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Answers for life.

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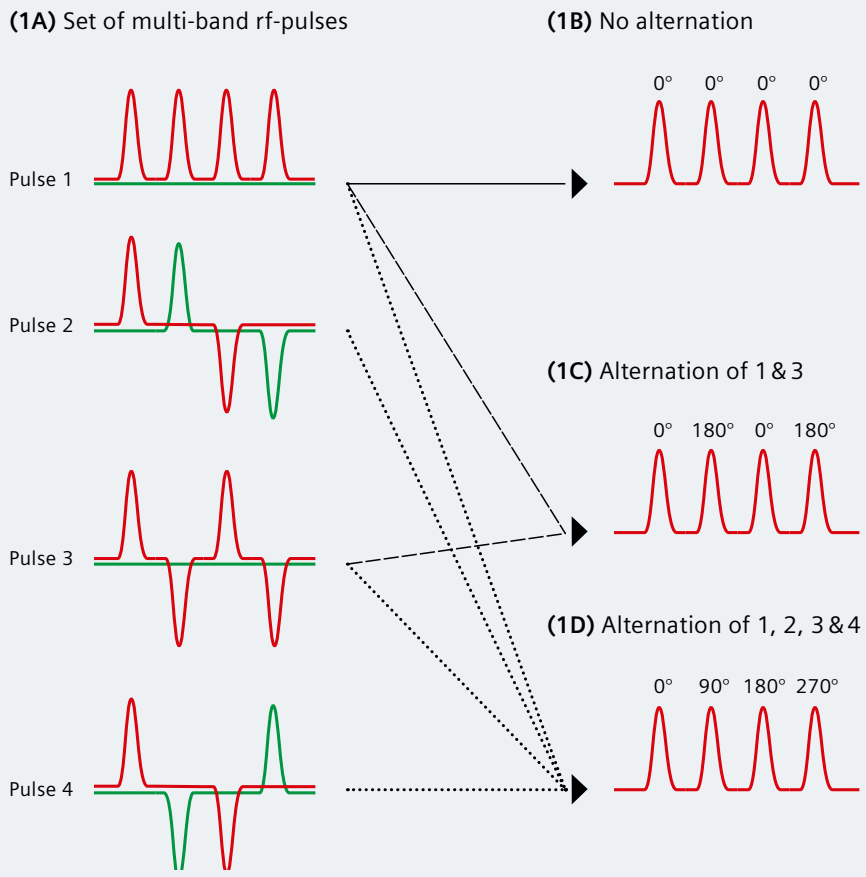

1 Multislice excitation with alternating rf-pulses taken from **(1A)** a set of 4 rf-pulses with different phase modulations (pulse 1 to 4) allows one to provide the individual slices with well-defined phase-cycles along the phase-encoding direction. The real part (red) and imaginary part (green) of 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

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) toward 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 **CAIPIRINHA** 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 order to use the coil encoding power of the underlying receiver array to full capacity. The concept has been success-

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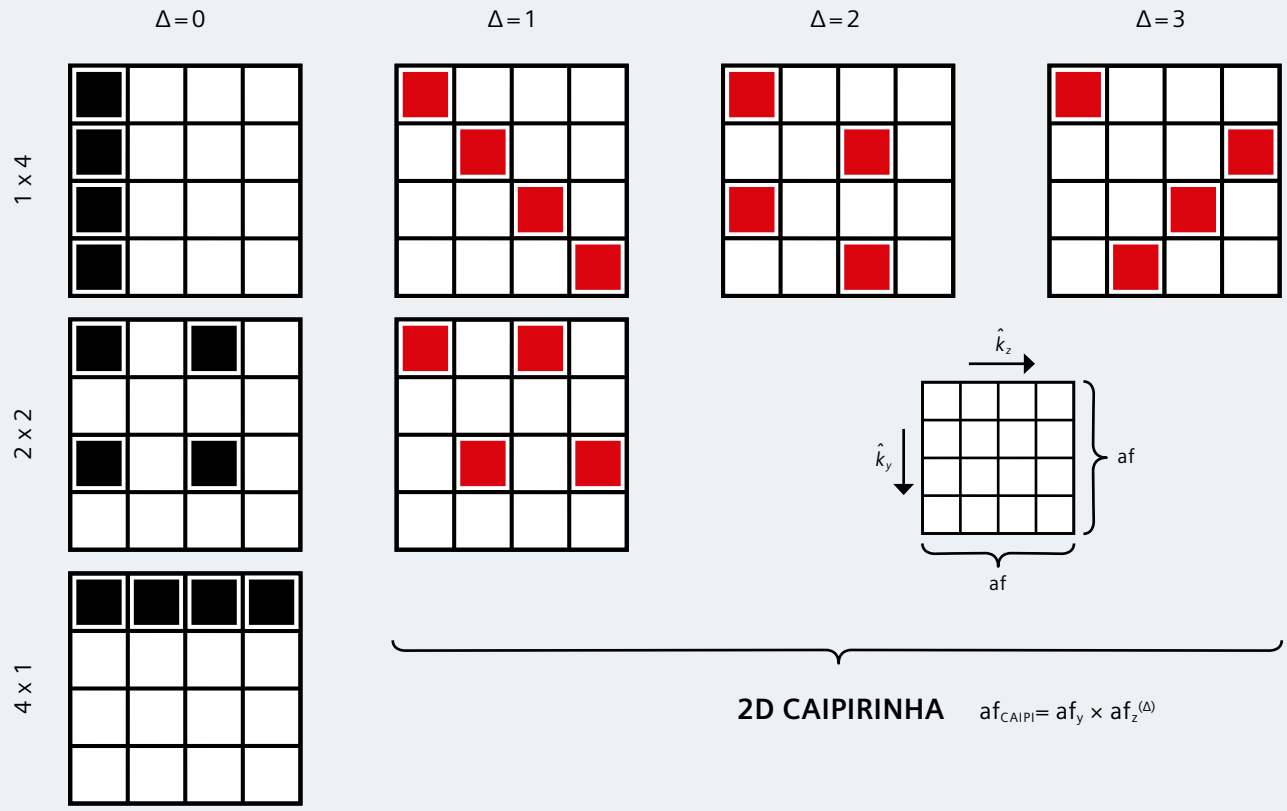
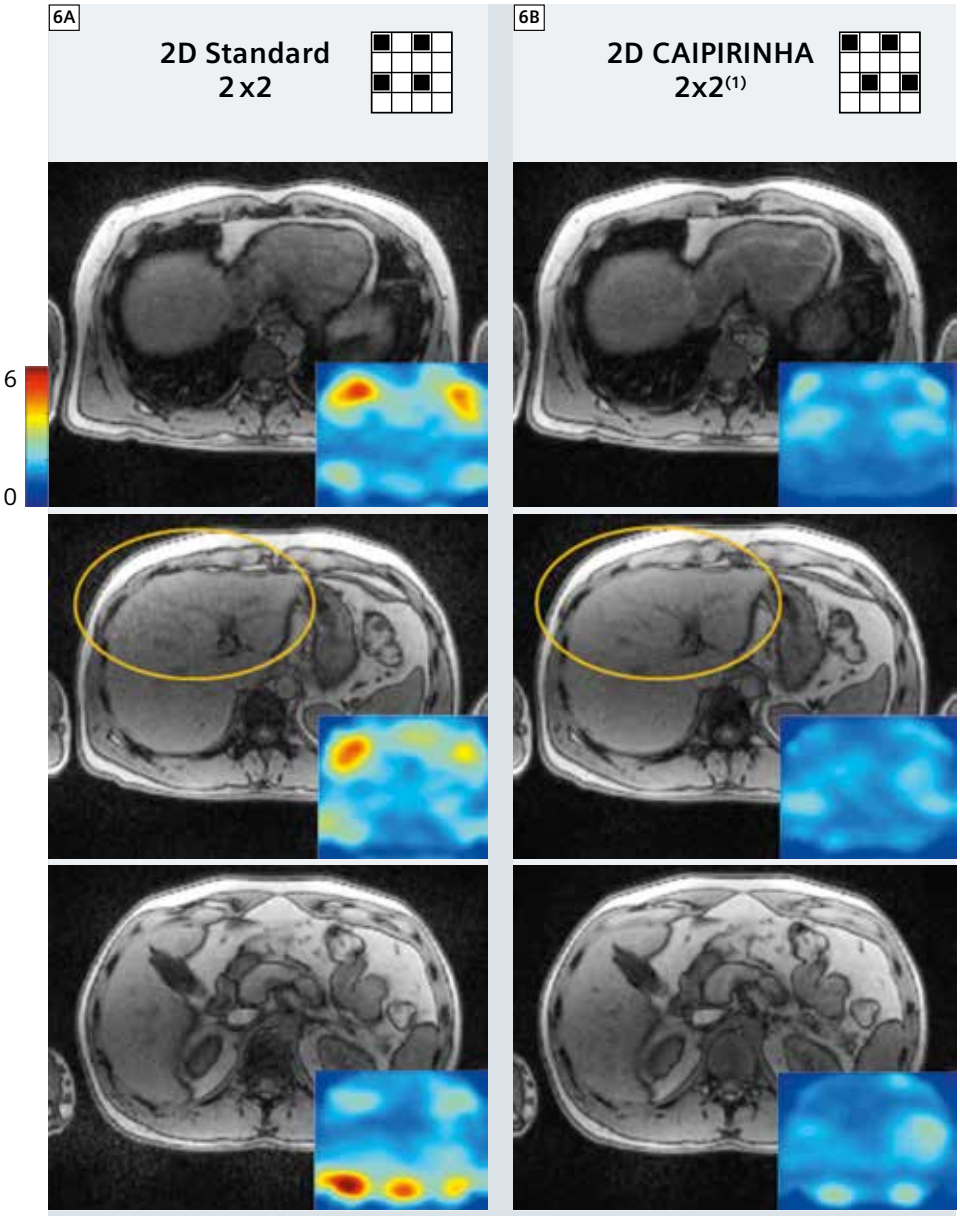


5 Procedure of generating 2D CAIPIRINHA sampling patterns for a given total acceleration factor, here af = 4. All possible sampling schemes can be represented by an af x af elementary cell with af sampling positions to fill. For each undersampling rate in the k_y direction (af_y) multiple patterns can 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.

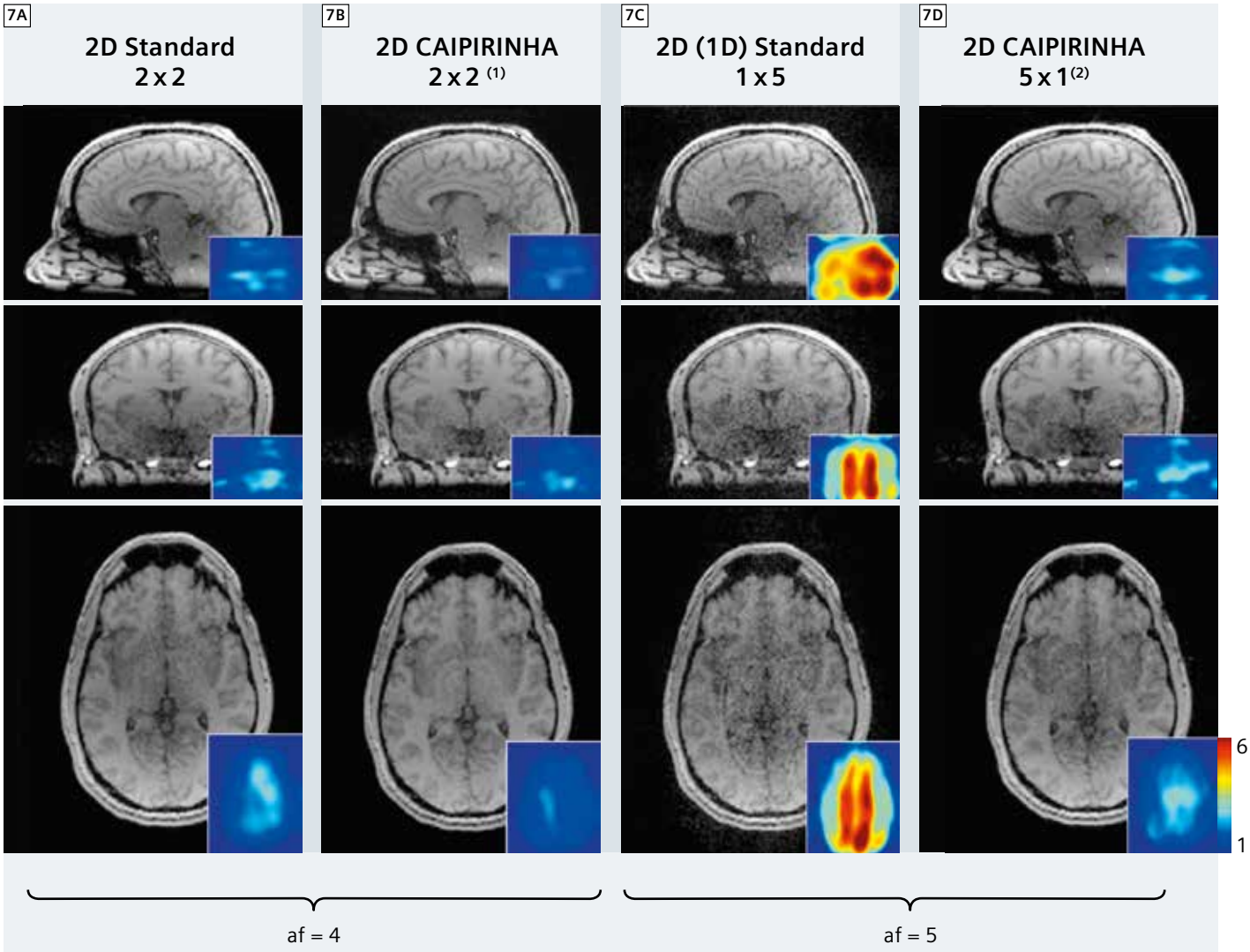


6 In vivo liver example; volunteer: Compared are GRAPPA reconstructions (3 example slices) derived from two different reduction schemes **(6A)** Standard 2x2 and **(6B)** 2D CAIPIRINHA 2x2(1). In addition, the corresponding GRAPPA g-factor maps are displayed. In the indicated region, the SNR benefit 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.

fully applied so far to simultaneous multislice imaging (MS-CAIPIRINHA) [6] and 3D imaging where data reduction can be carried out in two phase-encoding directions (2D CAIPIRINHA) [6]. 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 [7]. The following provides a brief overview of 2D CAIPIRINHA.

Improving parallel imaging performance with 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 strategy has shown the potential to allow for higher total image accelerations compared to undersampling schemes restricted to only one direction [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



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).

sampling positions on so-called 'sheared grids,' which form periodic lattices [9] 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 multislice 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 [6]. 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 [10].
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)
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 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 multislice imaging (MS-CAIPIRINHA) and 3D imaging (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 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 that 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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References

1 Pruessmann KP, Weiger M, Scheidegger B, Boesiger P. SENSE: sensitivity encoding for fast MRI. Magn Reson Med 1999; 42:952-962.

2 Griswold MA, Jakob PM, Heidemann RM, Nittka M, Jellus V, Wang J, Kiefer B, Haase A. GeneRalized Autocalibrating Partially Parallel Acqusitions (GRAPPA). Magn Reson Med 2002; 47:1202-1210.

3 Weiger M, Pruessmann KP, Boesiger P. 2D SENSE for faster 3D MRI. MAGMA. 2002 Mar; 14(1):10-9.

4 Larkman DJ, Hajnal JV, Herlihy AH, Coutts GA, Young IR, Ehnholm G. Use of multicoil arrays for separation of signal from multiple slices simultaneously excited. J Magn Reson Imaging. 2001 Feb; 13(2):313-7.

5 Breuer FA, Kannengiesser SA, Blaimer M, Seiberlich N, Jakob PM, Griswold MA. General formulation for quantitative G-factor calculation in GRAPPA reconstructions. Magn Reson Med. 2009 Sep;62(3):739-46.

6 Breuer FA, Blaimer M, Mueller MF, Seiberlich N, Heidemann RM, Griswold MA, Jakob PM. Controlled aliasing in volumetric parallel imaging (2D CAIPIRINHA). Magn Reson Med. 2006 Mar;55(3):549-56.

7 Breuer FA, Moriguchi H, Seiberlich N, Blaimer M, Jakob PM, Duerk JL, Griswold MA. Zigzag sampling for improved parallel imaging. Magn Reson Med. 2008 Aug;60(2):474-8.

8 Blaimer M, Breuer FA, Seiberlich N, Mueller MF, Heidemann RM, Jellus V, Wiggins G, Wald LL, Griswold MA, Jakob PM. Accelerated volumetric MRI with a SENSE/GRAPPA combination. J Magn Reson Imaging. 2006 Aug;24(2):444-50.

9 Willis NP and Bresler Y. Optimal scan design for time varying tomographic imaging {II}: Efficient design and experimental validation. IEEE Trans. Image Processing, 1995 May; 4: 654-666.

10 Wu B, Millane RP, Watts R, Bones PJ. Improved matrix inversion in image plane parallel MRI. Magn Reson Imaging. 2009 Sep;27(7):942-53.

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