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Selection and Evaluation of Optimal Two-Dimensional CAIPIRINHA Kernels Applied to Time-Resolved Three-Dimensional CE-MRA

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Abstract

Purpose—To develop and validate a method for choosing the optimal two-dimensional CAIPIRINHA kernel for subtraction contrast-enhanced MR angiography (CE-MRA) and estimate the degree of image quality improvement versus that of some reference acceleration parameter set at R 8.

Methods—A metric based on patient-specific coil calibration information was defined for evaluating optimality of CAIPIRINHA kernels as applied to subtraction CE-MRA. Evaluation in retrospective studies using archived coil calibration data from abdomen, calf, foot, and hand CE-MRA exams was accomplished with an evaluation metric comparing the geometry factor (g-factor) histograms. Prospective calf, foot, and hand CE-MRA studies were evaluated with vessel signal-to-noise ratio (SNR).

Results—Retrospective studies show g-factor improvement for the selected CAIPIRINHA kernels was significant in the feet, moderate in the abdomen, and modest in the calves and hands. Prospective CE-MRA studies using optimal CAIPIRINHA show reduced noise amplification with identical acquisition time in studies of the feet, with minor improvements in the hands and calves.

Conclusion—A method for selection of the optimal CAIPIRINHA kernel for high (R 8) acceleration CE-MRA exams given a specific patient and receiver array was demonstrated. CAIPIRINHA optimization appears valuable in accelerated CE-MRA of the feet and to a lesser extent in the abdomen.

Keywords

CAIPIRINHA; parallel imaging; optimization; acceleration

INTRODUCTION

Two-dimensional (2D) parallel imaging methods such as those based on 2D SENSE (1) or volumetric generalized autocalibrating partially parallel acquisitions (GRAPPA) (2) use means for reduction of the number of acquired samples across the 2D phase-slice (k_Y – k_Z) encoding plane of three-dimensional Fourier transform (3DFT) acquisition (1–3).

The 2D Controlled Aliasing in Parallel Imaging Results IN Higher Acceleration (2D CAIPIRINHA) (3) is a modification of the 2D SENSE undersampling kernel where the directionalities and patterns of the aliases in the accelerated image are moved off of the principal phase-slice encoding axes. This process can potentially better disperse a larger part of the aliases onto otherwise unoccupied regions of the image, reducing the overall number of aliases in the reconstructed image. As the acceleration R increases, the number of possible CAIPIRINHA kernels also increases, leading to increased uncertainty as to which kernel might work best in a given application. For example, for $R=8$, 10, and 12, there are 15, 18, and 28 possible kernels, respectively.

Given that many CAIPIRINHA kernels can provide acceleration factor R , it is of interest to determine which kernel provides the best or “optimal” performance for that R . This can be done by evaluating all candidate kernels according to some quantitative criterion. For example, the criterion could be the mean or the maximum geometry factor or “g-factor” (4) over the entire 3D volume. The kernel which provides the minimum value of either of these criteria is chosen as optimal. This could further be specified as the mean g-factor in a limited region of the volume such as for a specific organ as recently applied in Stemkens et al (5), i.e., abdomen, or in a reduced field of view (FOV) like cardiac imaging. The focus of this work is on the use of high ($R \geq 8$) acceleration factors in subtraction contrast-enhanced MR angiography (CE-MRA). In our experience particularly noisy patches throughout the volume interfere with interpretation, and it is desirable to reduce or eliminate such patches. Moreover, use of the mean g-factor over the volume as an optimization criterion provides inadequate differentiation between acceleration parameter sets which provide noticeably different performance.

Recently several works (4,6,7) have studied how CAIPIRINHA can provide improved performance in accelerated acquisition. The work of Breuer et al (4) used low resolution g-factor maps to show the optimal pattern at $R=4$ in acquisitions of the brain as evaluated by the average g-factor. In Deshpande et al (6), the abdomen of a subject was imaged at several possible CAIPIRINHA kernels for $R=3$ and 4 and the optimal kernel found of those tested. In Wright et al (7) a kernel determined using results from a trial set of subjects was selected which was shown to consistently provide improvement at $R=4$ versus standard $R=2 \times 2$ 2D SENSE. Although these works represent advances in how to best implement CAIPIRINHA, extension to higher acceleration factors, allowance for other imaging applications, and accommodation of patient-to-patient variability was desired.

The purpose of this work is multi-fold. First, we define a specific optimization criterion for the application of CAIPIRINHA to subtraction 3D CE-MRA and show how the optimal CAIPIRINHA kernel based on this criterion can be selected prospectively in a manner similar to that used for Acceleration Apportionment (8). Second, we describe a methodology which allows prediction of whether or not the optimum kernel provides an improvement in performance compared with some reference 2D SENSE acceleration. Finally, we apply the methodology to 3D CE-MRA of multiple anatomic regions to identify those situations in which CAIPIRINHA optimization appears to be particularly useful. This work is all done with acceleration factors $R \geq 8$. Also, as described in Weavers et al (8) for Acceleration Apportionment, the implementation allows determination of the optimum kernel within

seconds of acquisition of the coil sensitivity images, enabling immediate use in the accelerated acquisition.

METHODS

Selection of Optimal CAIPIINHA Kernel

In SENSE-type parallel imaging the g-factor can be used to evaluate the suitability of a given acceleration factor based on the fully sampled coil sensitivity information (9). For 2D CAIPIRINHA application the g-factor formalism also applies and similarly may be used to evaluate parallel imaging performance. Given the specific subject, receiver coil array and acquired coil sensitivity images for the 3D volume, we have previously shown how this g-factor information may be used in choosing optimal 2D SENSE acceleration parameters for application to CE-MRA (8).

The process of estimating the optimum acceleration parameters requires a balance between susceptibility to noise in the calibration data, and sensitivity to subtle quality reduction as indicated by increased g-factors. For this reason a multi-step process has been devised to assign a scalar quality metric to a given CAIPIRINHA kernel. First, the 3D g-factor map of the trial kernel is calculated, with the regions of the FOV outside the imaged object excluded from calculation as determined by masking (9). Next, to reduce the impact of calibration data noise, the 3D g-factor volume is summed along the frequency encode axis, resulting in a 2D image while maintaining resolution in the phase encoding directions. This summation reduces the sensitivity of the metric to outlier points of high noise which often occur along the edges of the air/tissue mask boundary. Finally, the maximum of this 2D image is chosen as the quality metric representative of the entire volume for the CAIPIRINHA kernel under consideration. The maximum of this image is chosen in order that a large g-factor-based image quality degradation is detected without regard for position in the phase encode plane, removing the blunting effect of the summation. The quality metric output by this process predicts relative image quality between two acquisitions identical but for acceleration parameters, as such is not an absolute measure of clinical image quality. This process is repeated over all plausible trial kernels, and the kernel that minimizes the quality metric is chosen.

This process, diagrammed in Figure 1, has been implemented on a GPU system interfaced directly to the MRI scanner (10). The time required to compute the performance metric for a high-resolution 3D dataset using 16 receiver array elements at R=12 is approximately 65 ms per trial CAIPIRINHA kernel. Repeating the process for all possible kernels, determining the optimal kernel, and loading it into the accelerated pulse sequence can be done within several seconds.

The criteria we have chosen for the selection of the CAIPIRINHA kernel are expected to result in a superior quality image in large part due to penalizing kernels with high g-factors by means of the maximum operation. In subtracted CE-MRA, a maximum intensity projection highlights any area of localized increased noise amplification, and, therefore, penalizing high g-factors reduces incidence of these regions. The reduction of influence of calibration data noise through the summation operation ensures that errantly included pixels

on the edge of the object contributing only noise to the image do not contribute unduly to the chose CAIPIRINHA kernel.

Estimation of Change in Image Quality

As opposed to a metric to be used for the selection of the optimal acceleration parameter set (8) or CAIPIRINHA kernel, it is of interest to generate a metric to predict whether the optimal set provides a significant improvement in image quality versus that provided by some reference parameter set. The selection metric is not suitable for this purpose as it was tuned to penalize small regions of highly degraded image quality, rather than to describe overall image quality. Because improvement in image quality is largely seen as a reduction of noise in particularly noisy regions of the image, a scalar value was developed to represent the reduction of high g-factors as allowed by the trial acceleration parameter set relative to some reference acceleration. Although the average of the 3D g-factor volume can be used (4), it was desired that the image quality comparison metric be sensitive mainly to the higher g-factor values.

The estimate of image quality improvement of a candidate acceleration parameter set versus a reference is based on the histogram of g-factors over the 3D volume. This is illustrated in Figure 2. Figure 2a is an image of the g-factor for an axial partition taken from a 3D CE-MRA exam of the feet acquired with 2D SENSE acceleration $R_Y \times R_Z = 2 \times 4 = 8$. This acceleration parameter set is defined in this example as the reference. Other acquisition parameters for this study are indicated in Table 1. Figure 2c is a plot of the histogram of the entire masked 3D g-factor image from which Figure 2a was selected. Figure 2b shows the g-factor image of the same slice as (a), but formed using a candidate CAIPIRINHA kernel, 2×4 (2) ($(R_Y=2) \times (R_Z=4)$ (Kernel offset=2)) in this example, which provides the same $R=8$ overall acceleration as Figure 2a and which used the same coil calibration dataset as Figure 2a. As suggested by the reduced g-factors in Figure 2b, the candidate kernel is expected to provide some improved performance. The corresponding g-histogram of the volume is shown in Figure 2d.

Starting with the g-factor histograms of the reference (Fig. 2c) and the candidate (Fig. 2d), the image quality improvement metric is determined by first taking the difference between histograms, reference – candidate, shown in Figure 2e. The shape of Figure 2e is characteristic of that found in the overall analysis. For an acquisition in which the candidate kernel outperforms the reference, one observes positive values at the largest g-factor values in the difference histogram and negative values near unity, reflecting an overall shift or reduction of the g-factor from larger to smaller values across the entire volume, as is generally desired. To quantify this overall shift, the area under the difference curve from the highest occurring zero crossing to the largest g-factor with nonzero bin count was calculated, with each histogram count weighted by the g-value for that bin, resulting in a g-factor weighted area (GWA). Dividing the GWA by the total number of voxels in the masked 3D imaging volume normalizes for image matrix size and generates the desired metric for how much the g-factor-based noise is reduced. This is defined as the histogram improvement (HI) factor, with positive values indicating improvement with optimization. Additionally, the average g-factors for both the reference and trial kernels were calculated,

and the trial average subtracted from the reference average to generate the average g-factor, $\bar{g}$, for comparison. This $\bar{g}$ would also be positive for improvement using the trial kernel.

As previously stated, the motivation for creating the HI metric was to have a value more sensitive to the reduction of high g-factors. We can express this by assuming $p(g_i)$ to be the histogram of the g-factors of the reference acceleration parameter set, and $p'(g_i)$ that of the candidate parameter set. Assume that both are normalized by N, the number of pixels in the masked 3D volume. These can each then be regarded as probability density functions. The mean values, m and m' can each be computed, and the difference in mean g-factors, average g-factor, is then

$$\begin{aligned}\Delta\bar{g} &= m' - m = \frac{1}{N} \sum_{i=1}^{\infty} g_i(p'(g_i)) - \frac{1}{N} \sum_{i=1}^{\infty} g_i(p(g_i)) \\ &= \frac{1}{N} \sum_{i=1}^{\infty} g_i(p'(g_i) - p(g_i))\end{aligned}\quad [1]$$

where the sums run over all possible g-factor values. Defining $\Delta p(g_i) = p'(g_i) - p(g_i)$, the last sum can be split into two:

$$\Delta\bar{g} = \frac{1}{N} \sum_{t=1}^{t-1} g_t \Delta p(g_t) + \frac{1}{N} \sum_{i=t}^{\infty} g_i \Delta p(g_i) \quad [2]$$

where g_t is the first g-value occurring just beyond the highest zero crossing with positive bin count and is identified in Figure 2e. The HI factor is simply the second of these sums. Within this summation the difference $\Delta p(g)$ is positive for all values of g-factor, given that there is some improvement afforded by the candidate CAIPIRINHA kernel. In this case, and noting that the sum of $\Delta p(g)$ overall g-factors must be zero, the first sum tends to be negative. Although not a rigorous proof, this suggests that for g-factor histograms such as the ones in Figure 2 that are typical for the case when a candidate kernel reduces high g-factor values, HI tends to be larger than $\bar{g}$.

Retrospective Studies

Multi-coil calibration data from 41 archived CE-MRA studies from several anatomic regions were analyzed using the aforementioned HI and average g-factor metrics. The regions used were the abdomen (11) (n=10 studies, reference 2D SENSE acceleration $R=R_Y \times R_Z=4 \times 2=8$), bilateral calves (12) (n=9, $R=4 \times 2=8$), bilateral hands (13) (n=9, $R=4 \times 2=8$), bilateral feet (13) (n=8, $R=2 \times 4=8$), and bilateral hands (14) (n=5, $R=6 \times 2=12$). Linear receiver coils comprised of 8 to 12 elements were used, placed circumferentially around the targeted region in all cases.

Each individual dataset was subjected to the described optimization process resulting in an optimal CAIPIRINHA kernel at the chosen acceleration factor R. The predicted performance of the optimal kernel was then compared with the reference 2D acceleration using the HI and $\bar{g}$ metrics. Additionally the optimal 2D SENSE acceleration pair (R_Y, R_Z) was determined using Acceleration Apportionment and compared with the reference acceleration as well as to the optimal CAIPIRINHA kernel.

Prospective CE-MRA Studies

Prospective CE-MRA studies were undertaken to assess the methodology for selecting the optimal CAIPIRINHA kernel. Studies were done in four subjects: two bilateral studies of the feet ($R=8$), one bilateral study of the calves ($R=12$), and one unilateral study of a hand ($R=8$).

For each study a healthy volunteer was recruited for two separate visits following an IRB approved protocol with minimum contrast clearance time of 48 hours between visits, one visit done with reference 2D SENSE acceleration parameters previously reported (13,15), and one with the optimal CAIPIRINHA kernel. A linear circumferential receiver coil specific to each anatomy was used for each of the different extremities scanned (13,15,16). At each visit 20 mL of gadobenate dimeglumine (MultiHance; Bracco Diagnostics; Princeton, NJ) injected at 3 mL/s followed by 20-mL saline flush injected at 3 mL/s was administered intravenously into the arm by means of power injector (Spectris Solaris; Medrad Inc; Indianola, PA). Each acquisition was performed at 3 Tesla (T) with the CAPR view-shared time-resolved CE-MRA sequence (17) at a view-share factor of 4. Table 1 details additional scan parameters.

The two contrast-enhanced scans for each subject were compared by the HI and metrics and by vessel signal-to-noise (SNR) measurements. For vessel SNR, 10 or more identical vessel segments in each of the two studies were identified for comparison. Each vessel segment was primarily longitudinally oriented so that its cross section was contained in an axial plane, corresponding to the plane of 2D acceleration. The signal was measured in the axial plane using a 3×3 pixel ROI, and the standard deviation of a nearby patch of at least 200 pixels of subtracted unenhanced tissue was measured as noise. The ratio of the vessel intensity to the background tissue standard deviation was then calculated and reported as SNR. These results of the reference and optimized accelerations were compared using the paired Student's t-test (18).

RESULTS

Retrospective Studies

Figure 3 shows the HI results. Individual plots (Figs. 3a–e) distinguish the retrospective studies by anatomic region and acceleration factor, while results for all four prospective studies are shown together in Figure 3f. Each figure plots the mean g-factor change $\bar{g}$ versus HI. Each closed square represents the values for the optimal CAIPIRINHA kernel for an individual study, and the linked open circle indicates the performance of Acceleration Apportionment for that study. Note the change in axes scales from Figures 3a–c, d, and e,f.

Figure 4 summarizes which CAIPIRINHA kernels and apportioned (R_Y , R_Z) values were selected as optimal for all 36 retrospective studies done with $R=8$. Note the variability of the optimal kernel from one anatomic region to the next as well as the variability amongst subjects for a given region. The k-space sampling pattern of each kernel that was selected as optimal in any study is shown as an inset. The scatter plot (Fig. 4b) shows the mix of (R_Y , R_Z) values selected by Acceleration Apportionment for these same 36 $R=8$ studies.

Prospective Studies

Figure 5 shows results from the prospective CE-MRA feet studies. Figures 5a,b compare sagittal MIPs of the peak-contrast time frame of the left foot of Study # 1 using the reference 2D SENSE acceleration (Fig. 5a) and the optimal CAIPIRINHA kernel (Fig. 5b), 2×4 (2). Vessel SNR measurements (Fig. 5c) show significant improvement ($P=0.0014$) for the optimal CAIPIRINHA sampling. The measured $\bar{g}$ and HI values comparing acquisitions illustrated in Figures 5a,b were 0.17 and 0.31, respectively. Figures 5d,e show targeted zoomed MIPs of Feet Study #2 using the reference 2D SENSE acceleration (Fig. 5d) and the optimal CAIPIRINHA kernel (Fig. 5e), 1×8 (3) with $\bar{g}$ and HI values comparing acquisitions illustrated in Figures 5d,e of 0.10 and 0.17. Vessel SNR measurements (Fig. 5f) show a nonsignificant, but trending improvement in vessel SNR from reference to optimal acceleration.

Figure 6, shows results from the prospective CE-MRA study of the calves: coronal MIPs from the peak-contrast time frame acquired with reference 2D SENSE $R=12$ ($R_Y=6$; $R_Z=2$) (Fig. 6a), and the optimal CAIPIRINHA 6×2 (1) (Fig. 6b). Figure 6c is a comparison of SNR measurements in 19 vessel segments. The measured $\bar{g}$ and HI values were 0.0048 and 0.01, respectively.

DISCUSSION

We have developed a g-factor-based optimality criterion and a method to automatically select the optimal CAIPIRINHA kernel on a patient- and anatomy-specific basis for highly accelerated ($R \geq 8$) 3D acquisitions as used for subtraction CE-MRA. We have further defined a histogram improvement “HI” metric, also based on the g-factor, which attempts to predict the degree of improvement in image quality resulting from use of optimal acceleration parameters versus some reference. The method was applied to 41 retrospective CE-MRA studies from four anatomic regions. Compared with the reference accelerations used, which were all shown previously as having provided diagnostic quality results, the comparisons suggest that CAIPIRINHA optimization consistently provides significant improvement in SNR in CE-MRA of the feet, consistent but moderate improvement in the abdomen, and modest or negligible improvement in CE-MRA of the calves and hands. Furthermore, we have shown patient-to-patient variability in the optimal kernel at a given acceleration, indicating that the optimal kernel may vary between patients even for the same anatomy and receiver coil array. This work extends previous work (4,6,7) by studying multiple anatomic areas, multiple receiver coil configurations, and higher R factors in not only retrospective but also prospective studies.

Results from the prospective CE-MRA studies are consistent with these predicted trends, at least for the feet and calves. Feet Study #1 (HI=0.317) showed significantly improved vessel-to-background SNR of optimized CAIPIRINHA (Fig. 5c). Results from Feet Study #2, which had a reduced HI (0.17) versus Study #1, demonstrated improved visual image quality for CAIPRINHA but marginally nonsignificant improvement in SNR. Results from the Calves Study (Fig. 6) demonstrated equivalent visual image quality and SNR for a HI metric near zero (HI=0.01). This suggests that a trial acceleration parameter set having an HI value exceeding 0.37, as noted in Feet Study #1, should be expected to provide significantly

improved vessel SNR versus the reference used. As HI diminishes below this value, the likelihood of improvement does as well.

The metrics for the selection of the optimal acceleration parameters as well as for the comparison of the parameter sets are both calculated from the 3D g-factor map. By design, the g-factor metric used for selecting the optimal acceleration parameters minimizes peak g-factors so as to prevent any areas of highly degraded image quality. In contrast to the selection metric, the HI metric reflects the overall shift in high g-factors from reference to lower g-factors in the optimal acceleration. It can be seen in Figures 3b and 3d, for example, that the $\bar{g}$ metric ascribes improvement in many cases when the HI metric indicates otherwise. We note that as g_t approaches $g_{\max}$ only a small set of high g-factors are reduced, and the HI is likely to be small. Also if $p(g)$ frequently crosses zero for large g-factor values, then $p(g) = p'(g)$, again suggesting little improvement.

A byproduct of this work was a comparison of the expected performance of CAIPIRINHA and Acceleration Apportionment (AA). The anatomic region for which the predicted improvement in SNR was greatest was the feet (Fig. 4e) in which case the HI was at least 0.17 for all eight retrospective studies for CAIPIRINHA, and for which CAIPIRINHA was predicted to have superior performance to AA. AA also was predicted to provide improvement but not to the same degree. We believe the large improvement for both techniques is due to the 3D FOV used for feet imaging having considerable space with zero signal in which to flexibly place aliases (8) and effectively use coil sensitivity maps in performing the unaliasing step, and CAIPIRINHA more effectively does this than AA. The anatomic region showing the next greatest improvement was the abdomen (Fig. 4c), which has the lowest amount of empty space in the 3D FOV of the regions considered. We speculate that the ability to better disperse the aliases with CAIPIRINHA versus 2D SENSE (3) allowed for the g-factor reduction. Of the 10 retrospective studies of the abdomen, CAIPIRINHA had a higher HI versus AA in four, lower in four, and the two were equivalent, within .02 of each other, in the other two. Finally, the regions of the calves and hands showed less improvement according to the HI metric, which suggests that the reference accelerations were already well suited to those regions.

These results are specific to the receiver coil arrays and acceleration factors used. We note that it is possible to perform retrospective studies similar to that done here to estimate the potential benefit of optimal CAIPIRINHA kernel selection for other imaging situations.

Note that when the $p(g)$ and HI metrics disagree, as in a few cases in Figure 3, it can be ascribed to the interaction of receiver coil array and patient being unsuitable for the method. It is occasionally observed that incomplete coverage of the FOV with the receiver arrays can invalidate the assumptions that go along with this work. Note especially the outlier in Figure 3d: the edges of the FOV are acquired using the methods in Haider et al (12) are subject to significant signal fall-off at the S/I edges of the FOV. A sensible alteration of the cost function in this situation would be to lower the weighting of those edges in the generation of the cost function. Each prospective study presented has full FOV coverage from the receiver arrays.

The method has been implemented such that the optimal CAIPIRINHA kernel can be determined within several seconds of acquisition of the coil sensitivity data, allowing practical use of an optimal patient-specific kernel. Even if the optimal CAIPIRINHA kernel minimally improves image quality, this implementation is fast and robust enough that it may be widely applied with no deleterious effects.

The HI metric was specifically developed to characterize the relative improvement in g-factors between optimal and reference accelerations. It may be the case that an alternate metric will prove more applicable in some situations. For example, use of the fully projected g-factor selection metric might not be desired in cardiac imaging where the area of interest is generally central within the FOV (19).

This study was applied to precalibrated SENSE-type imaging, and uses the fully sampled coil sensitivity information before the accelerated exam begins. Similar methodology conceivably may be applicable to data-driven parallel imaging methods such as GRAPPA (20,21), and a real-time implementation might use the first part of an accelerated scan to guide what form of GRAPPA kernel to use for the remainder of that scan.

A limitation of this study is that only a few prospective contrast-enhanced comparisons of the technique were performed. Application to more anatomic regions in a larger number of subjects with different receiver coil arrays and at different targeted parallel imaging acceleration rates would allow for additional characterization. However, CAIPIRINHA has already been demonstrated to reduce parallel imaging artifacts for lower ($R = 4$) accelerations (22,23), and standard 2D SENSE acceleration has been shown to work at higher ($R = 8$) accelerations (12,15,24,25). Based on these two it seems justifiable to assert that a judicious choice of acceleration kernel at high acceleration will result in good images using the CAIPIRINHA technique.

CONCLUSIONS

An optimality criterion based on the 3D g-factor map has been defined for selecting and evaluating the optimal CAIPIRINHA kernel for specific patient, anatomic region, and receiver coil array for acceleration factors $R = 8$ as used for subtraction CE-MRA. Retrospective studies suggest that CAIPIRINHA can provide significant improvement in SNR in CE-MRA of the feet, moderate improvement in the abdomen, and minor improvement in the calves and hands versus currently used 2D SENSE accelerations. Prospective CE-MRA studies are consistent with these findings. The method is general and is applicable to other applications of accelerated imaging.

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References

1. Weiger M, Pruessmann KP, Boesiger P. 2D SENSE for faster 3D MRI. *MAGMA*. 2002; 14:10–19. [PubMed: 11796248]

2. 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 Mag Reson Imaging*. 2006; 24:444–450.
3. 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; 55:549–556. [PubMed: 16408271]
4. Breuer, FA.; Blaimer, M.; Mueller, MF.; Heidemann, RM.; Griswold, MA.; Jakob, PM. Finding the optimal sampling pattern in 2D parallel imaging for a given receiver coil configuration. Proceedings of the 13th Annual Meeting of ISMRM; Miami, Florida, USA. 2005; p. Abstract 2665
5. Stemkens, B.; Tijssen, R.; Andreychenko, A.; Crijns, S.; Sbrizzi, A.; Langendijk, J.; van den Berg, C. Optimizing CAIPIRINHA multi-band acquisition scheme for 2D multi-slice experiments in the abdomen. Proceedings of the 22nd Annual Meeting of ISMRM; Milan, Italy. 2014. p. Abstract 0645
6. Deshpande, V.; Nickel, D.; Kroeker, R.; Kannengiesser, S.; Laub, G. Optimized Caipirinha acceleration patterns for routine clinical 3D imaging. Proceedings of the 20th Annual Meeting of ISMRM; Melbourne, Australia. 2012; p. Abstract 0104
7. Wright KL, Harrell MW, Jesberger JA, Landeras L, Nakamoto DA, Thomas S, Nickel D, Kroeker R, Griswold MA, Gulani V. Clinical evaluation of CAIPIRINHA: comparison against a GRAPPA standard. *J Mag Reson Imaging*. 2014; 39:189–194.
8. Weavers PT, Borisch EA, Johnson CP, Riederer SJ. Acceleration apportionment: a method of improved 2D SENSE acceleration applied to 3D contrast-enhanced MR angiography. *Magn Reson Med*. 2014; 71:672–680. [PubMed: 23450817]
9. Pruessmann KP, Weiger M, Scheidegger MB, Boesiger P. SENSE: sensitivity encoding for fast MRI. *Magn Reson Med*. 1999; 42:952–962. [PubMed: 10542355]
10. Borisch, EA.; Weavers, PT.; Riederer, SJ. GPU-enabled individualized acceleration apportionment for SENSE and CAIPIRINHA. Proceedings of the 21st Annual Meeting of ISMRM; Salt Lake City, Utah, USA. 2013; p. Abstract 2224
11. Mostardi PM, Glockner JF, Young PM, Riederer SJ. Contrast-enhanced MR angiography of the abdomen with highly accelerated acquisition techniques. *Radiology*. 2011; 261:587–597. [PubMed: 21900616]
12. Haider CR, Glockner JF, Stanson AW, Riederer SJ. Peripheral vasculature: high-temporal- and high-spatial-resolution three-dimensional contrast-enhanced MR angiography. *Radiology*. 2009; 253:831–843. [PubMed: 19789238]
13. Haider CR, Riederer SJ, Borisch EA, et al. High temporal and spatial resolution 3D time-resolved contrast-enhanced magnetic resonance angiography of the hands and feet. *J Magn Reson Imaging*. 2011; 34:2–12. [PubMed: 21698702]
14. Haider, CR.; Glockner, JF.; Stanson, AW.; Johnson, CP.; Riederer, SJ. High temporal and spatial resolution 3D CE-MRA of the peripheral vasculature using 12-fold 2D SENSE. Proceedings of the 17th Annual Meeting of ISMRM; Honolulu, Hawaii, USA. 2009; p. Abstract 2734
15. Weavers, PT.; Cline, CC.; Johnson, CP.; Rossman, PJ.; Hulshizer, TC.; Riederer, SJ. A 16 element phased array surface coil for time-resolved CE-MRA at SENSE accelerations up to 12. Proceedings of the 20th Annual Meeting of ISMRM; Melbourne, Australia. 2012; p. Abstract 2648
16. Weavers, PT.; Hulshizer, TC.; Rossman, PJ.; Young, PM.; Riederer, SJ. Unilateral 8ch receiver array for ultra high resolution time-resolved 3D CE-MRA of the hand. Proceedings of the 21st Annual Meeting of ISMRM; Salt Lake City, Utah, USA. 2013; p. Abstract 4395
17. Haider CR, Hu HH, Campeau NG, Huston J III, Riederer SJ. 3D high temporal and spatial resolution contrast-enhanced MR angiography of the whole brain. *Magn Reson Med*. 2008; 60:749–760. [PubMed: 18727101]
18. Rosner, B. Fundamentals of biostatistics. Boston: Brooks/Cole, Cengage Learning; 2011. p. xviip. 859
19. Hu X, Parrish T. Reduction of field of view for dynamic imaging. *Magn Reson Med*. 1994; 31:691–694. [PubMed: 8057824]
20. Griswold MA, Jakob PM, Heidemann RM, Nittka M, Jellus V, Wang J, Kiefer B, Haase A. Generalized autocalibrating partially parallel acquisitions (GRAPPA). *Magn Reson Med*. 2002; 47:1202–1210. [PubMed: 12111967]

21. Breuer FA, Kannengiesser SAR, Blaimer M, Seiberlich N, Jakob PM, Griswold MA. General formulation for quantitative G-factor calculation in GRAPPA reconstructions. *Magn Reson Med*. 2009; 62:739–746. [PubMed: 19585608]
22. Riffel P, Attenberger UI, Kannengiesser S, Nickel MD, Arndt C, Meyer M, Schoenberg SO, Michaely HJ. Highly accelerated T1-weighted abdominal imaging using 2-dimensional controlled aliasing in parallel imaging results in higher acceleration: a comparison with generalized autocalibrating partially parallel acquisitions parallel imaging. *Invest Radiol*. 2013; 48:554–561. [PubMed: 23462674]
23. Michaely HJ, Morelli JN, Budjan J, Riffel P, Nickel D, Kroeker R, Schoenberg SO, Attenberger UI. CAIPIRINHA-Dixon-TWIST (CDT)-volume-interpolated breath-hold examination (VIBE): a new technique for fast time-resolved dynamic 3-dimensional imaging of the abdomen with high spatial resolution. *Invest Radiol*. 2013; 48:590–597. [PubMed: 23538886]
24. Hadizadeh DR, von Falkenhausen M, Gieseke J, Meyer B, Urbach H, Hoogeveen R, Schild HH, Willinek WA. Cerebral arteriovenous malformation: Spetzler-Martin classification at subsecond-temporal-resolution four-dimensional MR angiography compared with that at DSA. *Radiology*. 2008; 246:205–213. [PubMed: 17951352]
25. Fenchel M, Doering J, Seeger A, Kramer U, Rittig K, Klumpp B, Claussen CD, Miller S. Ultrafast whole-body MR angiography with two-dimensional parallel imaging at 3.0 T: feasibility study. *Radiology*. 2009; 250:254–263. [PubMed: 19001152]

FIG. 1.

The workflow of optimal CAIPIRINHA imaging. The main addition is the optimization step, taking place after the acquisition of calibration data, and before the accelerated scan.

FIG. 2.

Schematic overview of the calculation of the HI quality metric. Show axial sections of the 3D geometry factor (g-factor map) for both the reference 2D SENSE acceleration (a) as well as the optimal CAIPIRINHA pattern (b). This happens to be data for Feet Study #1. (c,d) The g-factor histograms for those accelerations used in (a,b). (e) The difference of the histogram in (d) subtracted from the histogram in (c). Shaded in (e) is the area representing the number of voxels in which the g-factors have reduced from higher (for this study, approx. $g > 1.3$ at threshold index t) to lower levels. Each bin in the shaded area is multiplied by its center, and added together to generate the g-factor weighted area (GWA). This value is then used to calculate the HI by dividing GWA by the total number of voxels considered in the reconstruction. The HI metric is constructed such that a positive HI indicates an increase in image quality with optimal CAIPIRINHA versus a reference acceleration per prospective studies noted in Figures 5 and 6.

FIG. 3.

The library of retrospective studies (RS) plotted against the histogram improvement metric. The anatomic regions shown here are: hands at R=8 (**a**), hands at R=12 (**b**), abdomen (**c**), calves (**d**), and feet (**e**). Note difference in axes scales between (a-c), (d) and (e,f). (**f**) Four prospective studies are shown and labeled as to the anatomic region imaged. Note that positive HI and $\bar{g}$ values denote improvement with acceleration optimization. Each open circle represents an Apportioned acceleration, whereas closed squares represent an optimal CAIPIRINHA kernel. When the magnitude of the HI metric is small we note indistinguishable quality changes as shown in Figure 6 indicating there may be some minimum HI for improvement to be visible, below which variances in these metrics may be insignificant. Note outliers in panels (b,c,e) for Apportioned accelerations. Once per panel the cost function fails to produce minima which results in favorable metrics for the Apportionment technique: this is further described in the discussion section.

FIG. 4. Summary of optimal acceleration parameters for the 36 R=8 retrospective studies. Summary of optimal CAIPIRINHA kernels (**a**) and plot of optimal 2D SENSE acceleration pairs (R_Y, R_Z) (**b**) as directed by Acceleration Apportionment.

FIG. 5.

A comparison feet CE-MRA study of the same subject using reference SENSE (a,d) and optimal CAIPIRINHA (b,e) acceleration in two separate CE-MRA exams, both at $R=8$. Each image is a sagittal MIP of a single foot acquired with a 6.6 s frame time. Images compare selected full sagittal MIPs of SENSE and optimal CAIPIRINHA accelerated feet in Feet Study #1 (a,b), with showing vessel SNR measurements between the two (c). (d,e) Images are zoomed, reduced FOV images of a single foot from Feet Study #2, with (f) showing the vessel SNR measurements. Note how the area of patchy increased noise in the reference SENSE image (d, arrows) is substantially eliminated using optimal CAIPIRINHA (e).

FIG. 6.

A comparison of two CE-MRA calf studies of the same subject using reference acceleration (a) and optimal CAIPIRINHA acceleration (b) for $R=12$. (c) Vessel SNR measurements shown. Note minimal improvement in MIP image quality despite a significant improvement in vessel SNR.

Table 1Acquisition Parameters for Contrast-Enhanced MRA Comparison Studies^a

Anatomic region	Feet	Calves	Hand
FOV (cm ³)	30×19.8×24	42×33.6×13.2	30×15×7.2
Resolution (mm ³)	0.75×0.75×0.9375	1	0.7
2D SENSE acceleration	R = 8	R = 12	R = 8
2D Homodyne acceleration	R _{HD} = 1.80	R _{HD} = 1.88	R _{HD} = 1.73
Overall acceleration	R _{net} = 14.4	R _{net} = 22.6	R _{net} = 13.8
Coil array (reference)	8ch differential (13)	16ch calf (15)	8ch unilateral hand (16)
TR/TE (ms)	5.684/2.652	6.032/2.744	6.064/2.764
Update time/temporal footprint	6.66 s/24.35 s	3.57 s/12.61 s	2.79 s/9.49 s
Reference acceleration R _Y × R _Z	2 × 4	6 × 2	4 × 2
Optimal CAIPIRINHA R _Y × R _Z (Shift)	2 × 4 (2) (Study #1) 1 × 8 (3) (Study #2)	6 × 2 (1)	2 × 4 (2)

^a Note high (R = 8) parallel imaging combined with homodyne compensation.