Improvements in the single-shot Burst imaging method

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Synopsis
The ultra-rapid imaging technique Burst was introduced in 1988. Although it had a number of highly attractive features for single-shot acquisitions, it has been seen as a low signal-to-noise ratio (SNR) method and few significant results have been presented since 1997. Here, we show images obtained using a technically much improved version of the sequence with higher spatial resolution and better SNR. We discuss briefly the causes of remaining image artifacts.

Introduction
Burst [1] is an ultra-fast single-shot sequence that takes approximately twice as much time as Echo-Planar Imaging (EPI), but only about half that of a single shot Turbo Spin-Echo sequence (ssTSE). The most striking feature of Burst, when compared with EPI is the lack of artifacts associated with magnetic susceptibility variations in the volume of interest. This is because each echo in the imaging train can be made separately a spin-echo. For the same reason, there is no fat-related shift in the phase-encoding direction and so no pre-saturation of fat is required as with EPI when imaging the abdomen. Compared with ssTSE, the power deposition is greatly reduced, since the number of 180° pulses remains small, allowing rapid multi-slice imaging in areas of the body where this would be prohibited using ssTSE. In addition, a more flexible choice of T1 and T2 contrast is available with Burst than with any of EPI, ss-TSE or ultra-rapid gradient echo acquisitions.

However, the initial Burst imaging sequence had a number of limitations, primarily concerning poor SNR, lack of slice selection and low resolution. Although some of these were addressed by previous authors [2,3], the sequence never came into widespread clinical or even research use. The most recently presented results appeared in abstract form in 1997 [4], but, although they were encouraging, no full publication followed.

Methods
A number of different variants of the basic Burst concept have been implemented, in work aimed at improving the SNR and resolution of Burst in the head and investigating the novel application of Burst extra-cranially. The first important step is to replace the non-selective low flip-angle pulses used in [1, 2] with optimised slice-selective pulses. We pursued two approaches: Type (1) direct replacement of all 64 low-angle pulses in our basic sequence and Type (2) the use of a smaller number of excitation pulses followed by additional 180° pulses during the readout period to refocus a number of trains of spin echoes. The latter approach is similar to the Burst Spin Echo (BSE) described by Zha et al. [4], although with a somewhat different choice of parameters and acquiring a larger number of echoes for improved spatial resolution. An important consequence of the use of slice-selective excitation pulses is that the constant read gradient in [1,2] during the excitation period must be replaced with a series of read gradient lobes (one after each low-angle pulse), interspersed with slice gradient lobes at the time of the pulses. This process places high demands on the RF and gradient subsystems and we spent a considerable time investigating the different ways in which this can be done to minimise image artifacts, such as the “slice definition inconsistencies” alluded to by Zha et al. in [4]. The second key feature that we optimised was the refocusing train of 180° pulses and their associated spoiler gradients. The new sequences all use partial Fourier acquisition and images are reconstructed using the POCS algorithm.

Results and Discussion
All work was performed on a Siemens Vision MR scanner. Figure 1 is a comparison of the image quality obtained using (from left to right) a high-resolution TSE sequence (acquisition time ~30s), a 128×128 clinical EPI sequence (~150 ms), our original 64×64 Burst imaging sequence (~80 ms) and the new 128×128 Burst sequences (~200 ms). The sagittal slices on the top row all have FOV 300×300 mm². The Burst image has slice thickness 10 mm and is obtained with a sequence of Type (1). It is evident that the Burst images are much more distorted than the equivalent EPI. The images on the bottom row all have FOV 230×230 mm² and the slice thickness is 7 mm in all cases, so that we can obtain a direct comparison of SNR. The Burst image on the right is obtained with a sequence of Type (2). When normalised to take account of the different acquisition matrix sizes, the SNR’s of the sequences are in the ratios TSE : EPI : original Burst : improved Burst 124 : 52 : 4 : 41 for the region of brain of interest in fMRI. Type (1) Burst sequences are virtually free of artifacts, but have a lower SNR than Type (2), because the pulse flip angle is smaller. Type (2) sequences still have some residual blurring in the phase-encode direction. This is primarily due to inconsistencies in phase between the echo trains obtained from successive refocusing pulses. Our current solution to the problem is to acquire a non-phase-encoded reference scan (as is often done in EPI). This approach appears to work well in the head and adequately in the pelvis (see Figure 2). However, Burst imaging of the liver, which moves between the reference and main image acquisitions still remains a problem on our current scanner. We believe that this problem will be alleviated on a more modern scanner with improved gradient performance and more flexible pulse sequence programming.

Figure 1: Comparison of images obtained with the conventional and new sequences. Top row sagittal section, bottom row transverse section. In each case, from left to right: High-resolution TSE, EPI, original Burst sequence, new Burst sequences with higher resolution and improved SNR (this can be verified by enlarging the page on the electronic version of the abstract).

Figure 2: Single-shot pelvic image obtained using Burst. Note the absence of any susceptibility artifact or fat-shift artifact, as would be obtained with EPI. The low power deposition allows us to obtain multi-slice datasets faster than with HASTE.

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References