decrease of acquisition times in almost all clinical applications. Thus, today, pMRI plays a substantial role in everyday clinical routine.

pMRI operates by reducing the amount of data necessary to form an image. In the Cartesian case, this is usually accomplished by uniformly undersampling the k-space (e.g. skipping every other phase-encoding line) resulting in so-called 'aliasing artifacts' in the image domain. pMRI reconstruction methods seek to compensate the lack of spatial encoding by taking into account the spatial sensitivity information, provided by a multi-coil receiver array. Unfortunately, the pMRI concept is intrinsically associated with a signal-to-noise (SNR) loss compared to a fully encoded image. The SNR is

a) reduced by the square root of the acceleration factor, simply due to the fact that less data is acquired, and
b) by the so-called g-factor, depending strongly on the encoding capabilities of the underlying receiver array.

Thus, pMRI is often limited to applications with sufficiently high base SNR, such as volumetric imaging methods. With the newest generation of MR scanners



2 **(2A)** Standard 4 times accelerated simultaneous 4-slice-experiment. The same pulse is applied for subsequent excitations (phase encoding lines), resulting in all 4 slices projected onto each other. **(2B)** CAIPIRINHA 4-times accelerated simultaneous 4-slice-experiment alternating between e.g. pulses 1 and 3 for subsequent excitations, thereby providing slice 2 and slice 4 with a 180° phase cycle. According to the Fourier Shift Theorem, slices 2 and 4 appear shifted by half of the FOV with respect to slices 1 and 3 in the resulting folded image. **(2C)** CAIPIRINHA 4-times accelerated simultaneous 4-slice-experiment alternating between pulse 1, 2, 3 and 4 for subsequent excitations, thereby providing each slice with a different phase-cycle (0°, 90°,180°, 270°). Each slice appears shifted by FOV/4 with respect to their adjacent slice in the folded image.

providing up to 128 independent receiver channels, further scan time reductions are potentially achievable. However, in conventional 2D clinical imaging, parallel imaging today is still restricted to relatively moderate scan time reductions (acceleration factors of 2 to 3) due to intrinsic limitations in the coil sensitivity variations along one phase encoding direction (1D parallel imaging). In 3D and simultaneous multi-slice imaging, parallel encoding can be carried out in two encoding directions (2D parallel imaging), thereby employing the sensitivity variations in both directions, as has been demonstrated in, for example, 2D SENSE [3] and MS SENSE [4]. This concept has been shown to significantly improve the reconstruction conditions, allowing for higher accelerations of the acquisition (>3). However, both techniques require sufficient sensitivity variations in two encoding directions for successful image reconstruction and therefore strongly depend on the underlying coil geometry. As mentioned above, spatial encoding with a receiver array is associated with a certain noise amplification known as 'g-factor noise'. Quantitative g-factor estimation methods have been derived for SENSE [1] and GRAPPA reconstructions [5] and serve as a quality metric for pMRI reconstructions. One important approach to reduce this g-factor noise for a given application is the optimization of the receiver array geometry (e.g. number of coils, coil arrangement) towards the application at hand. However, hardware limitations, the diversity of patient weight and size, the need for flexibility regarding a wider range of applications, as well as sequence or protocol specific considerations, hamper the viability.

The **CAIPRINHA** concept (**C**ontrolled **A**liasing **I**n **P**arallel **I**maging **R**esults **IN** **H**igher **A**cceleration) allows one to partially overcome these requirements and limitations by modifying the aliasing conditions in a well defined manner. This is done already during the data acquisition by modifying the rf-excitation or gradient encoding scheme in

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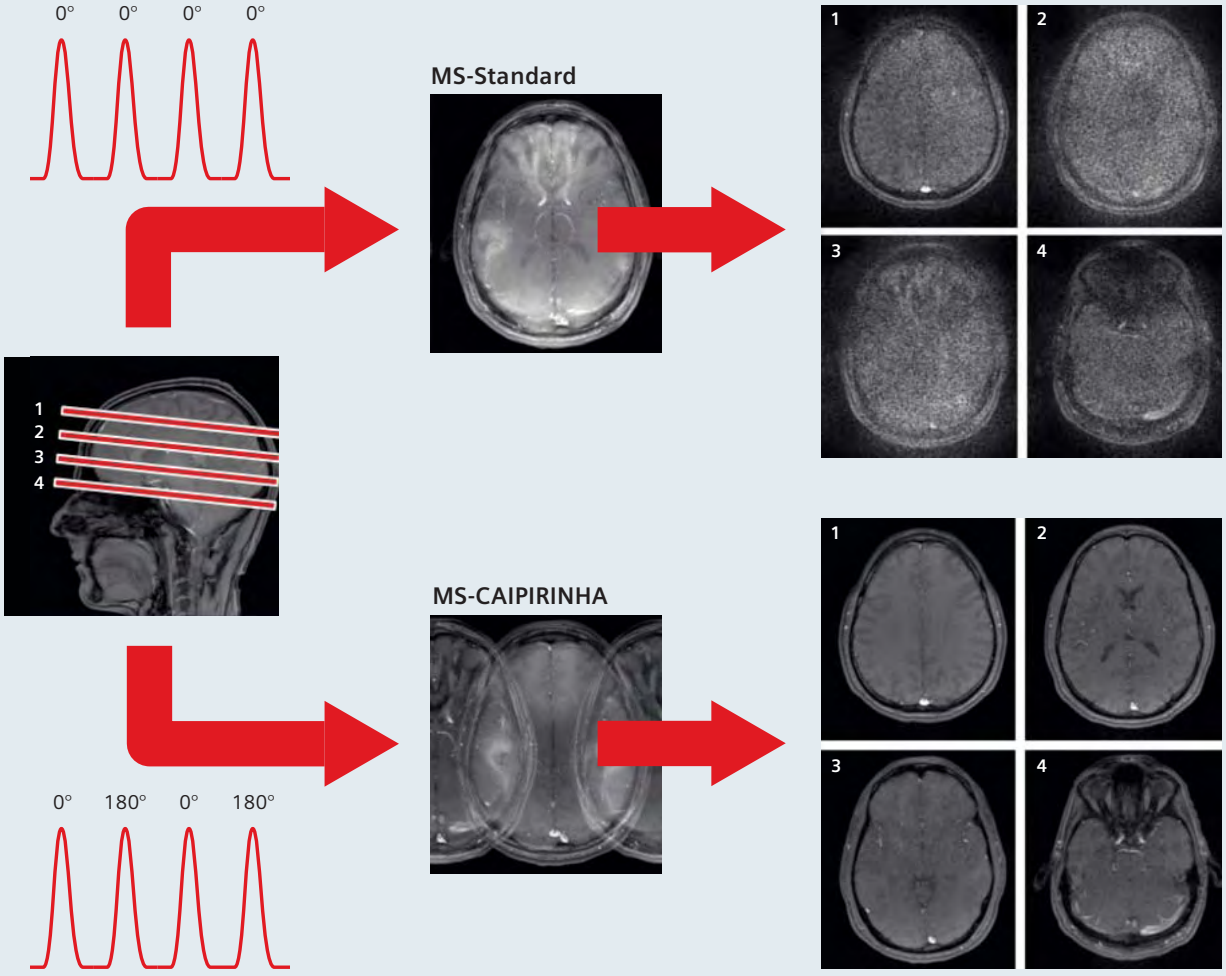


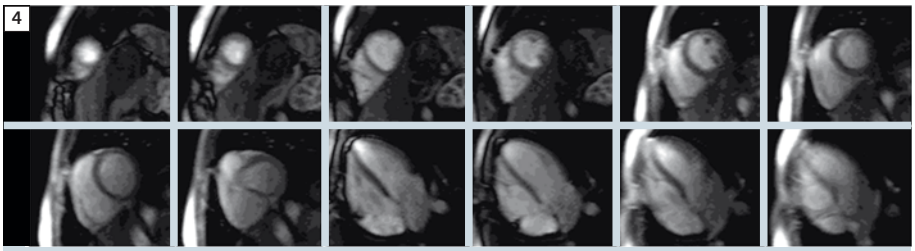
3 In vivo brain example: 4 x accelerated simultaneous 4-slice experiment using no phase cycling (MS-Standard) results in superimposition of all the slices directly on top of each other. Due to the lack of sufficient sensitivity variations along the slice direction strong noise amplifications can be observed after GRAPPA reconstruction. Using MS-CAIPIRINHA, employing 2 alternating multi-band rf-pulses, slices 2 and 4 appear shifted with respect to slices 1 and 3 in the folded image. In this way sensitivity variations in the phase encoding direction (LR) can be used in addition to the sensitivity variations available in the slice direction. The concept results in significantly improved image quality after GRAPPA reconstruction. Imaging parameters: 3T MAGNETOM Skyra, TE 3.4 ms, TR 100 ms, FA 50°, FOV 178 x 220 mm², matrix 208 x 320, slice thickness 4 mm, distance factor 300 %.

order to use the coil encoding power of the underlying receiver array to full capacity. The concept has been successfully applied so far to simultaneous multi-slice imaging (MS-CAIPIRINHA) [6] and 3D imaging where data reduction can be carried out in two phase-encoding directions (2D-CAIPIRINHA) [7]. In addition, both strategies can be extended to the third remaining direction, namely the read-out direction, by utilizing e.g. zig-zag shaped read-out trajectories [8]. The following provides a brief overview of MS-CAIPIRINHA and 2D-CAIPIRINHA.

Improving parallel imaging performance with CAIPIRINHA

MS-CAIPIRINHA

Simultaneous multi-slice imaging offers an SNR benefit over standard single-slice imaging and comprises rf-excitations with specialized multi-band pulses as displayed in Figure 1. After multi-band excitation the received signals will accrue from all the slices (bands) and thus are subject to the subsequent gradient encoding sequence. Simply replacing the standard single-slice excitation pulse with a multi-slice pulse in an MR imaging sequence will therefore result in an image with all the simultaneously excited slices projected onto each other (Fig. 2A). As mentioned above, the parallel imaging concept provides an elegant way to separate multiple image signals which are aliased into one image pixel. Thus, sufficient sensitivity variations of the underlying receiver array



4 In vivo cardiac example: The MS-CAIPIRINHA approach enables the acquisition of up to 6 slices per cardiac cycle. Here, 12 slices are acquired within 2 cardiac cycles (8 slices in short axis view and 4 in the long axis). Imaging parameters: 1.5T MAGNETOM Avanto, 32-channel cardiac array (Rapid Biomedical, Würzburg, Germany); Sequence: SR-TrueFISP, CAIPIRINHA phase cycle +90°/-90°; FOV 320 x 260 mm², matrix 128 x 77, resolution 2.5 x 3.4 mm², slice thickness 10 mm, distance factor (two-slice pulse) of short/long axis: 200%/100%; partial Fourier 6/8, measurements: 20, TR 2.8 ms, TI 120 ms, TE 1.4 ms, FA 50°, reconstruction algorithm GRAPPA (R=3). *Images courtesy of Daniel Stäb.*

along the slice direction will then allow for separation of the slices using adapted standard pMRI reconstruction algorithms [4, 9]. However, in cases where the sensitivity variations along the slice direction are not sufficient e.g. as a result of small slice distances or suboptimal coil geometry, the pMRI reconstruction will fail and result in large noise amplification. Sensitivity variations, potentially available along the other spatial directions, here the phase-encoding direction, are not employed.

It has been demonstrated that increasing the field-of-view (FOV) by the number of simultaneously excited slices allows the individual slices to be shifted with respect to each other in an extended FOV (along the phase-encoding) [10, 11] such that the slices show no superposition. A similar concept is Hadamard aided rf-encoding [12]. The required shifts mentioned above can be accomplished by employing dedicated alternating multi-band rf-pulses providing the individual bands with well-defined phase-cycles along the phase-encoding direction (e.g using the set of rf-pulses displayed in Fig. 1). Due to the volumetric excitation this approach offers a benefit in SNR efficiency of square root of the number of simultaneously excited slices compared to single slice acquisitions, however at the cost of increased pulse energy deposition.

Using this concept in combination with image acceleration (fewer phase-encoding steps) superimposed slices with individual shifts along the phase-encoding direction can be realized by employing alternation of rf-pulses taken e.g. from the set of pulses given in Fig. 1. A four-slice excitation at an acceleration of af = 4 using only rf-pulse 1 yields a superimposition of 4 image pixels originating from all the 4 slices at the same location in the phase-encoding direction (Fig. 2A). Employing an alternation of rf-pulses (e.g. pulse 1 and pulse 3, or pulses 1, 2, 3 and 4) the individual slices can be shifted with respect to each other in the FOV (Figs. 2B, C). In this way, as demonstrated in the corresponding z_y-plots, aliased pixels may now originate from both different slices and different locations in the phase-encoding direction in a well defined manner (MS-CAIPIRINHA), thereby allowing the pMRI reconstruction to take advantage of sensitivity variations in the slice and the phase encoding direction.

The benefit of MS-CAIPIRINHA is demonstrated in vivo employing a 4 times accelerated simultaneous 4-slice experiment: Figure 3 shows 4 slices in a volunteer's brain (slice positions are indicated in the sagittal brain image), which are excited simultaneously using specialized multi-band rf-pulses taken from the set of pulses given in Figure 1A. In the case of non-alternating rf-pulses [4] (MS-Standard), each slice is subject to the same phase cycle along the phase-encoding direction (LR). The slices appear projected directly on top of each other, thereby allowing the pMRI reconstruction (here GRAPPA-SENSE hybrid [9]) only to use sensitivity variations available in the slice direction. Due to the relatively small slice distances, the relatively high acceleration factor (af = 4) and the limited sensitivity variations provided by the coil array in the slice-direction, the reconstruction results in large noise amplifications and thus unacceptable image quality. However, using a MS-CAIPIRINHA acquisition in combination with an adapted GRAPPA reconstruction, the folded image pixels can now be separated almost without any noise amplification. In this example, an MS-CAIPIRINHA scheme as depicted in Figure 2B has been employed. Alternation of pulses 1 and 3 provides slices 2 and 4 with a 180° phase-cycle along the phase encoding direction causing these slices to appear shifted by FOV/2 with respect to the slices 1 and 3 which had no phase modulation. Thus, in this case, MS-CAIPIRINHA allowed the acquisition of 4 slices in the same time normally required or a single slice without losing SNR.

In addition, the applicability of MS-CAIPIRINHA to cardiac perfusion imaging is demonstrated in Figure 4. A two-slice CAIPIRINHA saturation recovery TrueFSIP sequence has been employed using a total acceleration of af = 3. This allows for the acquisition of 12 slices (8 slices in the short axis view and 4 in the long axis) in only two cardiac cycles. A repetition of the sequence during contrast agent uptake has the potential for cardiac perfusion imaging with significantly increased spatial coverage in high temporal resolution [13].

2D-CAIPIRINHA

In contrast to conventional 2D imaging where only one phase-encoding direction is available for scan-time reduction (1D pMRI), 3D volumetric imaging with a second phase encoding direction offers the potential to choose the direction in which undersampling is performed, or even to split the acceleration between the two phase-encoding directions (2D pMRI). Given a receiver array geometry providing sensitivity variations in both phase encoding directions, this

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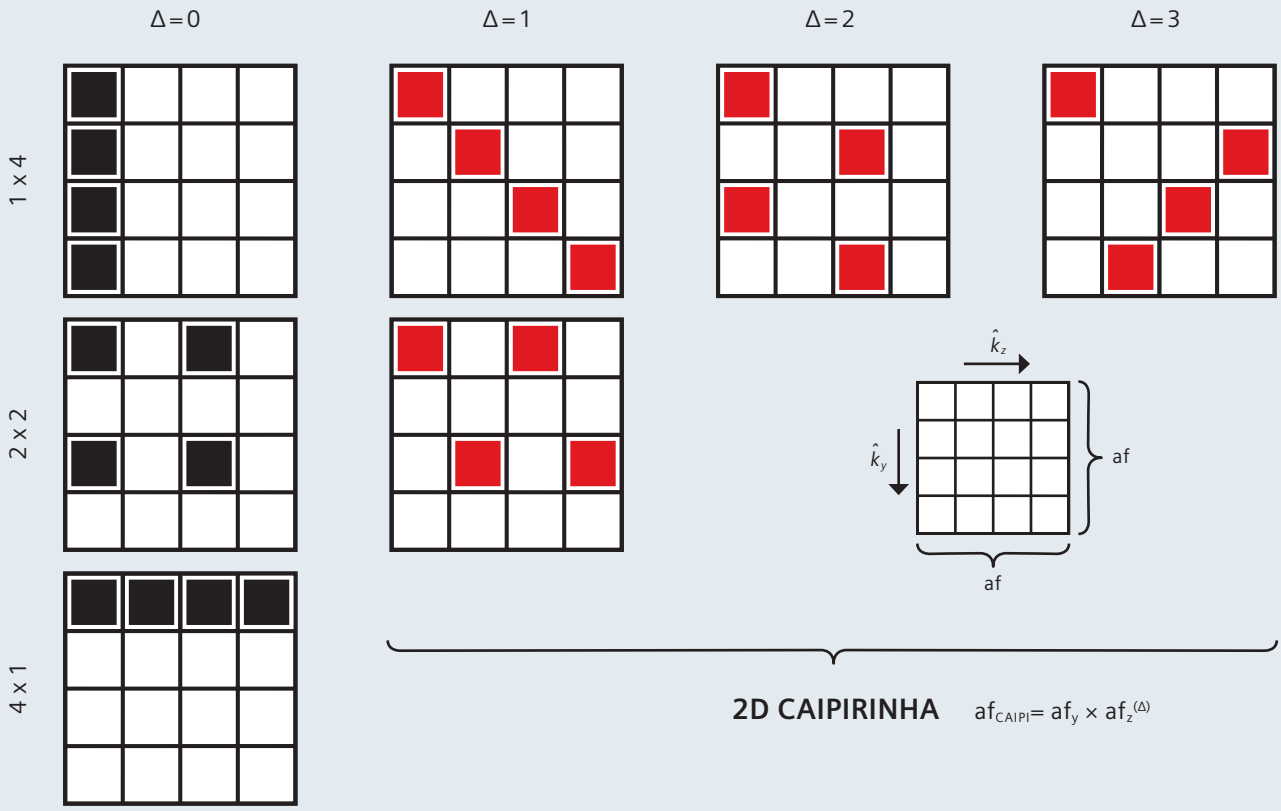
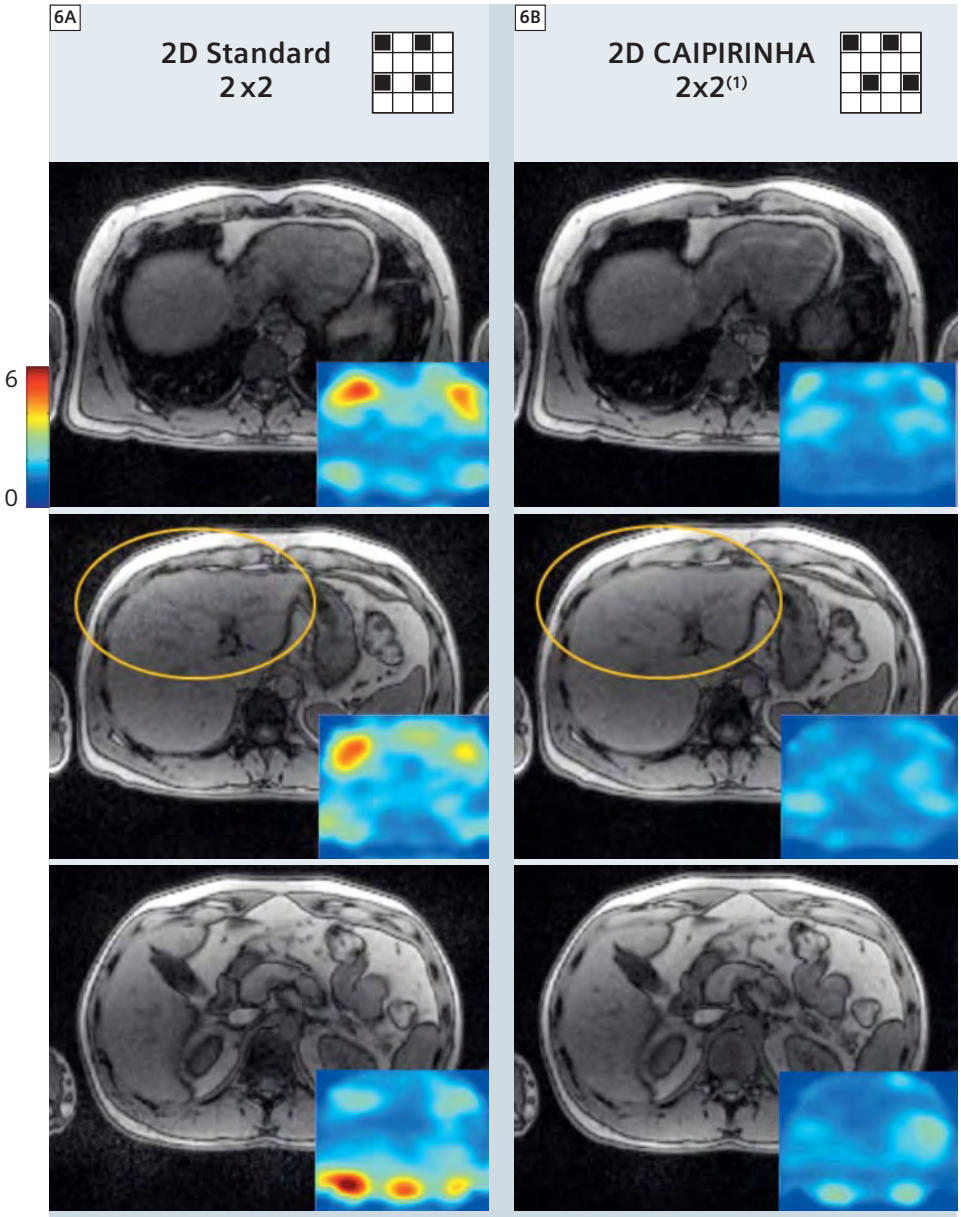


5 Procedure of generating 2D-CAIPIRINHA sampling patterns for a given total acceleration factor, here af = 4. All possible sampling schemes can be represented by an af x af elementary cell with af sampling positions to fill. For each undersampling rate in the k_y direction (af_y) multiple patterns can be created by shifting sampling positions at row k_y in the k_z direction by a different amount Δ, whereas Δ runs from 0 to af_z-1, and $af_z = af/af_y$. Sampling patterns without shift (Δ = 0) are 2D standard acquisitions, while all the other patterns are represented by 2D-CAIPIRINHA-type acquisitions ($af = af_y \times af_z^{(\Delta)}$) indicated by the red sampling positions.

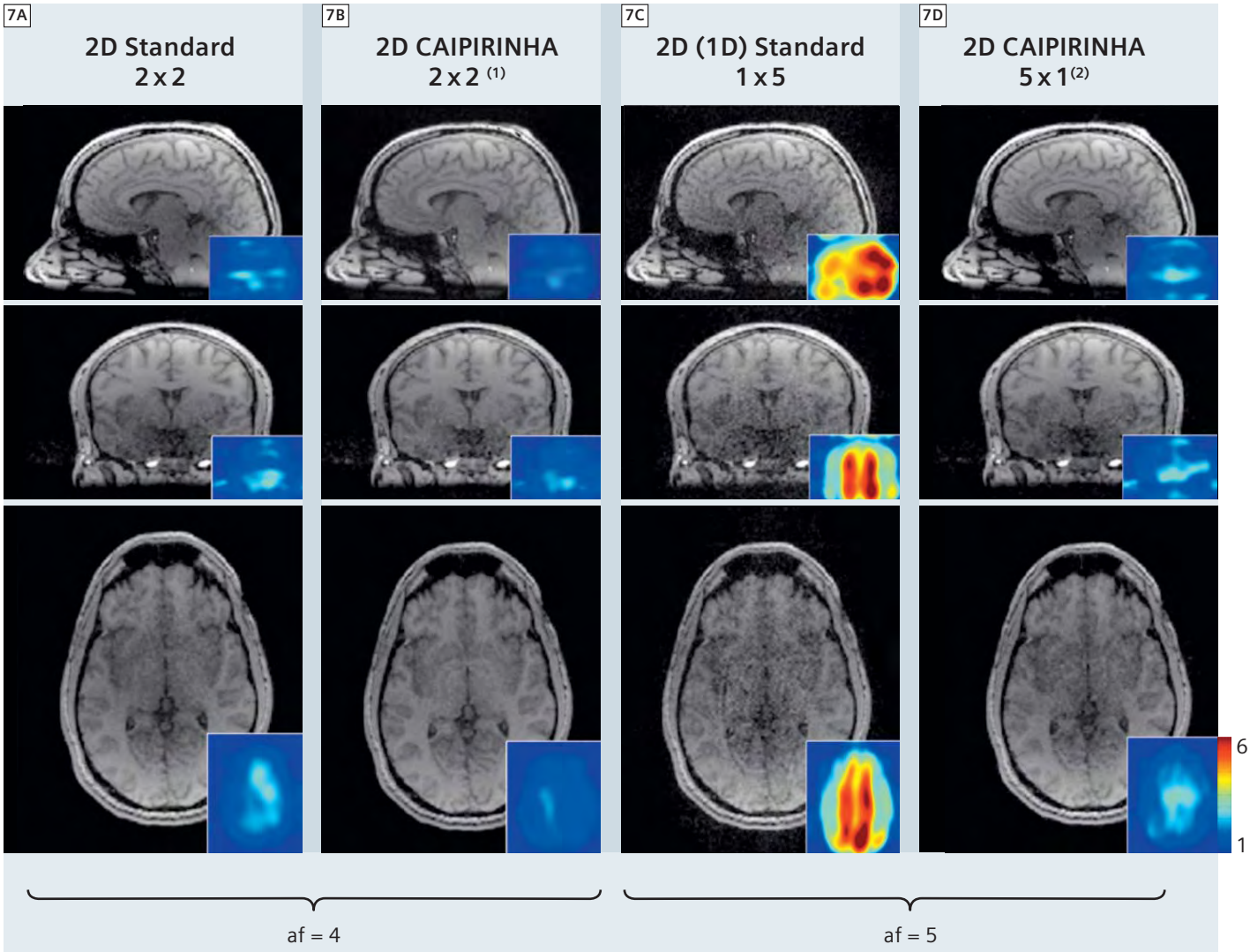
strategy has shown the potential to allow for higher total image accelerations compared to undersampling schemes restricted to only one direction [4, 5]. However, since the sensitivity variations available for the pMRI reconstruction depend not only on the coil geometry but also on the image position and orientation, the choice of the FOVs and encoding directions as well as the object position, size and shape, the right choice of the undersampling rate for the individual phase-encoding directions is not easily predictable and remains a challenging task. Thus, in many applications the reconstructed images suffer from severe residual artifacts or strong noise amplifications, depending on the choices made by the operator.

Again, the CAIPIRINHA concept has shown to partially overcome these limitations. It has been realized that, besides the standard rectangular sampling patterns with undersampling using simple integer reductions, many other patterns are conceivable where the sampling positions are shifted from their original positions in the 2D phase encoding scheme. Here we restrict ourselves to sampling positions on so-called 'sheared grids' which form periodic lattices [14] resulting in exactly af superimposed image pixels at an acceleration factor of af as it is the case in all standard rectangular patterns. The procedure of generating the available 2D-CAIPIRINHA patterns is schematically displayed in Figure 5 for a total image acceleration of af = 4. The sampling schemes can be represented by an af x af elementary cell with af sampling positions to fill. For each undersampling rate in the k_y direction (af_y) multiple patterns can be created by shifting sampling positions at row k_y in the k_z direction by a different amount d, whereas d runs from 0 to af_z-1, and $af_z = af/af_y$. Sampling patterns without shift (d = 0) are 2D standard acquisitions, while all the other patterns are represented by 2D-CAIPIRINHA-type acquisitions. This concept can also be used for prime number accelerations (af = 2, 3, 5 ...) where standard accelerations only allow undersampling in one of the phase encoding directions. The required shifts in k-space can simply be realized by applying additional gradient offsets to the phase encoding gradient tables. These 2D-CAIPIRINHA sampling patterns, analogous to the phase-cycles in simultaneous



6 In vivo liver example; volunteer: Compared are GRAPPA reconstructions (3 example slices) derived from two different reduction schemes **(6A)** Standard 2x2 and **(6B)** 2D-CAIPIRINHA 2x2(1). In addition, the corresponding GRAPPA g-factor maps are displayed. In the indicated region the SNR benefit of 2D-CAIPIRINHA can be appreciated.
Imaging details: 1.5T MAGNETOM Avanto, 6-channel body matrix coil and 6-channel spine matrix coil; VIBE af = 4, extra reference scan matrix 32 x 24 x 24. FOV 400 x 312.5 mm², matrix 320 x 170 x 50 GRAPPA, total acquisition time 9 s breath-hold.

multi-slice imaging, modify the appearance of aliasing in 2D parallel imaging compared to conventional rectangular reduction schemes and have the potential to relax the requirements of integer reductions to great extent. This is demonstrated in more detail in the original publication [7]. By shifting the sampling positions in a well-directed manner, aliasing can be shifted in such a way that sensitivity variations provided by the underlying receiver array are employed more efficiently. In some cases the amount of aliasing can even be reduced. These modified aliasing conditions may then result in a further improvement in parallel imaging reconstruction conditions and therefore in better image quality. Recently, this concept has also been extended to more generalized sampling schemes which are not restricted to sheared grids [15].
In order to demonstrate the benefit of 2D-CAIPIRINHA in vivo, two subsequent accelerated (af = 4) abdominal 9 s breath-hold VIBE experiments have been carried out on a volunteer. In Figure 6 GRAPPA reconstructions from three out of 50 slices from
a) a standard 2x2 and
b) a 2D-CAIPIRINHA 2x2(1)
acquisition are displayed. In addition, the corresponding g-factor maps of the GRAPPA reconstructions are displayed as a quantitative measure of image quality. As indicated by the lower g-factor values in the 2D-CAIPIRINHA reconstructions the improved image quality can clearly be observed, even on a visual scale (see region indicated by the orange circle).
Furthermore, the improvements in image quality associated with 2D-CAIPIRINHA are demonstrated taking four different T1-weighted 3D FLASH experiments of a volunteer's brain with different acceleration factors and acquisition schemes (Fig. 7). The acquisitions compared are
a) standard 2x2,
b) 2D-CAIPIRINHA 2x2(1),
c) 2D-CAIPIRINHA 1x5(2) and
d) standard 5x1 scheme.
Displayed are the central sections of the reconstructed 3D image data in the sagittal, coronal and axial view in addition to the corresponding quantitative g-factor maps. Comparing reconstruction results from af = 4 (7A) and (7B) the improvement of 2D-CAIPIRINHA can clearly be appreciated. Comparing results from af = 5 (7C) and (7D) the gain in SNR is even more obvious. In this case the parallel imaging performance of 2D-CAIPIRINHA 1x5(2) (7C) compares pretty well with the standard af = 4 (2x2) acquisition employed in (7A). While the 2D-CAIPIRINHA patterns in general appear to be more tolerant against user influence and suboptimal patient positioning, the automatic



7 In vivo 3D FLASH brain imaging using different acceleration schemes: **(7A)** Standard 2x2 **(7B)** 2D-CAIPIRINHA 2x2[1] **(7C)** Standard 5x1 **(7D)** 2D-CAIPIRINHA 1x5[2]. Displayed are central slices in the sagittal, coronal and axial view. In addition the corresponding GRAPPA g-factor maps are shown.
Imaging details: 3T MAGNETOM Skyra, 20-channel head neck matrix coil, 3D FLASH, GRAPPA with extra reference scan, matrix 32 x 32 x 32, TE / TR 4.3 ms / 16 ms, FA 35°, FOV 256 x 208 x 204 mm³, matrix 256 x 168 x 144; partial Fourier factor 7/8, total scan time 1 min 40 s (af = 4) and 1 min 16 s (af = 5).

extraction of the optimal pattern for the given imaging setup remains a challenging task and has not been sufficiently answered.

Conclusion

In all current parallel imaging techniques, aliasing artifacts resulting from an undersampled acquisition are removed by a specialized pMRI image reconstruction algorithm. The CAIPIRINHA concept aims at modifying the appearance of the aliasing artifacts already during the acquisition to improve the following parallel image reconstruction procedure. Specifically, this concept has been successfully applied to simultaneous multi-slice imaging (MS-CAIPIRINHA) and 3D imaging (2D-CAIPIRINHA).

MS-CAIPIRINHA

Aliasing in simultaneous multi-slice acquisitions can be modified already during the acquisition by employing alternating rf-pulses for subsequent phase encoding lines, thereby allowing the imprint of the individual slices with individual phase-cycles causing the slices to appear shifted with respect to each other thereby improving the reconstruction process minimizing g-factor related noise enhancements. Thus, a CAIPIRINHA-type 4 slice excitation with low g-factor values (close to 1) allows the acquisition of 4 slices in the same time usually required for 1 slice without

loss of SNR. Recently, the MS-CAIPIRINHA concept has been also successfully applied to more advanced acquisition schemes such as SSFP [13] EPI [16] and radial [17] simultaneous multi-slice imaging.

However, it is important to note that multi-slice excitations are associated with significantly increased energy deposition, currently limiting the method to a moderate number of simultaneously excited slices, and/or to low flip angles. However, recently, a promising concept for reducing the rf power of multi-band pulses has been introduced [18]. Thus, MS-CAIPIRINHA is expected to become a powerful strategy in the near future allowing to significantly accelerate many clinical protocols in almost preserved image quality.

2D-CAIPIRINHA

In conventional pMRI accelerated 3D imaging, data reduction is performed in two spatial dimensions simultaneously by integer-valued undersampling in each phase encoding direction. Though sensitivity variations can be exploited in two spatial dimensions, this sampling strategy provides suboptimal encoding performance. The 2D-CAIPIRINHA strategy similar to MS-CAIPIRINHA modifies aliasing in a controlled manner already during the data acquisition. This is accomplished by shifting sampling positions in the two dimensional phase encoding scheme with respect to each other. In this way, at certain image acceleration values, an optimal sampling pattern can be found which minimizes signal overlap and at the same time allows one to efficiently take advantage of all the sensitivity variations provided by the coil array in the 2D phase encoding plane. Thus, 2D-CAIPIRINHA provides optimal reconstruction performance given a certain coil configuration and object shape, and therefore results in optimal image reconstruction quality.

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