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# A New Steady-State Imaging Sequence for Simultaneous Acquisition of Two MR Images with Clearly Different Contrasts

H. BRUDER, H. FISCHER, R. GRAUMANN, AND M. DEIMLING

*Siemens, Medical Engineering Group, Henkestrasse 127, 8520 Erlangen, West Germany*

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We present a new steady-state imaging sequence, which simultaneously allows in a single acquisition the formation of two MR images with clearly different contrasts. The contrast of the first image is FISP-like, whereas the second image is strongly $T_2$-weighted. In principle the $T_2$ values in the image can be calculated from the combination of the first and second images. We also show calculated $T_2$ images. © 1988 Academic Press, Inc.

Rapid magnetic resonance (MR) imaging with an imaging time <1 min is useful in many applications, for example, in suppression of respiratory motion artifacts if the acquisition time is reduced to about 10 s, in dynamic studies, or in 3D imaging. However, the possibilities for contrast variation are limited, as opposed to Spin-echo imaging. Up to now the standard steady-state imaging techniques generate in one acquisition only one MR image, the contrast of which is determined by the parameters of the sequence, namely the repetition time $\tau$, the echo time $T_E$, and the pulse angle $\alpha$. The new sequence presented here allows the formation of two MR images with different contrasts in one measurement.

Thus additional information for clinical diagnosis is available, especially for neuroradiological applications, without a significant prolongation of the imaging time.

## THEORY

If rf pulses with pulse angles $\alpha$ are repeated with a short repetition time ($\tau \ll T_1$) in the presence of a constant gradient between the pulses, then focusing points of the transverse magnetization will be established just after and before each rf pulse (Fig. 1). If the $X'$ axis of the rotating frame defines the direction of the rotation of the magnetization, caused by the rf pulses, then the transverse magnetization just after and before each pulse can be calculated from (*2*)

$$
\begin{aligned}
M_x^- &= M_0(1 - E_1)E_2 \sin\alpha \sin\theta/D \\
M_x^+ &= M_x^- \\
M_y^- &= M_0(1 - E_1)(E_2 \sin\alpha \cos\theta - E_2^2 \sin\alpha)/D \\
M_y^+ &= M_0(1 - E_1)[(1 - E_2 \cos\theta)\sin\alpha]/D \\
D &= (1 - E_1 \cos\alpha)(1 - E_2 \cos\theta) - (E_1 - \cos\alpha)(E_2 - \cos\theta)E_2 \\
E_1 &= \exp(-\tau/T_1), \qquad E_2 = \exp(-\tau/T_2),
\end{aligned}
\tag{1}
$$

where $\tau$ is the repetition time.



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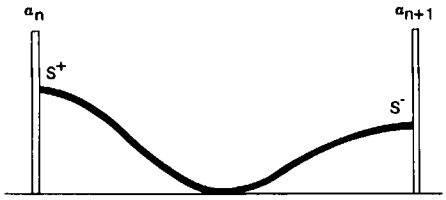


FIG. 1. Steady-state signal behavior between two rf pulses assuming a constant gradient between pulses.

The angle $\theta$ accounts for any phase rotation, which can arise from off-resonance effects due to inhomogeneities, susceptibility variations, or chemical shifts, or from any field gradients applied between the pulses. Because of these off-resonance effects the phase angle $\theta$ will take different values.

The signals $S_0^+$ and $S_0^-$ are both the sum of the magnetizations of isochromatic spin groups within a voxel,

$$S_0^+ = \int_{\theta_1}^{\theta_2} (M_y^+ + iM_x^+)d\theta \qquad [2]$$

$$S_0^- = \int_{\theta_1}^{\theta_2} (M_y^- + iM_x^-)d\theta. \qquad [3]$$

Equation [1] shows that $M_x^-$ and $M_x^+$ are point-symmetric functions of the phase angle $\theta$; thus the corresponding contributions to the integrals above vanish. The integration limits $\theta_1$ and $\theta_2$ depend on the value of the time-integral over the gradients within a repetition period, since the gradients are mainly responsible for the phase variations within a voxel. Note that the functions $M_y^-$ and $M_y^+$ have the periodicity $2\pi$.

If additional gradients with opposite sign are applied, the two signals $S_0^+$ and $S_0^-$, respectively, can be read out as echoes and used for imaging (Fig. 2).

The fast steady-state sequences FISP (*3*) and FLASH (*4*) make use of the signal $S_0^+$, while the sequence CE-FAST (*5*) utilizes the signal $S_0^-$ for imaging.

Generally steady-state sequences make use of gradient echoes. The time between the center of the rf pulse and the center of the echo leads to a reduction of the signal strength $S_0^+$ due to the irreversible $T_2$ decay and the influence of the main field inhomogeneities. It is a characteristic property of the steady state that inhomogeneities refocus within a repetition period (*2*). The $T_2$ decay is active during the whole repetition interval. If we take into consideration these effects, we must modify Eqs. [2] and [3],

$$S^+ = \int_{\theta_1}^{\theta_2} d\theta(M_y^+ + iM_x^+)\exp(-T^+/T_2)$$

$$S^- = \int_{\theta_1}^{\theta_2} d\theta(M_y^- + iM_x^-)\exp(+T^-/T_2),$$

where $T^+$ is the time between the $n$th rf pulse and the center of the $n$th echo $S^+$ and $T^-$ is the time between the center of the $n$th echo $S^-$ and the $(n+1)$st rf pulse.

In the equation above and in the following considerations we have assumed that the values for $T^+$ and $T^-$ can be chosen small enough that the influence of inhomo-

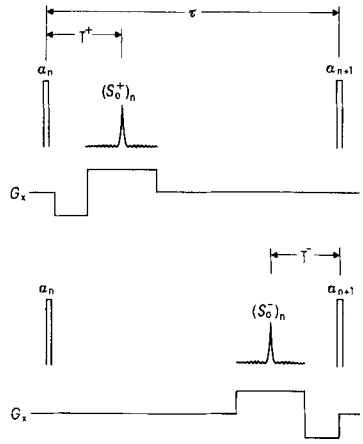


FIG. 2. The signals $S_0^+$ and $S_0^-$ respectively can be read out as echoes $(S^+)n$ and $(S^-)n$ by additional gradients as shown in the above figure.

geneities can be neglected. As we see later this assumption is justifiable, because the error in our $T_2$ calculations arising from this approximation is small (about 5%) compared with other contributions for possible errors. Another important point is the necessity of choosing the correct value for the repetition time $\tau$. Figure 3 shows that the signal $S_0^-$ decreases as $\tau$ increases, thus leading to a reduced signal-to-noise ratio. But as we discuss later, it is not useful to choose the lowest possible $\tau$ value, because shortening the repetition time will result in a similar contrast of the two images, which can be reconstructed from the signals $S_0^+$ and $S_0^-$.

It can be shown that for a pulse angle $\alpha = 90°$ the following relationship between the signals $S_0^+$ and $S_0^-$ is valid,

$$|S_0^-| = |S_0^+| \exp(-2\tau/T_2),$$

and for $S^+$ and $S^-$,

$$|S^-| = |S^+| \exp(-[2\tau - (T^+ + T^-)]/T_2). \qquad [4]$$

In this calculation we integrated the magnetization over a whole period $2\pi$; that means $\theta_2 = \theta_1 + 2\pi$.

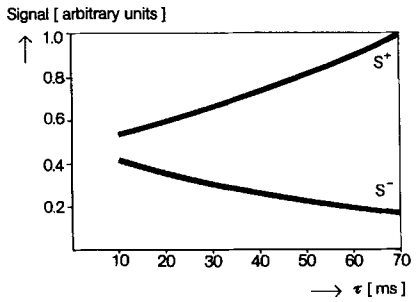


FIG. 3. The signals $S_0^+$ and $S_0^-$ as a function of the repetition time $\tau$ for $\alpha = 90°$, $T_1 = 600$ ms, $T_2 = 80$ ms.

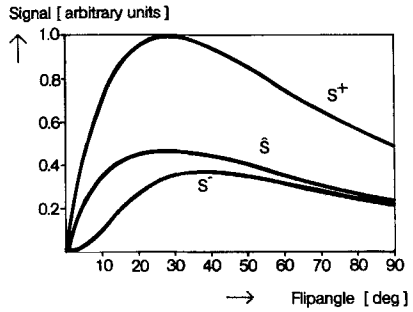


FIG. 4. The signals $S_0^+$, $\hat{S}_0 = S_0^+ \exp(-2\tau/T_2)$, and $S_0^-$ as a function of the pulse angle $\alpha$ ($\tau$ = 30 ms, $T_1$ = 600 ms, $T_2$ = 80 ms).

Equation [4] still remains approximately valid in a certain range below $\alpha = 90°$ (Fig. 4). The course of $\hat{S}$ is dependent on the relaxation times $T_1$ and $T_2$. The exponential relationship between $S^+$ and $S^-$ is the reason for a strong $T_2$ weighting of the image, which can be reconstructed from the signal $S^-$. In principle, different repetition times allow a variation of $T_2$ contrast. The $T_1$ contrast is nearly independent of the repetition time, because the signal $S^+$ is independent of $\tau$, as is shown below. If the repetition time is small enough, then the contrast of the image belonging to $S^-$ is similar to the contrast of the image resulting from $S^+$.

With the double-echo sequence shown in Fig. 5 it is possible to read out the signals $S^+$ and $S^-$ in a single acquisition. The acquisition time is slightly prolonged, compared with a standard steady-state sequence, because it takes time to read out the signal $S^-$. The sequence uses selective pulses as suggested in (*7*). It is important to switch the slice gradient $G_z$ symmetrically, since otherwise the signals will be reduced and the images will show artifacts. Moreover the symmetric switching of the slice gradient overcomes the problems of fast imaging sequences, connected with 90° pulses concerning the slice profile. Compared with the standard refocusing technique of the slice gradient, it can be shown that the symmetric switching yields a 25% higher signal from the slice of interest, and a uniform slice profile, as provided by a single 90° pulse.

The negative $G_x$ gradients just after and before the $n$th rf pulse, respectively, form the signals $S_0^+$ and $S_0^-$ to echoes. A similar work has been presented by Redpath (*8*).

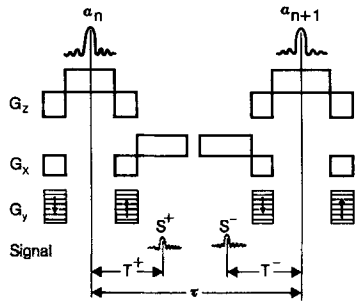


FIG. 5. Double-echo sequence for simultaneous readout of the differently weighted signals $S^+$ and $S^-$.

In principle the combination of the two MR images that can be obtained by this technique can be used for the determination of the transverse relaxation time $T_2$ pixel by pixel. According to Eq. [4] the ratio $S^+/S^-$ can be calculated from

$$|S^+/S^-| = \exp([2\pi - (T^+ + T^-)]/T_2).$$

According to the properties of the Fourier transformation, the equation above is also valid for the image intensities $I^+$ and $I^-$. This can be used for $T_2$ determination in the pixel at $(x_0, y_0)$,

$$T_2(x_0, y_0) = -(2\tau - (T^+ + T^-))/\ln(I^-(x_0/y_0)/I^+(x_0, y_0)),$$

The accuracy of this method is limited. Because of the rf-characteristics the pulse angle $\alpha$ varies over a wide range so that the deviations from the ideal pulse angle amount up to 30%. As we discussed above, Eq. [4] is valid only in a small range below $\alpha = 90°$. Furthermore we have neglected the influence of the main field inhomogeneities. The error caused by the inhomogeneities amounts to only about 5%. To estimate this value we have assumed that the inhomogeneity should not exceed 10 ppm at 1 T within a diameter of 20 cm.

The overall determining error arises from the reduced signal-to-noise ratio in the second image ($S^-$) and can be reduced only by averaging over more acquisitions. The total error depends on the number of averages and on the absolute $T_2$ value, but a 20% error should be realistic.

## EXPERIMENTAL RESULTS

Figure 6 shows an axial slice and Fig. 7 a sagittal slice of the head of a volunteer. The contrast of the first image ($S^+$) of each figure is FISP-like because the phase-

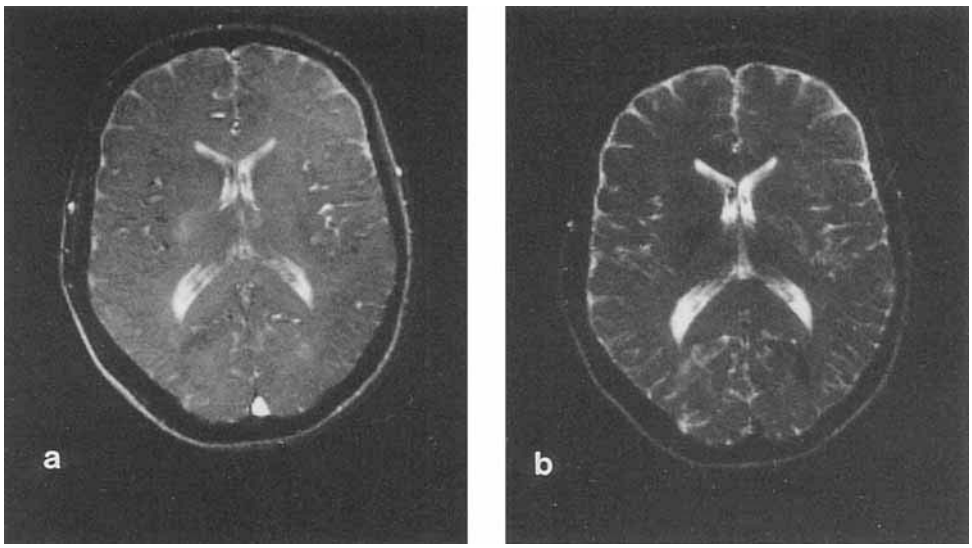


FIG. 6. Axial slice of the head of a volunteer: (a) first image (FISP), (b) second image (CE-FAST). The sequence parameters were $\alpha = 90°$, $\tau = 40$ ms, $T^+ = T^- = 10$ ms. Two acquisitions were made to enhance the signal-to-noise ratio.

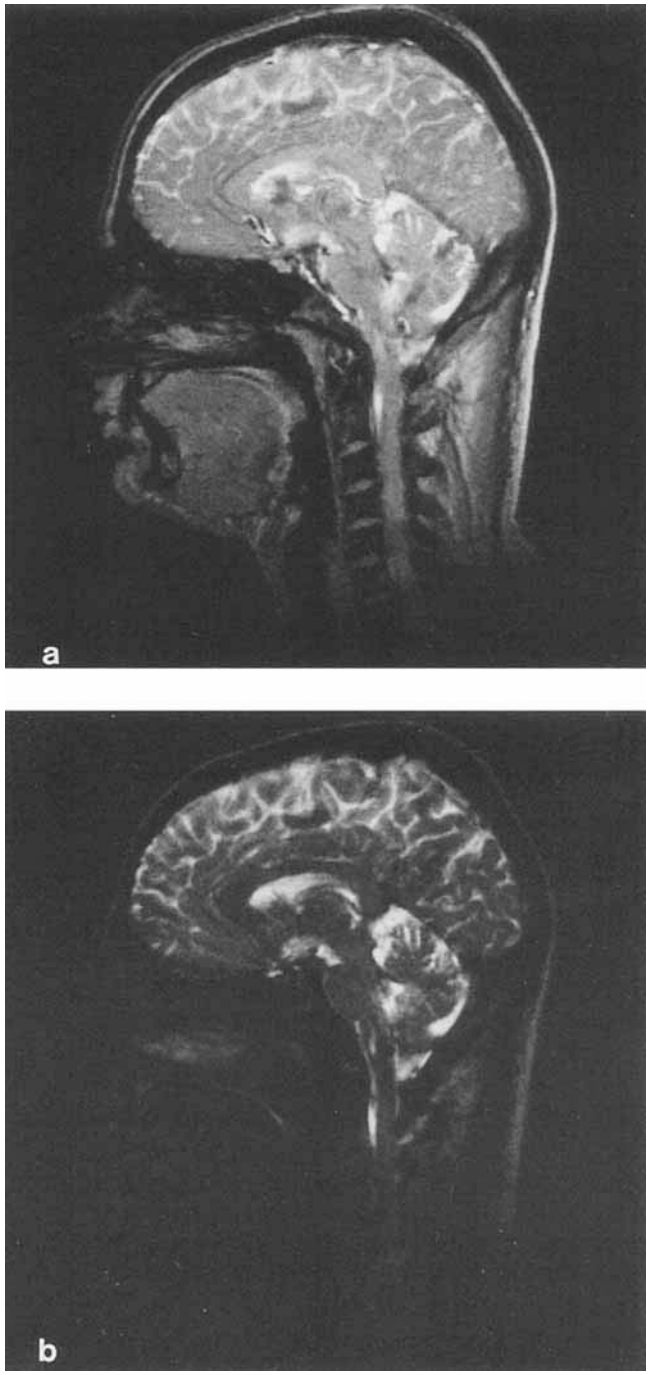


FIG. 7. Sagittal slice of the head of a volunteer: (a) first image (FISP), (b) second image (CE-FAST), showing strong $T_2$ weighting. The sequence parameters were $\alpha = 90°$, $\tau = 40$ ms, $T^+ = T^- = 10$ ms. Two acquisitions were made to enhance the signal-to-noise ratio.

encoding gradient is refocused in the double-echo sequence, presented here. It is mainly determined by the ratio $T_1/T_2$, because the FISP signal can be calculated approximately from (*6*)

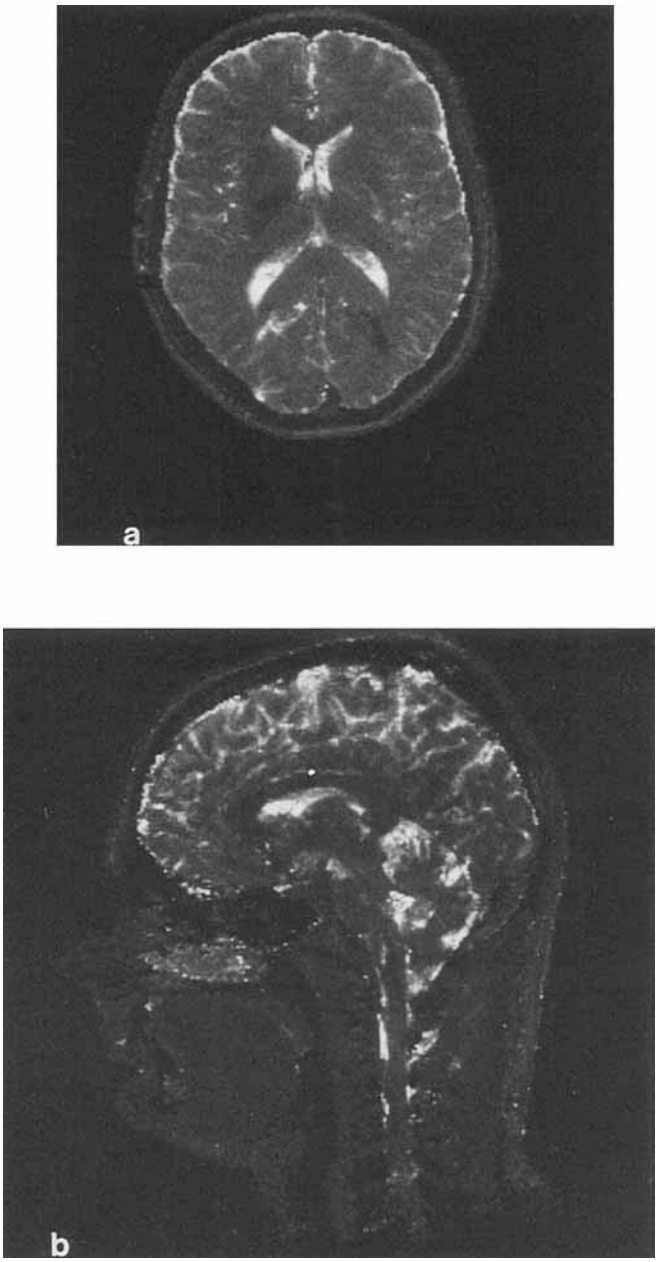


FIG. 8. Calculated $T_2$ images of (a) the axial slice in Fig. 6 and (b) the sagittal slice in Fig. 7.

$$S_{\text{FISP}} = \frac{\rho \sin \alpha}{(1 + T_1/T_2) + (1 - T_2/T_2)\cos \alpha}.$$

The second echo generates a strongly $T_2$-weighted MR image ($S^-$), the contrast of which is similar to the result of a CE-FAST experiment. As we see in Figs. 6 and 7, the second echo yields additional information due to the clear $T_2$ weighting. This additional information should improve the clinical diagnosis.

An interesting feature of the images resulting from the new sequence is the behavior of flow perpendicular to the slice. In the axial slice we see that the sinus appears very bright in the first image, whereas in the second image it has no intensity. The reason is that the corresponding blood flow is perpendicular to the axial slice so that not all flowing spins reach the steady state. A part of the flowing nuclear spin groups undergoes only a few rf pulses. As a consequence the signal from these spins is high compared with the signal of the stationary spins, which are in a steady state. In the second echo there is low flowing spin signals, because $S^-$ increases slowly with the number of rf pulses.

Following Eq. [1] we have calculated $T_2$ images from both the axial slice and the sagittal slice (Fig. 8). The $T_2$ image should be useful, especially if the relaxation times of different tissue have the same ratio $T_1/T_2$, but different $T_2$ values. Thus, it may be that they cannot be differentiated in the first image ($S^+$), but the $T_2$ image separates them clearly.

## CONCLUSIONS

We have developed a new steady-state imaging sequence, which allows the simultaneous acquisition of two MR images with clearly different contrasts. The first is FISP-like and, therefore, as is well known, mainly determined by the ratio $T_1/T_2$; the second is strongly $T_2$-weighted. Up to now imaging with standard steady-state sequences has been restricted to the acquisition of only one image in one measurement. The increase in the acquisition time compared to standard fast imaging sequences is negligible. In principle the combination of the two MR images makes it possible to determine the transverse relaxation $T_2$ pixel by pixel. The calculation of a $T_2$ image only requires knowledge of the time parameter of the double-echo sequence, presented here. The results show that this method should allow an improved clinical diagnosis, particularly in neuroradiological applications.

## REFERENCES

1. R. GRAUMANN, "Deutscher Röntgenkongress," Vol. 68, p. 853, 1987. [Book of Abstracts]
2. R. FREEMAN AND H. D. W. HILL, *J. Magn. Reson.* **4**, 366 (1971).
3. A. OPPELT, R. GRAUMANN, H. BARFUSS, H. FISCHER, W. HARTL, AND W. SCHAJOR, *Electromedica* **1**, 15 (1986).
4. A. HAASE, J. FRAHM, D. MATTHAEI, W. HÄNICKE, AND K. D. MERBOLDT, *J. Magn. Reson.* **67**, 217 (1986).
5. M. L. GYNGELL, N. D. PALMER, AND L. M. EASTWOOD, "Fifth Annual Meeting of the Society for Magnetic Resonance in Medicine, Montreal," p. 666, 1986. [Book of Abstracts]
6. W. S. HINSHAW, *J. Appl. Phys.* **47**, 8 (1976).
7. GARROWAY, *et al., J. Phys. C.* **7**, L 457 (1974).
8. T. W. REDPATH, R. A. JONES, AND J. R. MALLARD, "Sixth Annual Meeting of the Society for Magnetic Resonance in Medicine, New York," p. 228, 1987. [Book of Abstracts]