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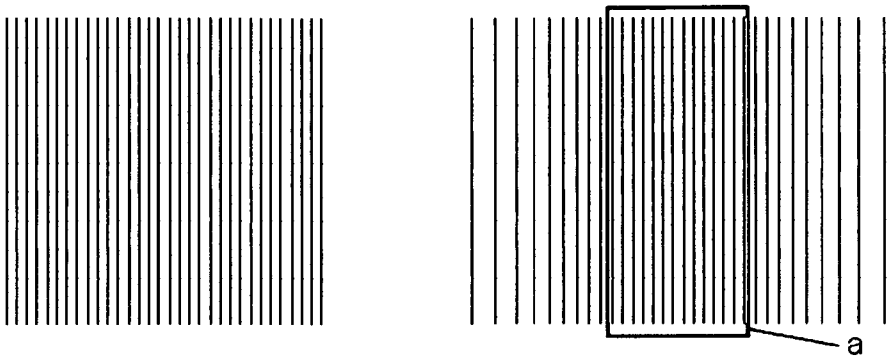
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(54) **Title:** PARALLEL MRI INVOLVING DENSITY WEIGHTING OR ACQUISITION WEIGHTING



(57) **Abstract:** A method of magnetic resonance imaging, comprising the step of sampling the k-space by means of the method of density weighted or acquisition weighted data acquisition such as density weighted phase encoding or acquisition weighted phase encoding, wherein in the sampling at least one region of the k-space is undersampled with respect to the Nyquist criterion. The reconstruction of the MR image is performed at least in part with the aid of a parallel MR imaging mediod and the influence in the image of the missing k-space data required to fulfil the Nyquist criterion is reduced.

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PARALLEL MRI INVOLVING DENSITY WEIGHTING OR ACQUISITION WEIGHTING

The present invention concerns a method of MR imaging (MR, magnetic resonance) according to the preamble of claim 1.

In MR imaging based on the phenomenon of nuclear magnetic resonance, the test object is exposed to a static homogeneous magnetic field, whereby the atom's nuclear spins, which at first are randomly oriented, align along the axis of the static magnetic field. The ordered nuclear spins are excited by magnetic high frequency fields. For image formation, the nuclear spin signal is recorded as a voltage which is induced into one or several coils under the influence of a suitable sequence or progression of high frequency and gradient pulses in the time domain. Finally, the image reconstruction in general is carried out by means of a Fourier transform of the time signals. The number, the time interval, the duration and the strength of the gradient pulses employed define the mode of sampling of the reciprocal so-called "k-space", which in turn determines the FOV (field of view) to be imaged as well as the image resolution.

In the advancement of the art of MR imaging methods, numerous efforts have already been made to shorten the image acquisition time and/or to improve the achievable spatial resolution.

To accelerate the imaging process, parallel MR techniques for example take advantage of the phenomenon that the spatial sensitivity profile of the receiver coils provide the nuclear spin signal with spatial information, which can be used for the image reconstruction. By using in parallel several separate receiver coils, each with a different sensitivity profile, and combining the respective nuclear spin signals detected, the acquisition time for an image can be reduced in comparison to a conventional reconstruction by a factor which is limited by the number of receiver coils and therr spatial arrangement.

Further, for MR images where the side lobes of the SRF (spatial response function) entail severe contamination, it has been proposed to use a k-space filter which enhances the data in the centre of the k-space. While this filtering reduces the side lobes of the SRS and thereby

the contamination from neighbour voxels, it also affects the spatial resolution and reduces the signal to noise-ratio (SNR). One approach to overcome this problem involves the use of density weighted or acquisition weighted imaging methods. This way, k-space filtering is already applied during data acquisition. If acquisition weighted methods are employed, the Nyquist criterion is generally met in the entire k-space. However, due to the greater number of acquisitions in the centre of the k-space and the need for a greater number of phase encoding steps for the same size of the FOV and the same spatial resolution as in an unweighted method, the minimum number of scans is restricted and thereby limited to the shortest recording duration. To overcome this problem, Greiser and von Kienlin have introduced the density weighted phase encoding method, which is described in more detail in the article "Greiser A, von Kienlin M., Efficient k-Space Sampling by Density-Weighted Phase-Encoding. Magn Reson Med 2003; 50: 1266 - 1275". In contrast to the acquisition weighted methods, this method is based on the variation of the density of the sampling points in the k-space. The acquisition weighted sampling is a density weighted sampling in which specifically the k-space density is varied by a variable number of acquisitions on a fixed k-space grid. The density weighted sampling has proven to be a highly efficient method of k-space sapling. The modification of the sampling scheme makes it possible to improve the shape of the SRF and thereby the localisation properties. In the kind of density weighted sampling presently known, some parts of the k-space violate the Nyquist criterion, which entails undersampling artefacts.

The object of the present invention is to suggest a method in which the above-mentioned drawbacks can at least partially be eliminated.

The object is achieved with the features of claim 1.

The method of magnetic resonance imaging comprises the step of sampling the k-space by means of the method of density weighted data acquisition, which includes acquisition weighted data acquisition, wherein in the sapling at least one region of the k-space is undersampled with respect to the Nyquist criterion. According to the invention, the reconstruction of the MR image is performed at least in part with the aid of a parallel MR imaging method. Thereby, in the image the influence of the missing k-space data required to

fulfil the Nyquist criterion is at least partly reduced. For example, the k-space data missing in a part of the undersampled region and required to meet the Nyquist criterion can be reconstructed or the image can be deconvoluted on a greater FOV.

The parallel MR imaging method replaces at least partially the very time-consuming spatial encoding. This way, only a fraction of the samples, generally phase encoding steps, must be performed, which yields an acceleration of the image acquisition while the full spatial resolution is maintained. In the acquisition weighted data acquisition, where otherwise the Nyquist criterion is fulfilled over the entire k-space, the sampling density is reduced and the violation of the Nyquist criterion at least in parts of the k-space is deliberately accepted. In the density weighted imaging as well, the Nyquist criterion is knowingly violated.

In one embodiment, in the parallel imaging method missing k-space lines are reconstructed for example by means of a SMASH- (Simultaneous acquisition of spatial harmonics), Auto-SMASH, (Autocalibrating simultaneous acquisition of spatial harmonics), VD-Auto-SMASH- (Variable density autocalibrating simultaneous acquisition of spatial harmonics), GRAPPA- (Generalized Autocalibrating Partially Parallel Acquisitions), or a GRAPPA-operator-algorithm.

In a further embodiment, in the parallel MR imaging method the reconstruction is performed in the real-space, for example with a SENSE-, Auto-SENSE-, mSENSE- (modified sensitivity encoding), T-SENSE- (Adaptive sensitivity encoding incorporating temporal filtering) oder PILS- (Partially parallel imaging with localized sensitivities) algorithm.

I yet another embodment, in the parallel MR imaging method the reconstruction is performed in the real-space as well as in the k-space, for example with the SPACE-RIP method (Sensitivity profiles from an array of coils for encoding and reconstruction in parallel).

In a further embodiment, the method of magnetic resonance imaging comprises a spectroscopic imaging method, for example chemical shift imaging.

In a further embodiment, the method comprises the density weighted and the acquisition weighted sampling, respectively, by means of phase encoding steps.

In a further embodiment, the method comprises the density weighted and the acquisition weighted sampling, respectively, not only by means of phase encoding steps, as for example by means of radial of spiral k-spacs sampling or sapling according to the PROPELLER (Periodically Rotated Overlapping ParallEL Lines with Enhanced Reconstruction) technique (Pipe JG. Motion correction with PROPELLER MRI: Application to head motion and free-breathing cardiac imaging. Magn Reson Med 1999; 42: 963:969).

In a further embodiment, the method comprises the density weighted and the acquisition weighted sampling, respectively, wherein the Nyquist criterion is violated in the entire k-space.

In a further embodiment, the method comprises an optimised density weighted and the acquisition weighted sampling mode, respectively, in case the signal intensity encoded in the k-space varies due to the imaging methods, as for example in RARE (rapid acquisition with relaxation enhancement), FSE (Fast spin echo), TSE (Turbo spin echo) oder EPI (Echo planar imaging) methods, or in case the noise in the reconstruction depends on the k-space density, for example a noise amplification factor g depending on an acceleration factor in parallel imaging.

In a further embodiment, the method comprises the density weighted and the acquisition weighted sampling, respectively, wherein not the data in the centre of the k-space are enhanced, as for example when the density distribution corresponds to a edge enhancement filter.

The method according to the invention combines the benefits of the density weighted and the acquisition weighted sampling mode, respectively, with the benefits of the parallel imaging methods. In comparison with imaging methods that make use of a uniform sampling mode, the method according to the invention at the same nominal spatial resolution and the

same number of scans yields a noticeable reduction of aliasing artefacts and, depending on the size of the specimen, an improved SNR (signal to noise-ratio).

The method according to the invention can be applied in one, two, or three spatial dimensions.

The method according to the invention can for example be used in conventional MR-imaging or spectroscopic MR-imaging, but also in MR angiography, dynamic or functional MR imaging or in phase contrast techniques.

In the following, the invention is further exemplified by means of examples with reference to the drawings, in which

Fig. 1 is the image of a test object, recorded by a conventional method;

Fig. 2 is the image of the same test object as in fig. 1, recorded with the method according to the invention;

Fig. 3 is the graphical representation of the contributions to the SRF for the conventional method and the method according to the invention;

Fig. 4 is a schematic representation of the sampling for non-weighted Cartesian imaging (left) and for PLANED imaging (right). The rectangle a shows the part of the k-space where the Nyquist criterion is fulfilled;

Fig. 5 shows the k-space weighting functions (top) and the corresponding SRFs (bottom) for the sampling of the PLANED experiment (a) and the final filtering (b) compared to a Cartesian experiment (c). The calculated SRFs (bottom) indicate that both experiments have the same spatial resolution. For comparison the original cosine square-function (d) and the corresponding spatial response function (dashed line) are also displayed;

Fig. 6 is a comparison of images recorded with standard Cartesian sampling (a and b), standard density weighted imaging (c and d) and the PLANED technique (e and f). The right column shows the same images zoomed to a quarter of the original image size. Images a, c, e and b, d, f are windowed equally, respectively;

Fig. 7 is a comparison of the measured spatial response functions resulting from the Cartesian image shown in figure 6a and the PLANED image in figure 6e;

Fig. 8 shows the position of the line where the SRFs shown in fig 7 were determined and position of the ROIs for the determination of the SNR; and

Fig. 9 are *in vivo* images of the human brain recorded with standard Cartesian sampling (left), standard density weighted imaging (middle) and with the PLANED technique (right). The stripes visible in figure 6e can be attributed to undersampling artefacts, whereas in 6d Gibbs ringing is visible.

Example 1

Fig. 1 shows the image of a test object formed by a conventional spin echo method with 256 phase encoding steps in horizontal direction. At the edges, a line pattern can be observed, which is a result of the absence of the spatial frequencies. Such artefacts are also called truncation- or Gibbs-artefacts.

Fig. 2 shown the image of the same test object as in fig. 1, the acquisition of the image, however, was now performed according to the example described below. In comparison to fig. 1, the artefacts are now considerably less pronounced and the signal to noise-ratio has significantly improved.

The method according to the invention comprises the step of density weighted or acquisition weighted sampling. In contrast to an MR experiment with a uniformly weighted k-space acquisition, the acquisition weighted and the density weighted encoding uses a continuous

weighting function. This makes it possible to optimise the SRF by means of using certain weighting functions as for example the Hanning- or the Gauss-functions. In the described example, a Hanning filter was used (see Mareci TH, Brooker HR. Essential considerations for spectral localization using indirect gradient encoding of spatial information. J Magn Reson 1991; 92: 229-246). For an acquisition weighted experiment, this weighting function is given by (see Pohmann R, von Kienlin M. Accurate phosphorous metabolite images of the human heart by 3D acquisition-weighted CSI. Magn Reson Med 2001;45:817-826):

$$w(k) = \frac{\Delta x_{nom} \cdot \beta}{2} NA_{tot} \left(1 + \cos\left(2\pi \frac{k \cdot \Delta x_{nom}}{\alpha} \right) \right).$$

To compare an unweighted experiment with an experiment that uses a non-uniform weighting, the nominal resolution Δx_{nom}, which in analogy to the optical resolution is given at 64% of the width of the SRF (siehe Kienlin M, Pohmann R. Spatial resolution in spectroscopic imaging. In: Blümler P, Blümich B, Botto R, Fukushima E editors. Spatially resolved magnetic resonannce, Weinheim: Wiley-VCH; 1998), and the total number of acquisitions must be the same in both cases.

In the mentioned example, the phase encoding steps were recorded density weighted in horizontal direction, i.e. the data acquisition yielded a data set in a non-Cartesian k-space. In the acquisition, the minimum density was limited to one third of the Nyquist density.

For reconstruction, the data was transferred into the x-k_y-space by means of a Fourier transform in reading direction. The k-space recorded in this example and the x-k_y-space can be divided in regions where the Nyquist criterion is met or exceeded and regions where the Nyquist criterion is violated. In order to be able to apply conventional FFT (fast fourier transform) algorithms for the reconstruction in the y-direction, the samples which are obtained by density weighted imaging are transferred o a Cartesian grid. To this end, a gridding algorithm as desribed in "Jackson JI, Meyer CH, Nishimura DG. Selection of a Convolution Function for Fourier Inversion Using Gridding. IEEE Transactions On Medical

Imaging 1991; 10:473:478" has been applied in the region of the k-space were the Nyquist criterion of met or exceeded.

Besides the step of acquisition weighted or density weighted imaging, the method according to the invention also comprises the step of a parallel imaging method. In the regions of the k-space where the Nyquist criterion is violated in the density weighted data acquisition, in the present embodiment the Cartesian lines required to achieve the Nyquist rate are reconstructed with the aid of the GRAPPA technique. In the embodiment, 4 neighbour lines were used to reconstruct the missing lines : "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". Thus, in the entire area a Cartesian k-space exists, in which the Nyquist criterion is met. In this embodiment, for image formation the k-space completely filled by the above method is multiplied with a filter according to the SRF so that the k-space denisty distribution (Fourier transform of the SRF) is essentially rebuilt. Instead of reconstructing over a Cartesian k-space with density correction it is also possible to reconstruct over a dense, non-Cartesian and not density-corrected k-space or over a dense, Cartesian and not density-corrected k-space.

In this example, in the oversampled regions of the k-space the reconstruction of a k-space is performed by means of known methods (regridding) with denity correction. Methods for density correction are discussed in "Pipe GP, Menon P. Sampling Density Compensation in MRI: Rationale and an Iterative Numerical Solution. Magn Reson Med 1999; 41:179:186". In this oversampled region the GRAPPA factors are determined.

Instead of GRAPPA factors, equivalent information such as coil profiles can be used. Instead of the oversampled regions, such information can also be obtained from prescans or additionally recorded k-space lines.

It should be noted that in principle methods where the reconstruction is performedin the k-space by means of reconstructing additional lines in the k-space as well as methods where the reconstruction is performed in the real-space into an expanded FOV, as for example in SENSE, can be used as parallel MR imaging methods, as well as so-called hybrid methods,

which can be performed in the real-space as well as in the k-space. Exemplarily, the following articles are referred to:

Jakob PM, Griswold MA, Edelman RR, Sodickson DK. AUTO-SMASH: A self-calibrating technique for SMASH imaging. Simultaneous acquisition of spatial harmonics. Magma 1998; 7:42:54;

Heidemann R, Griswold M, Haase A, Jakob PM. VD-Auto-SMASH imaging. Magn Reson Med 2001; 45:1066-1074;

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;

Kyriakos WE, Panych LP, Kacher DF, et al. Sensitivity profiles from an array of coils for encoding and reconstruction in parallel (SPACE RIP) Magn Reson Med. 2000;44:301-308;

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

Pruessmann KP, Weiger M, Bornert P, Boesiger P. Advances in sensitivity encoding with arbitrary k-space trajectories. Magn Reson Med 2001; 46:638:651;

Griswold MA, Jakob PM, Nittka M, Goldfarb JW, Haase A. Partially parallel imaging with localized sensitivities (PILS). Magn Reson Med 2000; 44: 602:609;

H. Köstler, J. Sandstede, C. Lipke, W. Landschütz, M. Beer, D. Hahn. AUTO-SENSE Perfusion Imaging of the Whole Human Heart. J. Magn. Reson. Imaging 2003; 18: 702 : 708;

Wang J, Kluge T, Nittka M,Jellus V, Kuehn B, Kiefer B, Parallel acquisition techniques with modified SENSE reconstruction (mSENSE). Proceedings of the 1st Würzburg workshop on parallel imaging, Würzburg 2001; 92; and

Kellman P, Epstein FH, Mc Veigh ER. Adaptive sensitivity encoding incorporating temporal filtering (TSENSE), Magn Reson Med 2001; 45: 846:852.

It is further conceivable to use an inherently self-calibrating method for non-Cartesian parallel imaging (see Yeh EN, Stuber M, McKenzie CA, Botnar RM, Leiner T, Ohliger MA, Grant AK, Willig-Onwuachi JD, Sodickson DK. Inherently Self-Calibrating Non-Cartesian Parallel Imaing. Magn Reson Med 2005; 54:1-8).

Further it is conceivable to perform a reconstruction acooriding to the method described in the article "Pruessmann KP, Weiger M, Scheidegger MB, Boesiger P. SENSE: sensitivity encoding for fast MRI. Magn Reson Med 1999; 42:952:962." The iterative non-Cartesian sensitivity encoding alogrithm is, however, used without density correction for the reconstruction of spirally and radially sampled images according to the method of the invention.

Fig. 3 shows a graphical representation of the contributions to the SRF which were obtained from the edge spread function for the conventional method (graph 1) and the method according to the invention (graph 2). Both images show a virtually identical image resolution, which corresponts to the widt of the SRF graphs. Further, it can clearly be seen that the truncation artefacts are considerably less pronounced in the metod according to the invention as compared to the conventional method.

Example 2

PLANED imaging combines density weighted imaging with parallel imaging techniques. In contrast to conventional Cartesian imaging (see figure 4 left) in PLANED imaging, the k-space sampling is performed in a density weighted manner (see figure 4 right). In order to achieve the same resolution as for an image acquired with a Cartesian sampling, i.e. the SRF at 64% of maximum intensity has the same width, $k_{y,max}$ has to be chosen larger in PLANED imaging (see *supra* Greiser A and von Kienlin M, 2003). The basic idea of PLANED imaging

is to fill the gaps of the undersampled parts of k-space by parallel imaging techniques, such as GRAPPA (see *supra* Griswold MA et al., 2002) or SENSE (see *supra* Pruessmann KP et. al, 1999) to avoid undersampling artefacts while maintaining the distribution of the sampling density defined by the weighting function.

If an oversampled part of k-space exists, it can be used for inherent self-calibration for the parallel imaging reconstruction (see Yeh EN, McKenzie CA, Ohliger MA, Sodickson DK (2005) Parallel Magnetic Resonance Imaging with Adaptive Radius in k-space (PARS): Constrained Image Reconstruction using k-space Locality in Radiofrequency Coil Encoded Data. Magn Reson Med 53:1383-1392; and Yeh EN, Stuber M, McKenzie CA, Botnar RM, Leiner T, Ohliger MA, Grant AK, Willig-Onwuachi JD, Sodickson DK (2005) Inherently Self-Calibrating Non-Cartesian Parallel Imaging. Magn Reson Med 54:1-8) and no additional data have to be recorded.

Written informed consent was obtained before each study. All measurements were performed on a 1.5 T scanner (Magnetom Symphony, Siemens) using a 12 channel array coil (6 channel body coil anterior and 6 channel posterior) and 8 receiver channels. Both phantom and *in vivo* measurements were carried out in transversal slice orientation and the array coils were positioned above and below the object.

Images generated with the PLANED method were compared to conventional Cartesian sampled images with the same spatial resolution and identical parameters, including number of phase encoding steps except for different phase encoding gradients ky. Images were recorded with a standard spin echo sequence (TE = 15 ms, TR = 525 ms). The raw data matrix was set to 256 x 512 for the phantom measurement and to 256 x 256 for the *in vivo* measurement. The density distribution of the phase encoding direction is presented in fig 5. The sampling density followed a cosine square function (Hanning function, $H(k)$) but was set to 1/3 in the outer part to avoid excessive noise enhancement in the parallel imaging reconstruction:

$$w(k) = H(k) \quad or \quad \frac{1}{3} \quad \text{whichever is larger .}$$

For 256 phase encoding steps $w(k)$ has to be set to 1/3 for k_y for the range indicated in figure 5.

PLANED and Cartesian imaging achieve the same spatial resolution, if the widths at 64% of the maximum intensity of their SRFs are equal (see Pohmann R, von Kienlin M (2001) Accurate phosphorous metabolite images of the human heart by 3D acquisition-weighted CSI. Magn Reson Med 45:817–826). Therefore the phase encoding steps of the PLANED experiments were adjusted as described by Pohmann and von Kienlin (see *supra* Polmann R and von Kienlin M, 2001), i.e. the value for w_{max} and $k_{y,max}$ were chosen larger for the PLANED sampling scheme. These parameters have been calculated to $k_{y,max}$=1.61 and w_{max}=1.23 for acquisition weighting in one dimension. This assures that the integral over w(k) yields exactly the same total number of accumulations NA and the same spatial resolution as for the corresponding unweighted Cartesian experiment. Figure 5 (bottom) displays the SRF for Cartesian, acquisition weighted and PLANED imaging, as it was used in the presented experiments and shows that the different used sampling schemes lead to the same resolution.

To calculate the signal-to-noise ratio two images with identical parameters were recorded for conventional Cartesian imaging and PLANED imaging, respectively. The signal-to-noise ratio within a region of interest was calculated by the ratio of the average and the standard deviation of the difference of both images.

The point spread function in phase encoding direction was determined as the derivative of the edge spread function. The latter was taken from a row perpendicular to the edge of the phantom as shown in figure 8.

Image reconstruction was carried out using custom-built software written in Interactive Data Language (Research Systems Inc., Boulder, CO, USA). To consider the coil profiles in x-direction the reconstruction process was carried out in the k_y-x-space, i.e. before the PLANED reconstruction the signal was Fourier transformed in the direction of the frequency encoding. The reconstruction was performed independently for each position in x-

direction. In the oversampled part data were gridded onto a dense Cartesian grid for every coil element using a Kaiser-Bessel-function as it is proposed in (see *supra* Pipe GP and Menon P, 1999). To reconstruct points for every coil element on a Cartesian grid in the undersampled part a PARS (see *supra* Yeh EN, McKenzie CA, Ohliger MA, Sodickson DK,2005) like GRAPPA (see *supra* Griswold MA et al., 2002) algorithm was applied. To calculate the signal for an aiming point on the Cartesian grid the four closest measured neighbour points were selected and the corresponding distances were determined. Using these distances the GRAPPA factors were determined in the oversampled part of k-space as described in detail below. This allowed the calculation of k-space points on a Cartesian grid fulfilling the Nyquist criterion also in the outer part.

The complete Cartesian k-space, i.e. the gridded inner part and the GRAPPA-reconstructed outer part, was multiplied by a cosine square filter to restore basically the sampled density distribution. The filtered k_y-spaces for every coil element were subsequently Fourier transformed and the final image was obtained by taking the square root of the sum of squares of the single coil images.

For every point on a Cartesian grid k_y in the outer part of the k-space GRAPPA reconstruction factors were determined independently. To calculate the signal for every aiming point on the Cartesian grid the $N_b = 4$ closest measured neighbour points ($k_{y,n1}$, $k_{y,n2}$, $k_{y,n3}$, $k_{y,n4}$) were selected and the corresponding distances were determined ($\Delta k_{y,n1} = k_{y,n1} - k_y$). In the oversampled part the measured data were sinc-interpolated in k_y direction. The measured points in the oversampled part of the k-space were used as auto calibration signals for four interpolated points at the calculated distances. The signal for the coil j at a point k_y is given by

$$S_j^{ACS}(k_y) = \sum_{l=1}^{L} \sum_{b=0}^{N_b-1} n(j,b,l) S_l\left(k_y + \Delta k_{y,nb}\right),$$

where l is the index of all coils and b denotes the individual neighbour point (0- N_b - 1). By setting up a system of equations for all measured point in the oversampled part of the k-space the different complex coil weights $n(j,b,l)$ were calculated by a singular value decomposition.

Using these coil weights the signals at Cartesian positions in the undersampled k-space were calculated.

To compare the PLANED method with standard density weighting and to investigate the influence of the aliasing artefacts due to the violation of the Nyquist criterion, the density weighted raw data were also reconstructed without parallel image reconstruction, i.e. using a regridding algorithm for the whole k-space. For this density weighted reconstruction the k-space was weighted by the same function as for the PLANED reconstruction.

Figure 6 shows the results of measurements of a test object recorded with the Cartesian, the PLANED and the density weighted technique. For the PLANED and the density weighted image the same raw data were used, but different reconstruction algorithms were applied. The Cartesian image has the lowest SNR and clearly reveals truncation artefacts in phase encoding direction, which can be observed as a line pattern at the edge of the phantom. The PLANED and the density weighted image show no Gibbs ringing in phase encoding direction. Both images exhibit the same SNR, but the density weighted image suffers from undersampling artefacts. These are visible as line patterns in the image at a distance at least 1/3 of the FOV from the signal producing structure. This corresponds to a maximum undersampling of a factor of 3. The best image quality is achieved by the PLANED imaging technique. The improved SRF results not only in the elimination of truncation and undersampling artefacts but also in the increase of SNR, in this example by 34 % compared to the Cartesian sampled image.

To test whether the spatial resolution Δy is identical for both images, the magnitude of the spatial response function was calculated from the edge spread function in the row indicated by a line in figure 8 for the Cartesian and the PLANED imaging technique. The magnitudes of the SRFs are presented in Figure 7. The results confirm experimentally that within the range of measurement errors both images have the same spatial resolution. Additionally the strong suppression of truncation artefacts for PLANED can be observed in the SRF, too. In contrast to the SRF obtained from the Cartesian image the SRF of the PLANED image shows much less pronounced side lobes.

An *in vivo* measurement of a human brain is presented in figure 9. The Cartesian image again reveals Gibbs ringing in the phase encoding direction. This is not observable in the density weighted image and in the PLANED image. While the density weighted image is corrupted by undersampling artefacts, these are not observable in the image, recorded with the PLANED technique.

The SNR measurements yield an enhancement of 17 % for the images obtained by the PLANED method.

The ROIs which were used for the calculation of the SNR, are indicated in figure 8.

In this work a new technique, parallel acquisition for effective density weighted imaging, is proposed and tested on phantoms and on *in vivo* applications. The proposed technique suppresses truncation artefacts and enhances considerably the signal-to-noise ratio of MR images of extended samples compared to conventional methods.

Acquisition and density weighted imaging methods have proven their capability to reduce the cross-talk between different pixels. At the same time these methods improve the SNR to an extent depending on the morphological geometry. The SNR gain originates from the suppression of the negative side lobes in the SRF compared to Cartesian imaging. Up to now the use of acquisition and density weighted imaging in clinical routine imaging was hampered by the increased mimimum scan time or, if the Nyquist criterion is violated, by aliasing artefacts.

One of the most important technical advances in the last decade was the introduction of parallel imaging techniques. Parallel imaging reconstruction allows to remove aliasing from undersampled data sets at the prize of a SNR amplification of a factor of g (see *supra* Griswold MA et al., 2002).

In this study the first results of a combination of density weighted and parallel imaging are presented. In the proposed implementation of PLANED imaging the sampling density was chosen according to a slightly modified cosine-square function. A cosine-square function is a

common k-space filter function that offers a favourable spatial response function (see *supra* Mareci TH and Brooker HR, 1991). The sampling density in our experiments was restricted to values larger 1/3 to keep the noise amplification factor (g-factor) by parallel image reconstruction low, even for our not optimized hardware. It could be shown, that the effect of small deviations from the cosine-square filter function on image SNR and SRF is minimal. The filling of the gaps in the sparse k-space was performed by a GRAPPA (see *supra* Griswold MA et al., 2002), or PARS (see *supra* Yeh EN, McKenzie CA, Ohliger MA, Sodickson DK, 2005) algorithm. But of course, PLANED imaging is not restricted to the use of the modified cosine-square filter or a GRAPPA reconstruction. Other filter functions like Kaiser-Bessel functions or numerically optimized functions (see *supra* Jackson JI et al., 1991) for a desired SRF may also be used. This also includes particular applications like edge enhancement filters for certain applications. Likewise the parallel imaging method included in our reconstruction can be substituted by any other parallel imaging technique and calibration method. In the implemented GRAPPA-reconstruction only 4 neighbouring points contributed to the reconstructed k-space points. According to the findings in (see *supra* Griswold MA et al., 2002) the usage of four neighbouring points results in the same reconstruction quality as a reconstruction with more points. But PLANED reconstruction is not limited to this particular choice. PLANED imaging may certainly profit from modified GRAPPA algorithms where regularization is performed after the singular-value decomposition for large coil arrays (see Lin FH, Kwong KK, Belliveau JW, Wald LL (2004) Parallel Imaging Reconstruction Using Automatic Regularization. Magn Reson Med 53:559-567; and Sodickson DK (2000) Tailored SMASH Image Reconstructions for Robust In Vivo Parallel MR Imaging. Magn Reson Med 44:243-251). But also other parallel techniques working either in the k-space [see *supra* Jakob PH et al, 1998; Heidemann R et al, 2001; Griswold MA, Blaimer M, Breuer F, Heidemann RM, Mueller M, Jakob PM (2005) Parallel Magnetic Resoance Imaging Using the GRAPPA Operator Formalism. Magn Reson Med 54:1553-1556; and Heidemann R, Griswold MA, Jacob PM (2003) Fast Parallel Image Reconstructions with non-Cartesian Trajectories Proc. Intl.Soc. Mag. Reson. Med 11: 2347], image space [see *supra* Pruessmann KP et. al, 2001, and Pruessmann KP et. al, 1999) or hybrid techniques see *supra* Kyriakos WE et al., 2000) can be used in the reconstruction of PLANED imaging. In this study coil sensitivity information was extracted from the densely sampled centre of the k-space. This self calibration can be applied to all cases were at least a

part of k-space fulfils the Nyquist criterion. But coil profile information may also be determined in a separate prescan (see *supra* Pruessmann KP et. al, 1999) or by completely filling a dense k-space in dynamic imaging (see Koestler H, Landschütz W, Sandstede J, Lipke C, Hahn D (2000) Abstract: Heart perfusion imaging using SENSE Radiology 217 (P) supplement, 464; Kellman P, Epstein FH, Mc Veigh ER (2001) Adaptive sensitivity encoding incorporating temporal filtering (TSENSE), Magn Reson Med 45: 846-852; Köstler H, Sandstede JJW, Lipke C, Landschütz W, Beer M Hahn D (2003) Auto-SENSE perfusion imaging of the whole human heart. J Magn Reson Imaging 18: 702-708; Köstler H, Beer M, Ritter C, Hahn D, Sandstede J (2004) Auto-sense view-sharing cine cardiac imaging. MAGMA Magn Reson Mater Phy 17:63-67; and Breuer FA, Kellmann P, Griswold PA, Jakob PM (2005) Dynamic autocalibrated parallel imaging using temporal GRAPPA (TGRAPPA). Magn Reson Med 53:981-985).

In the examples presented in this study the spatial resolution and the number of phase encoding steps were kept constant to compare Cartesian and PLANED imaging. But it is also possible to design a PLANED sequence that shows the same spatial resolution and the same SNR as the Cartesian technique. The presented results would correspond to a reduction of the acquisition time by a factor of about 1.4 to 1.8.

PLANED imaging is not restricted to a certain type of pulse sequence, as spin echo imaging. First results using gradient echo PLANED imaging (not shown in this work) indicate that the PLANED principle can be applied to all phase encoded pulse sequence.

The highest impact of PLANED imaging is expected in applications which are limited by the signal to noise ratio. Among these are fMRI studies, where the low SNR results in long acquisition times. But clinical images that SNR limited because they are recorded during breath hold or in real time should profit from the SNR gain in PLANED imaging. A special importance is expected for the application of the PLANED technique for 3D imaging. As in 3D imaging typically two dimensions are phase encoded the SNR gain in PLANED imaging should be squared compared to 2D MR imaging. PLANED imaging may substitute averaging and the need for higher field strength in some applications but it may also be combined with these techniques to allow new applications of MR imaging.

The new proposed technique, PLANED imaging, was applied to phantom and *in vivo* studies. The results showed that PLANED can improve the image quality by suppressing truncation artefacts and increasing considerably the signal to noise ratio. Thus PLANED opens the way for higher SNR and shorter scan times in most MR imaging applications.

It is an achievable advantage of the present invention to at least partially eliminate undersampling artefacts and to limit the scan time.

The entire content of the literature cited above in incorporated by reference into the present description.

Claims

1. A method of magnetic resonance imaging, comprising the step of sampling the k-space by means of the method of density weighted or acquisition weighted data acquisition, wherein in the sapling at least one region of the k-space is undersampled with respect to the Nyquist criterion,
characterised in that
the reconstruction of the MR image is performed at least in part with the aid of a parallel MR imaging method and the influence in the image of the missing k-space data required to fulfil the Nyquist criterion is reduced.

2. The method of claim 1,
characterised in that
missing k-space lines are reconstructed for example with the aid of a SMASH-, Auto-SMASH, VD-Auto-SMASH, GRAPPA, or a GRAPPA-operator-algorithm.

3. The method of claim 1,
characterised in that
in the parallel MR imaging method the reconstruction is performed in the real-space, for example with a SENSE, Auto-SENSE, mSENSE, T-SENSE or PILS-algorithm.

4. The method of claim 1,
characterised in that
in the parallel MR imaging method the reconstruction is performed in the real-space and the k-space, for example with the SPACE-RIP method.

5. The method of any one of claims 1 to 4,
charcterized in that
it comprises a spectroscopic imaging method, for example chemical shift imaging.

Fig. 1

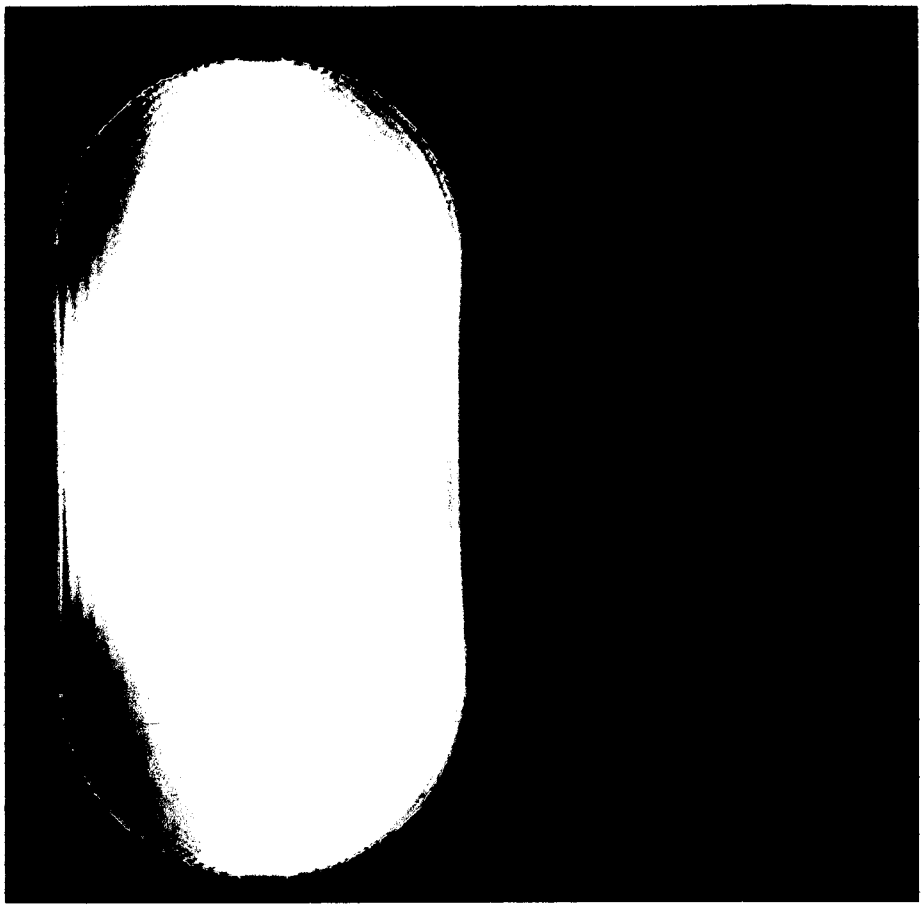


Fig. 2

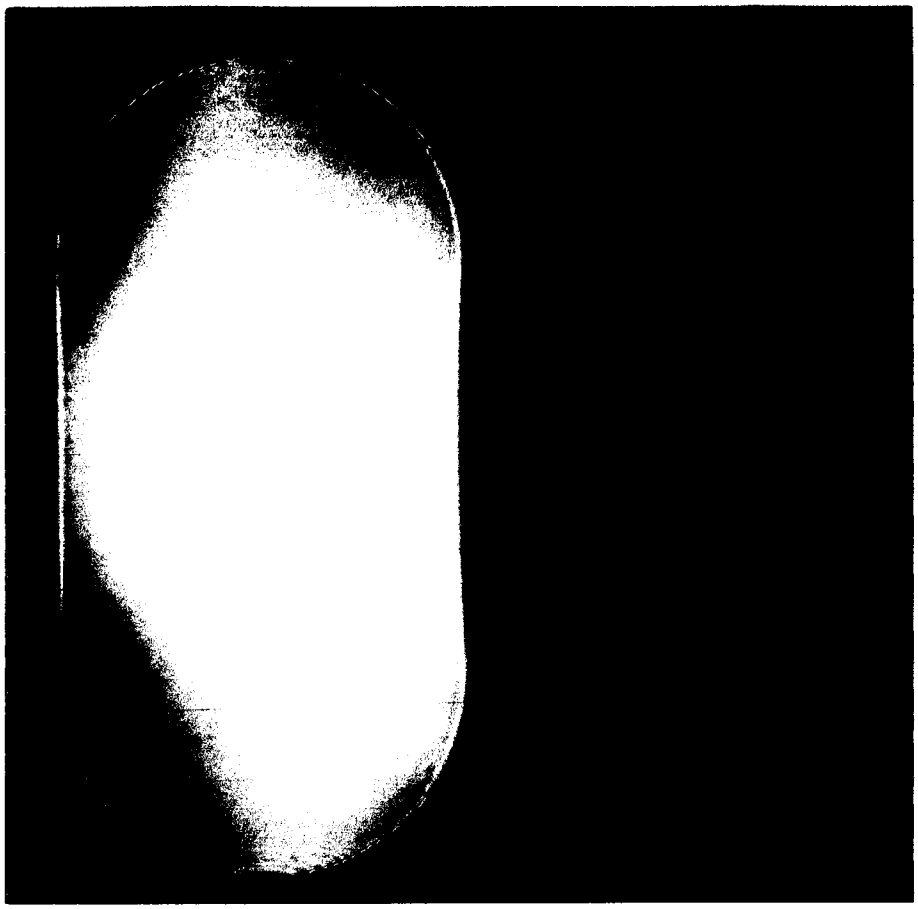


Fig. 3

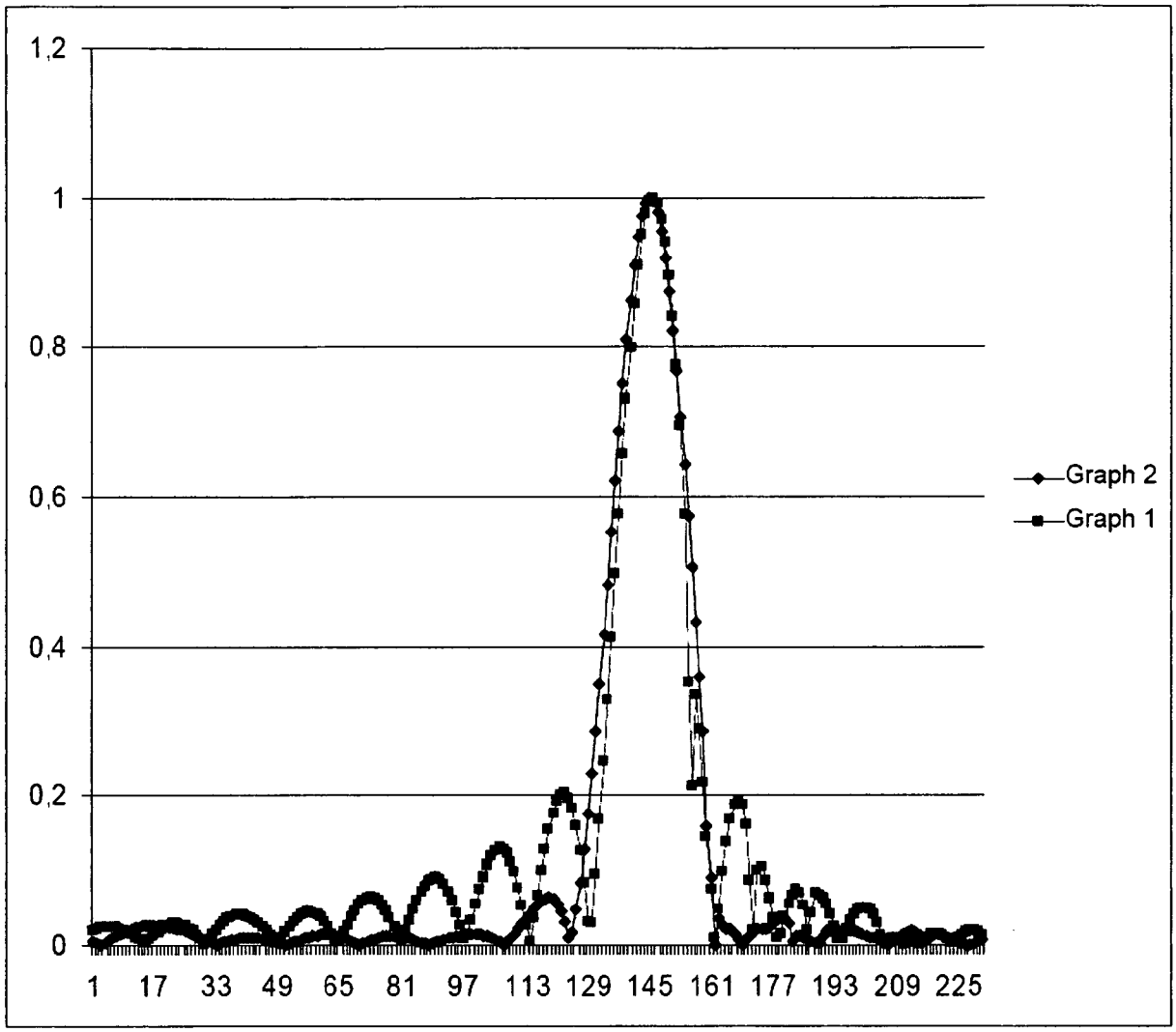

1,2
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0,6
0,4
0,2
0
1 17 33 49 65 81 97 113 129 145 161 177 193 209 225
Graph 2
Graph 1

Fig. 4

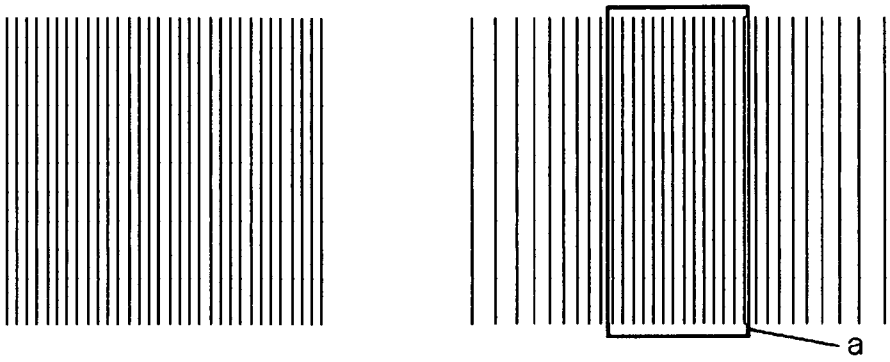

a

Fig. 7

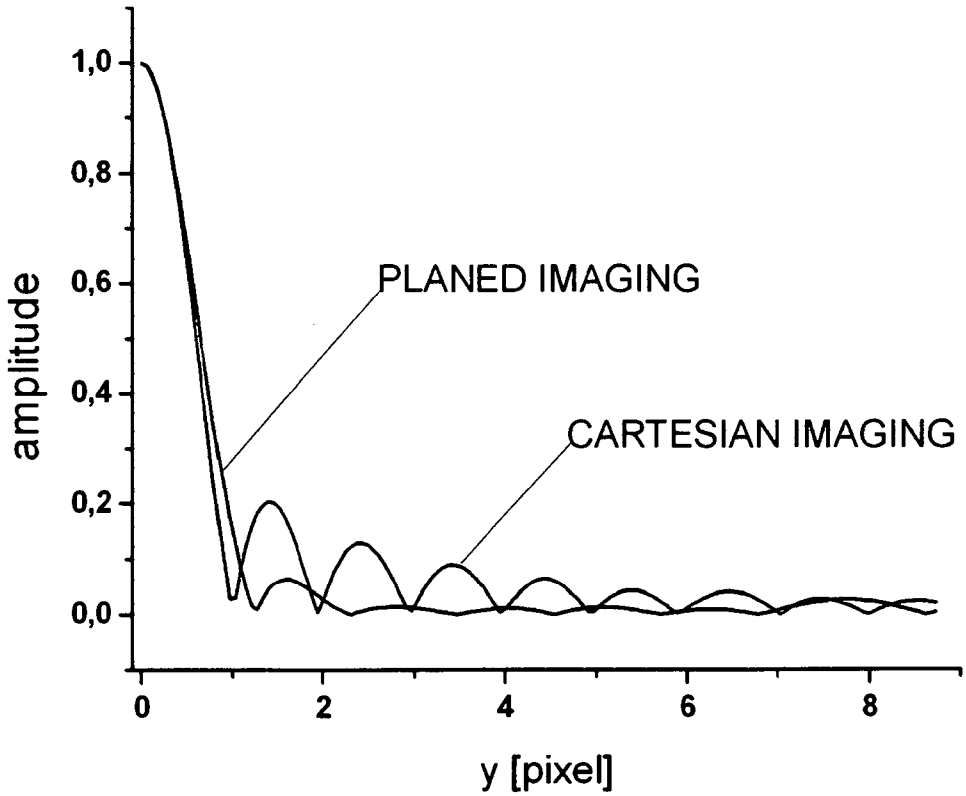

1,0
0,8
0,6
0,4
0,2
0,0
amplitude
PLANED IMAGING
CARTESIAN IMAGING
0
2
4
6
8
y [pixel]

Fig. 5

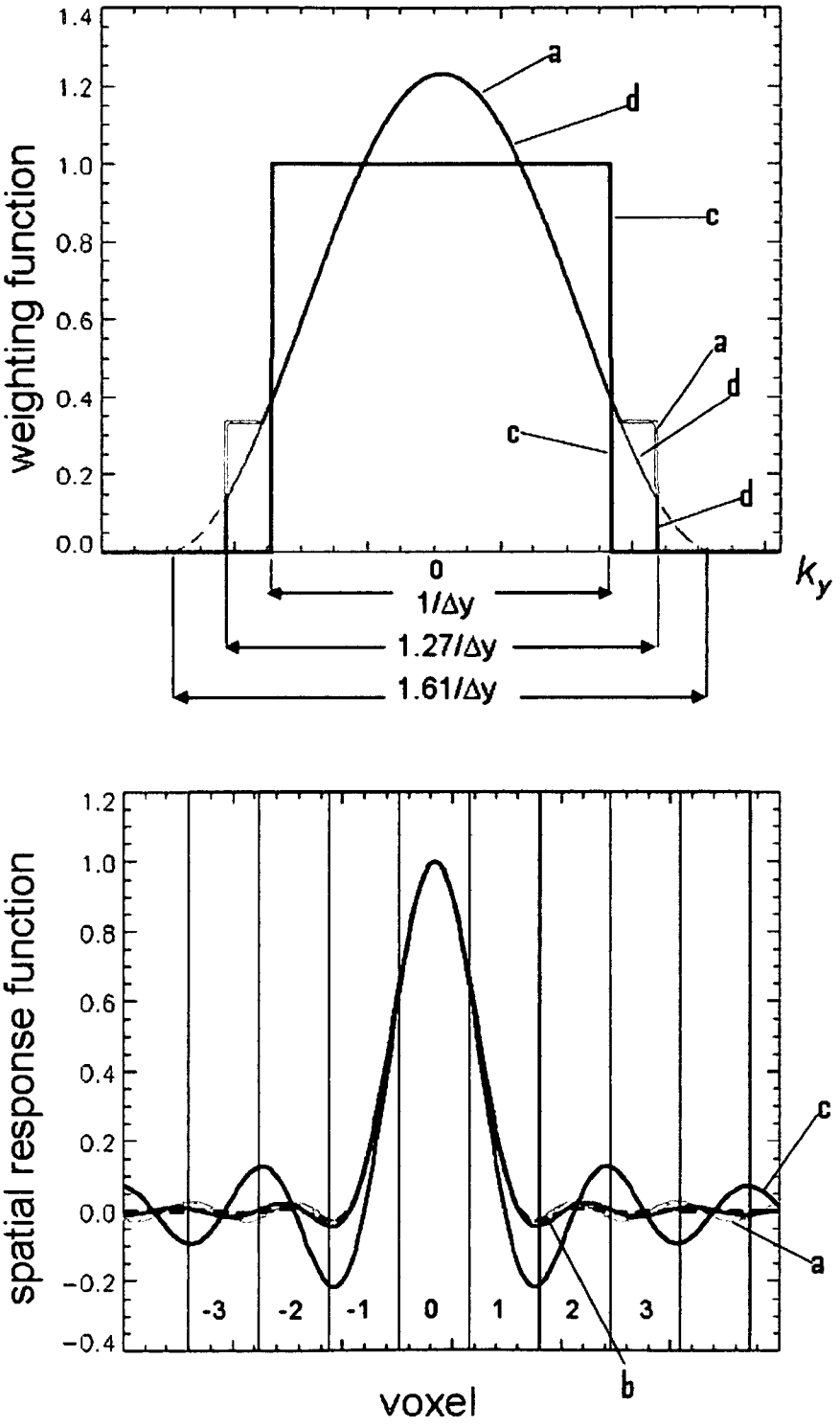

weighting function
1.4
1.2
1.0
0.8
0.6
0.4
0.2
0.0
a
d
c
a
d
c
d
0
1/Δy
1.27/Δy
1.61/Δy
k_y
spatial response function
1.2
1.0
0.8
0.6
0.4
0.2
0.0
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-2
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0
1
2
3
c
a
b
voxel

Fig. 6

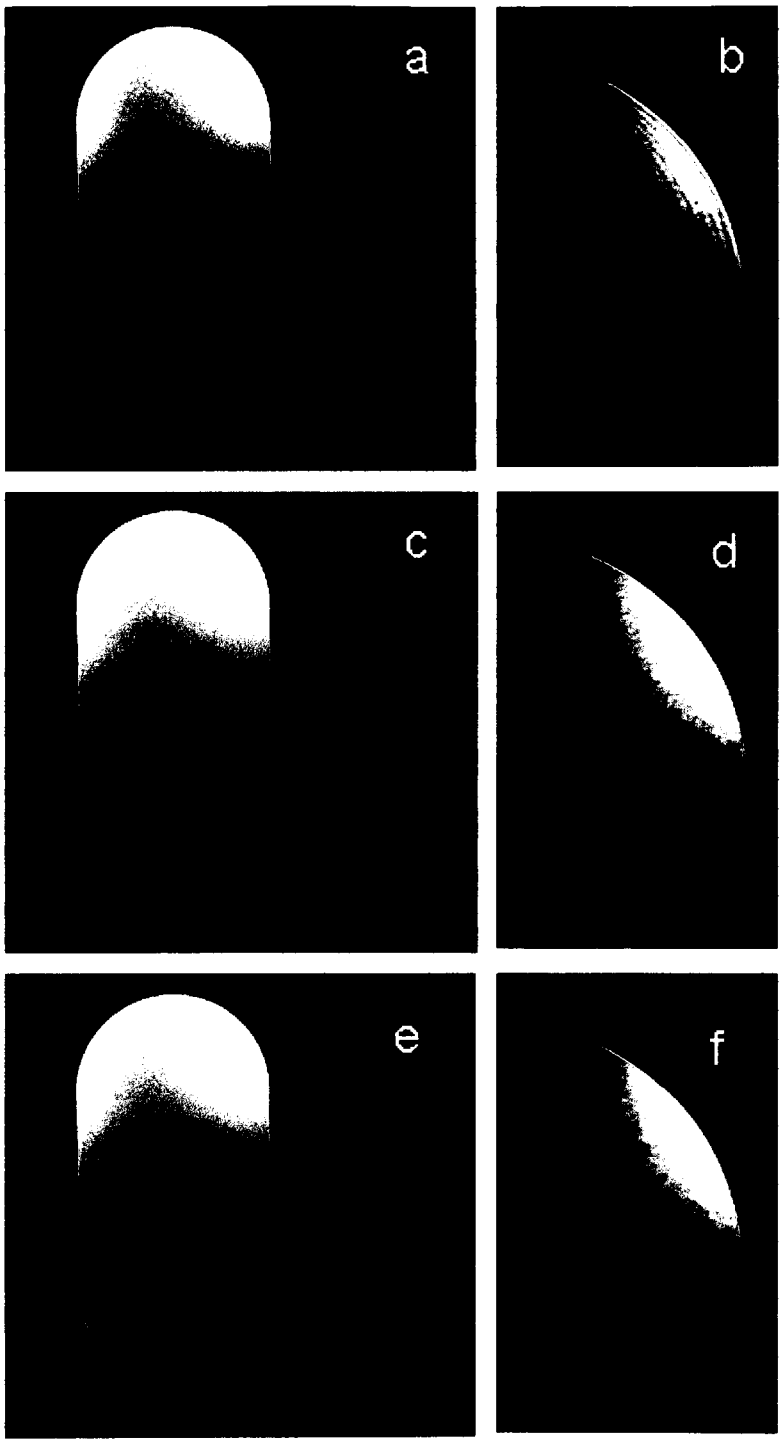


Fig. 8

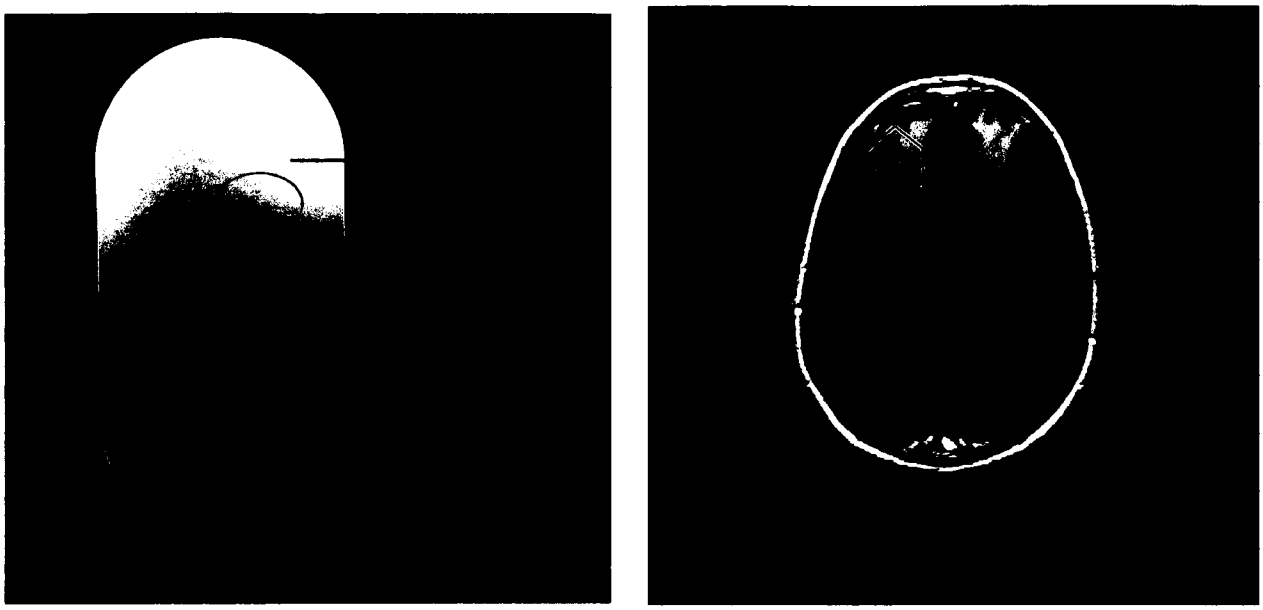
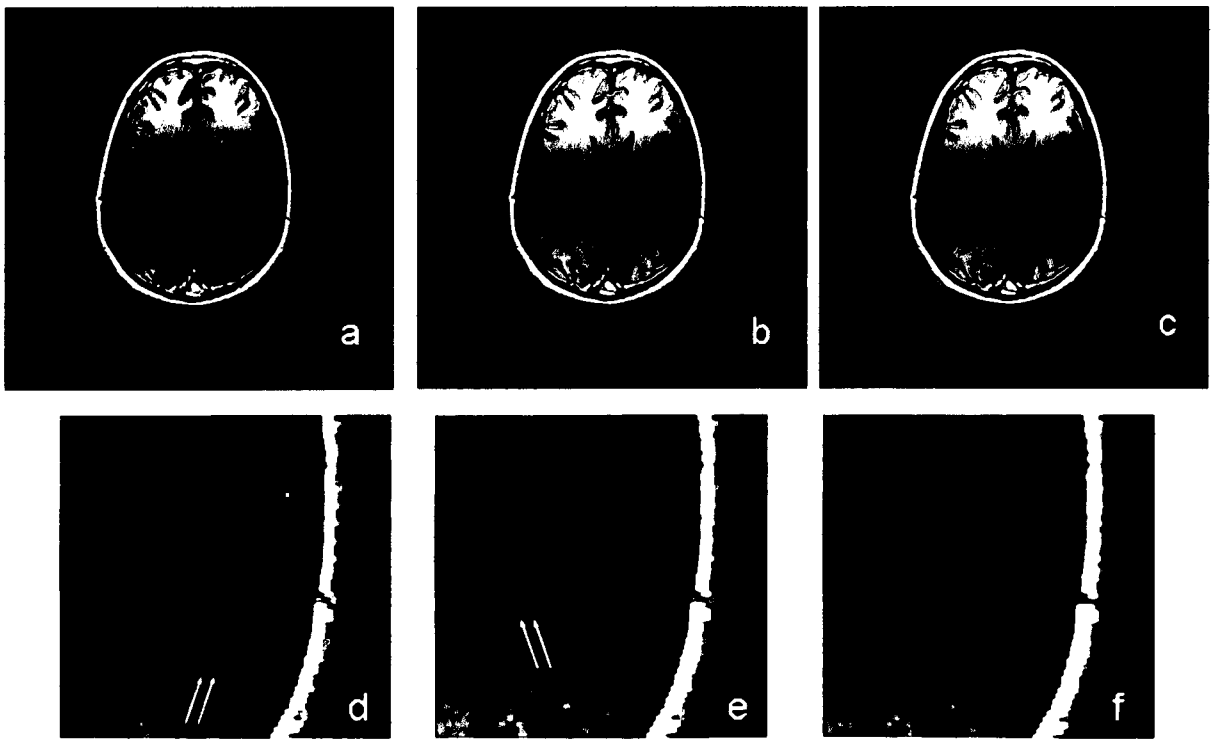


Fig. 9



INTERNATIONAL SEARCH REPORT

International application No
PCT/EP2007/000038

A. CLASSIFICATION OF SUBJECT MATTER
INV. G01R33/561

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
G01R

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)
EPO-Internal, WPI Data, COMPENDEX, INSPEC, BIOSIS, EMBASE, MEDLINE, IBM-TDB

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
T	GEIER OM ET AL: "Parallel acquisition for effective density weighted imaging: PLANED imaging" MAGNETIC RESONANCE MATERIALS IN PHYSICS, BIOLOGY AND MEDICINE, vol. 20, 20 January 2007 (2007-01-20), pages 19-25, XP002431531 the whole document	1-5
X	EP 1 236 436 A1 (HITACHI MEDICAL CORP [JP]) 4 September 2002 (2002-09-04) paragraph [0021] - paragraph [0056] figures 2-4	1-5

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[X] Further documents are listed in the continuation of Box C.

[X] See patent family annex.

* Special categories of cited documents :

"A" document defining the general state of the art which is not considered to be of particular relevance

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"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.

"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
27 April 2007	14/05/2007

Name and mailing address of the ISA/	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL – 2280 HV Rijswijk Tel. (+31-70) 340-2040, Tx. 31 651 epo nl, Fax: (+31-70) 340-3016	Streif, Jörg Ulrich

Form PCT/ISA/210 (second sheet) (April 2005)

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
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C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

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X	WO 03/093854 A (BRIGHAM & WOMENS HOSPITAL [US]; MADORE BRUNO [US]) 13 November 2003 (2003-11-13) the whole document	1,3,5
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X	ZHAO X ET AL: "The power spread function effect on low resolution sensitivity encoded imaging" PROCEEDINGS OF THE INTERNATIONAL SOCIETY FOR MAGNETIC RESONANCE IN MEDICINE, 9TH SCIENTIFIC MEETING AND EXHIBITION, GLASGOW, SCOTLAND, UK, 21-27 APRIL 2001, 21 April 2001 (2001-04-21), page 1799, XP002431533 the whole document	1,3,5
A	GREISER A, VON KIENLIN M: "Efficient k-space sampling by density-weighted phase-encoding" MAGNETIC RESONANCE IN MEDICINE, vol. 50, 2003, pages 1266-1275, XP002431534 cited in the application the whole document	1-5

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INTERNATIONAL SEARCH REPORT

International application No
PCT/EP2007/000038

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
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A	PIPE J G: "RECONSTRUCTING MR IMAGES FROM UNDERSAMPLED DATA: DATA-WEIGHTING CONSIDERATIONS" MAGNETIC RESONANCE IN MEDICINE, ACADEMIC PRESS, DULUTH, MN, US, vol. 43, no. 6, June 2000 (2000-06), pages 867-875, XP000932061 ISSN: 0740-3194 the whole document	1-5

INTERNATIONAL SEARCH REPORT

Information on patent family members

International application No
PCT/EP2007/000038

Patent document cited in search report	Publication date	Patent family member(s)	Publication date
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