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● *Technical Note*

ADVANCES IN CARDIAC APPLICATIONS OF SUBSECOND FLASH MRI

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Flow-suppressed, subsecond FLASH MR images of the normal human heart have been obtained from single cardiac cycles using a 2.0-T whole-body MRI/MRS system (Siemens Magnetom) equipped with conventional 10 mT m^{-1} gradients. The present results demonstrate further technical improvements as compared to a previous report on the same subject (*Magn. Reson. Med.* 13:150–157; 1990). Measuring times of 139 msec and 209 msec were achieved by reducing the repetition time to TR = 4.36 msec (TE = 2.8 msec) and the spatial resolution to 32 × 128 or 48 × 128 measured data points, respectively. The flip angle was optimized to 12°. Spatial pre-saturation of 60 mm thick sections adjacent to the imaging plane resulted in a suppression of the blood signal and a clear delineation of the myocardium. Oblique rotation of the imaging slice provides convenient access to the anatomical long axis and short axis views of the heart. EKG-triggered images from separate heartbeats but at different cardiac phases demonstrate that the *effective* time resolution is considerably less than the actual imaging time.

Keywords: **Heart, MR studies; Heart, anatomy; Heart, ventricles; Myocardium, MR studies; Magnetic resonance, pulse sequence; Magnetic resonance, technology; Magnetic resonance, physics.**

INTRODUCTION

The clinical rationale for the acquisition of high-speed MR images with their unavoidable trade-offs in spatial resolution and signal-to-noise is given by the need of studying patients with cardiac arrythmia. In such cases the image data has to be obtained from a *single cardiac cycle*. This condition is neither met by spin-echo MR images nor by cine FLASH MRI studies which both make use of data from multiple, typically 128 or 256, cardiac cycles for the reconstruction of a single image.

Previous attempts at high-speed imaging are almost exclusively based on echo-planar imaging (EPI) techniques.[1-4] However, even though its basic principles have been known since 1976 and recent implementations take advantage of high field strengths,[5,6] EPI has not made a clinical breakthrough. While some of the possible reasons may be of a more general nature related to high-speed imaging *per se*, e.g. the reduced spatial resolution, other aspects refer to specific disadvantages of the EPI technique. For example, echo-planar images suffer from (i) limited signal-to-noise due to the use of only a single rf excitation pulse and the need for very large receiver bandwidths (very short acquisition periods) to overcome signal losses due to T_2 relaxation, tissue susceptibilities, and motion-induced phase errors, (ii) the hitherto inability to acquire oblique or even nontransverse[7] images, (iii) the extreme technical requirements for the gradient hardware, and (iv) the resulting costs for adequate equipment.

In a preceding pilot study[8] an alternative to EPI has been demonstrated to overcome many of the above problems. The new approach is simply a high-speed version of a conventional, basic FLASH MRI sequence [9,10] using a minimized repetition time TR and a reduced number of phase-encoding steps. Nontriggered heart images (250-mm field-of-view, 64 × 128 measured data points) were acquired with subsecond FLASH MRI in a small fraction of the cardiac cycle (300 msec).

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No motion artifacts were observed and contrast between the myocardium and blood was achieved by flow suppression using spatial presaturation.

In this contribution we describe further significant advances in terms of imaging speed (100–200 msec imaging times), flow suppression, oblique imaging, EKG triggering, the effective time resolution, as well as anatomical and functional heart studies of healthy volunteers using series of sequential images of different cardiac phases from consecutive heartbeats. All images in this study represent single excitation images without data averaging.

METHODS

Cardiac studies using subsecond FLASH MRI were carried out on a 2.0-T whole-body MRI system (Siemens Magnetom) with conventional gradients that give a maximum strength of 10 mT m^{-1}. The FLASH sequence used employed gradient switching times of 300 μs, repetition times TR of 4.36 msec, echo times TE of 2.8 msec, and acquisition periods of 1.92 msec sampling 128 complex data points with a receiver bandwidth of 66.7 kHz (assuming oversampling in the frequency-encoding dimension). Although the basic sequence is formally identical to that reported previously,[8] more sophisticated sequence programming resulted in a further reduction of TR on this commercial system. The resulting total imaging times were 139 msec and 209 msec for images with 32 to 48 phase-encoding gradient steps, respectively. In all cases the phase-encoding gradient was stepped from negative to positive gradient amplitudes with zero phase-encoding in the center of the image acquisition as in conventional Fourier imaging. Heart images were acquired with a field-of-view (FOV) of 300 mm and a slice thickness of 8 mm using the standard circular polarized body coil of the system. For display purposes the images were interpolated to 256 × 256 pixels.

The flip angle of the section-selective rf pulses (0.64 msec duration, Gaussian shape) was chosen to be 12° based on a theoretical analysis of the signal strength during the approach to steady state.[11] Data acquisition in subsecond FLASH MRI starts immediately after the first rf excitation from the full longitudinal equilibrium magnetization without the application of "dummy" repetition cycles. In general, recording of the image was completed before the magnetization reached a steady state. It should be noted that the achieved speed was not limited by the existing gradient hardware. It has been chosen to avoid geometrical distortions (10%–15%) of the high-speed images at even faster gradient switches or reduced acquisition periods. The use of a different order of phase-encoding gradients, e.g., starting with zero phase-encoding followed by alternatively ascending and descending gradients, resulted in image distortions probably due to a phase mismatch of the positive and negative part of the raw data.

Blood signal in the heart was presaturated by applying a 90° section-selective rf pulse on either side of the imaging plane.[12] The pair of pulses was applied twice with a 20-msec interval leading to an overall preparation period of 50 msec. The pulses were sinc-shaped with a duration of 2.56 msec. The presaturated regions were 60 mm thick and positioned at 35 mm from either side of the imaging slice leaving a 10-mm gap for the 8-mm slice. In general, spatial presaturation is considered to be preferable to inversion recovery techniques to eliminate blood signals because the required long inversion delays would severely limit the application of EKG triggering.[13,14] Inversion techniques also lead to a signal reduction of the myocardium.

Transverse and coronal images were acquired to determine the angles and the slice shift defining the anatomical axes of the heart followed by single and double oblique images to obtain long or short axis views. EKG synchronisation was used to position the images at chosen points within the cardiac cycle. Sequential images with different positions or cardiac phases were recorded from separate heartbeats. No averaging was applied throughout this study. All investigations were performed on young healthy volunteers with informed consent prior to the investigation.

RESULTS

An important aspect of high-speed imaging is the achievable trade-off between spatial resolution and measuring time. Figure 1 compares a set of images (a–d) acquired in 139 msec and a data matrix of 32 × 128 pixels with images (e–h) acquired in 209 msec but at a higher resolution of 48 × 128 pixels. All images were EKG triggered with a,b,e,f recorded at systole and c,d,g,h at diastole. The EKG delay times given throughout this paper represent the total delay between the R wave of the EKG signal and the center of the image acquisition, i.e., the time of zero phase-encoding that dominates the gross anatomical appearance of a Fourier image.[15,16] Even the small structural changes seen in images at 50 msec apart, e.g., Figs. 1 (a), (b), and 1 (e), (f), were fully reproducible. These systolic images depict the gradual contraction and thickening of the myocardium, while the diastolic images in Fig. 1 delineate the much thinner left ventricular wall and septum during the opening of the ventricle. Papillary muscles inside the ventricle are seen in both

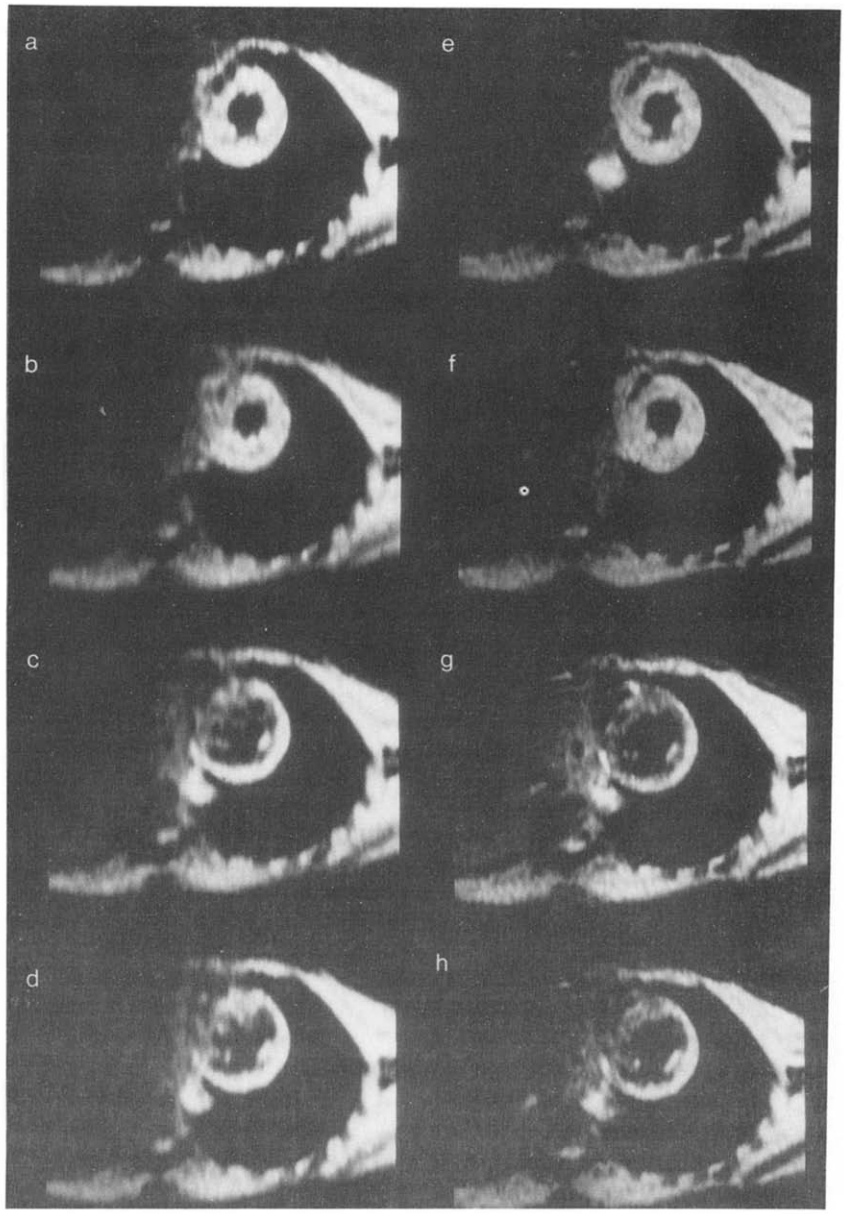

Fig. 1. 84 MHz (2.0-T) subsecond FLASH MR images of the heart of a young, healthy volunteer (field-of-view 300 mm, section thickness 8 mm, flip angle 12°). Images (a–d) were obtained within an acquisition time of 139 msec using 32 × 128 measured data points interpolated to a 256 × 256 pixel matrix for display purposes. Images (e–h) were recorded from the same volunteer and the same sections as in (a–d) but with an acquisition time of 209 msec (48 × 128 pixels). The two sets of single oblique (45° sagittal to transverse) images demonstrate the trade-off between spatial resolution and time resolution for the case of two systolic phases and two diastolic phases: images (a) and (e) refer to an EKG delay of 350 msec, (b) and (f) 400 msec delay, (c) and (g) 625 msec delay, and (d) and (h) 650 msec delay. In all cases the delay times represent the total delay between the R wave and the center of the image acquisition, i.e. the time of zero phase-encoding that dominates the gross anatomical appearance of a Fourier image.

sets of images obtained during diastole. In some images the blood signal in the aorta is not fully presaturated due to high-velocity flow.

The short axis views shown in Fig. 1 are obtained by oblique imaging rotating the orientation from a sagittal to a transverse plane by 45°. Figure 2 shows a related series of oblique images of the heart obtained with different rotation angles. Such series of images were used in order to locate the short axis view. The images in Figs. 2 (a)–(d) represent single oblique ac-

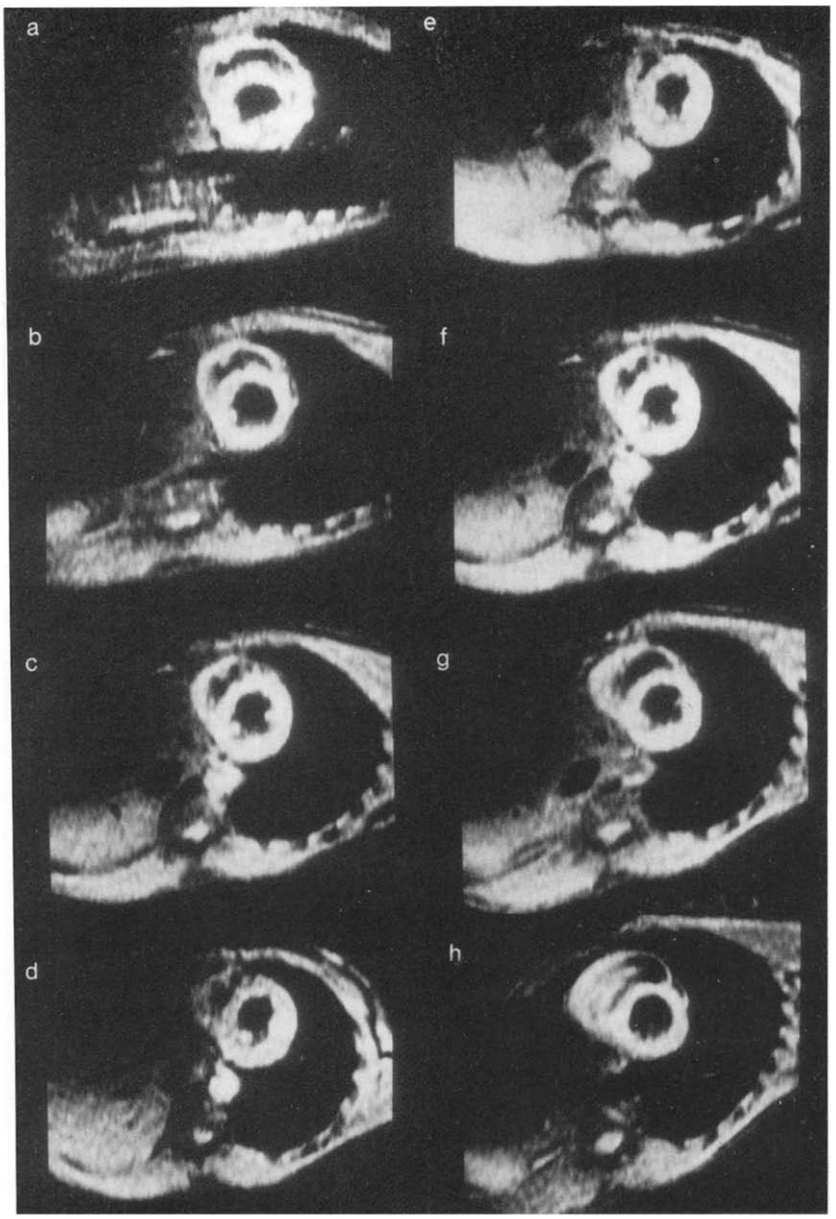

Fig. 2. 84 MHz (2.0-T) single and double oblique subsecond FLASH MR images of the heart of a young, healthy volunteer (imaging times 209 msec, data matrix 48 × 128, field-of-view 300 mm, section thickness 8 mm, flip angle 12°, EKG delay 300 msec). Images (a–d) were obtained in a single oblique mode with the section orientation rotated clockwise from sagittal to transverse by 15° (a), 30° (b), 45° (c), and 60° (d) to locate the short axis view. Images (e) and (f) were obtained in a double oblique mode rotating the image orientation from transverse to coronal by 10° (e), 0° (f) = (c), −10° (g), and −20° (h) in addition to a 45° rotation from sagittal to transverse.

quisitions with image orientation rotated clockwise from sagittal to transverse by 15° (a), 30° (b), 45° (c), and 60° (d). In a second step double oblique images were recorded selecting the 45° sagittal to transverse plane as a base image. Accordingly, the orientation of images shown in Fig. 2 (e)–(h) were further rotated from transverse to coronal by 10° (e), 0° (f) = (c), −10° (g), and −20° (h). Altogether, the sections show various systolic cuts at the same EKG delay of 300 msec through the left and right ventricle acquired in 209 msec (48 × 128 pixels) each.

Figure 3 shows a series of images acquired parallel to the short axis, but at 10 mm apart, demonstrating the normal anatomy of the heart from its apex to the

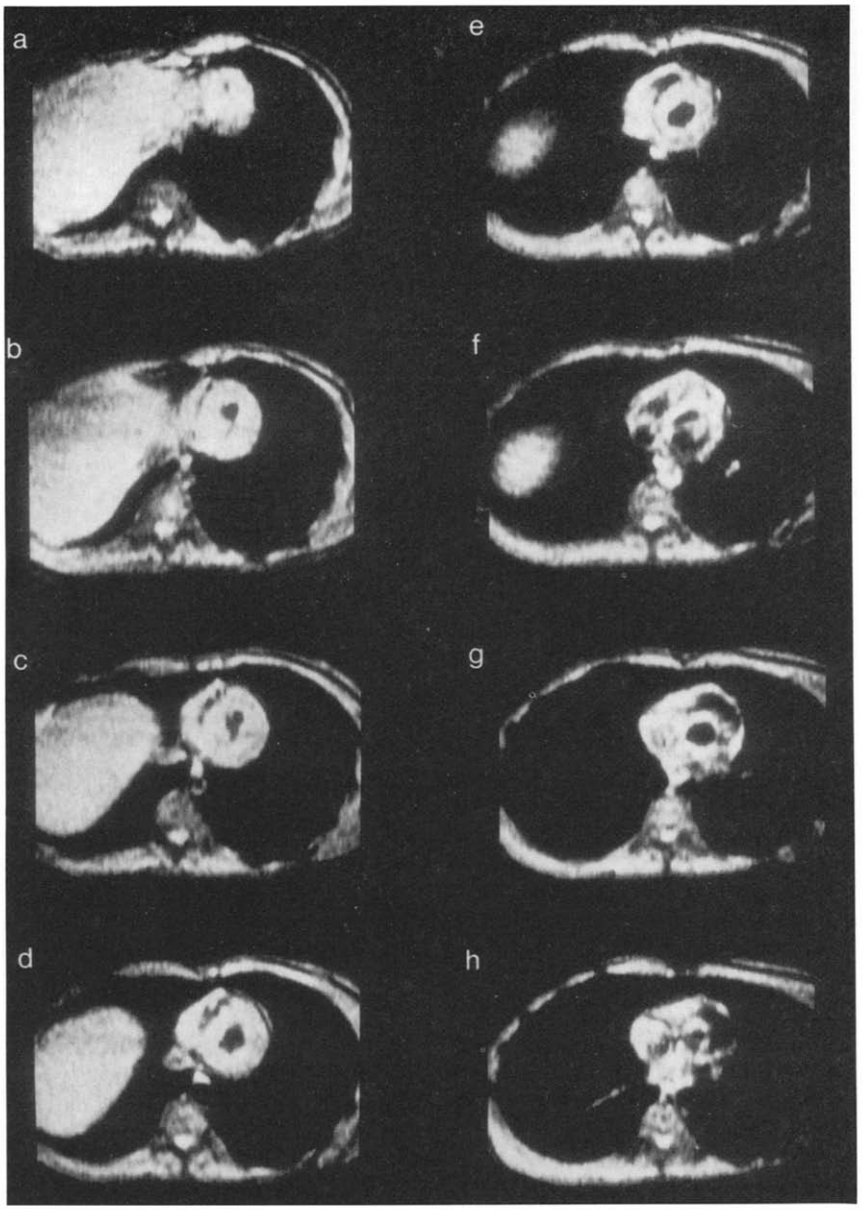

Fig. 3. 84 MHz (2.0-T) subsecond FLASH MR images of the heart of a young healthy volunteer (imaging times 209 msec, data matrix 48 × 128, field-of-view 300 mm, section thickness 8 mm, flip angle 12°, EKG delay 350 msec). Images (a)–(h) represent single oblique (60° sagittal to transverse) systolic views at different anatomical levels. The sections are 10 mm apart covering the heart from the apex (a) to the base (h) parallel to the short axis plane.

base in systole. All images were acquired at the same point in the cardiac cycle, i.e. using an EKG delay of 350 msec, but from different heartbeats. The cross-sections were taken with a single oblique rotation of 60° clockwise from sagittal to transverse.

Figure 4 shows a series of 12 time-sequential, short-axis images that cover the cardiac cycle from 150 msec to 700 msec (EKG delays) in steps of 50 msec. Again the images stem from separate cardiac cycles. They depict the gradual contraction and thickening of the ventricular walls during systole, followed by relaxation of the ventricles in diastole. In the early part of this phase, i.e., in Figs. 4 (h), (i) where the center of the image acquisition occurs 500–550 msec after the R wave, the ventricular wall exhibits a sudden and complete loss of signal. Signal recovery comes gradually in late diastole, starting at about 150 msec after the initial drop in intensity. The phenomenon was quantita-

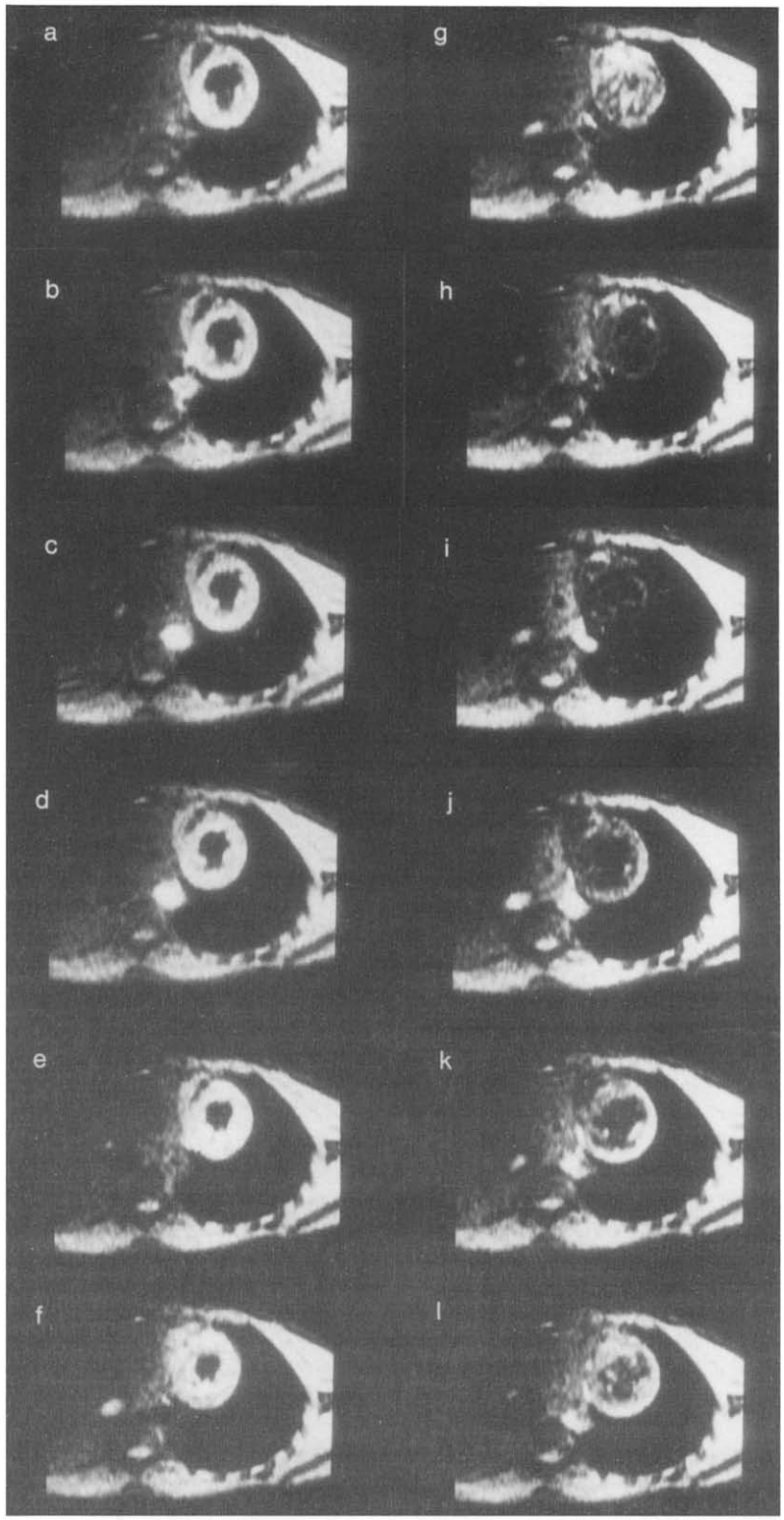

Fig. 4. 84 MHz (2.0-T) subsecond FLASH MR images of the heart of a young, healthy volunteer (imaging times 209 msec, data matrix 48 × 128, field-of-view 300 mm, section thickness 8 mm, flip angle 12°). Images (a–l) represent single oblique (45° sagittal to transverse) short axis views through the left ventricle from different heartbeats. The series of images covers the cardiac cycle in 12 distinct phases ranging from an EKG delay of 150 ms (a) to 700 msec (l) in 50-msec steps. (For details see text.)

tively reproducible with all volunteers. It is not yet clarified whether the signal loss is due to rapid motions during the relaxation/expansion of the ventricular wall immediately after systole, or caused by the rapid inflow of blood into the coronary vasculature. This blood carries paramagnetic, oxygenated hemoglobin that may create susceptibility gradients resulting in an extremely short effective T_2 relaxation time of the myocardium.

DISCUSSION

Motion artifacts along the phase-encoding direction as found in conventional Fourier MR images are not present in subsecond FLASH MR images regardless of the actual measuring time. This finding may be explained by both the extremely short echo time of 2.8 msec and the short acquisition period of 1.92 msec precluding the propagation of motion-induced phase errors.[16,17] When compared to previous results[8] the total imaging time of the subsecond FLASH sequence was further reduced by using a more sophisticated way of pulse programming as well as by reducing the number of phase-encoding steps. As suggested by the images shown in Fig. 1, 32–48 steps seem to provide an optimum for the present experimental conditions. This is confirmed by the fact that the spatial resolution of heart images acquired with data matrices of 64 × 128 points did not show any improvement. This is probably caused by enhanced image blurring[16,17] due to the prolonged imaging time of 279 msec.

An important result of this study is the clear demonstration that the effective time resolution of the subsecond FLASH MR images is better than the actual imaging time. This apparently paradoxical statement needs explanation. First, it should be emphasized that the *imaging time* must not be confused with the *time resolution*. The time resolution refers to the ability of two images to distinguish between two events that differ by a certain amount of time. In other words, the time resolution is given by the minimum time that has to evolve to create two different images of a moving structure. Secondly, as pointed out by Riederer et al.[15,16] and elegantly used for MR fluoroscopy experiments, the gross anatomical or structural features of a Fourier image are mainly determined by the low spatial frequencies (acquired using zero or close to zero phase-encoding gradients). In the present study the repetition cycles responsible for the low spatial frequencies cover a range of perhaps 50 msec (≈12 TR), which may be considered as the relevant time resolution.

Assuming 50 msec as the effective time resolution of the present subsecond FLASH MRI sequences, two images from different heartbeats that are 50 msec apart in the cardiac cycle should reflect myocardial movements. This is clearly proven by the pairs of systolic (50 msec difference) and diastolic (25 msec difference) images in Fig. 1 as well as by the full set of time-resolved images shown in Fig. 4 (50 msec differences). Moreover, consistent results are obtained for identical cardiac phases in healthy volunteers. Obviously, data acquisitions during the imaging process are *not uniformly weighted*, or in other words, the low spatial frequencies are considerably emphasized in a Fourier imaging experiment. This behavior is completely different to the apparently analog situation in spatial dimensions. For example, shifting a 3-mm thick slice by 1.5 mm may yield two images that look different, but does not create an "effective spatial resolution" of 1.5 mm since all signal contributions are *equally weighted* along the spatial coordinate.

In general, one may adopt the view that subsecond FLASH MRI yields static and well-resolved images as long as the motions are of the order of the total imaging time. If the motions become faster, the images remain "sharp" but at a somewhat reduced resolution due to an implicit sacrifice of high spatial frequencies. Only if the motions become so fast that major structures move significantly during the acquisition of the low spatial frequencies, the resulting images become degraded. Interestingly, the argument may be reversed in the sense that it is useless to acquire a large number of high spatial frequencies in the presence of rapid movements. This is the reason why images with 32 or 48 phase-encoding steps and measuring times of 100–200 msec seem to provide an optimum. The effect is experimentally confirmed by the images in Fig. 1, exhibiting little loss in information and only minor image degradation as the number of phase-encoding steps is decreased from 48 to 32. On closer examination of the images, the boundary of the myocardium is more clearly defined in the images obtained with higher spatial resolution, as expected for cardiac phases that do not show extreme movements of the ventricular wall.

CONCLUSION

Cardiac subsecond FLASH MR images with acquisition times of 100–200 msec have been obtained using a 2.0-T whole body MRI system, with data matrices of 32 × 128 or 48 × 128 pixels. These images, each acquired from a single cardiac cycle, show no motion artifacts. EKG triggering to different cardiac phases demonstrates an effective time resolution of about 50 msec, which is much less than the actual imaging time.

When compared to high-speed alternatives based on EPI, the FLASH approach offers considerable ad-

vantages such as oblique imaging and inexpensive availability on state-of-the-art MRI systems. It maintains flexibility in the choice of optimal imaging time and spatial resolution. Artifacts due to chemical shifts and susceptibilities are avoided by means of strong gradients and short echo times, respectively. While further technical improvements are possible, the next step should concentrate on the critical evaluation of functional cardiac parameters, e.g. myocardial wall thickening, ventricular volumes, and ejection fractions.

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