# 3-Directional Fast Acceleration Encoding

**F. Staehle[1], S. Bauer[1], B. A. Jung[1], J. Hennig[1], and M. Markl[1]**

[1]Department of Diagnostic Radiology, Medical Physics, University Hospital Freiburg, Freiburg, Germany

**Introduction:** For the assessment of myocardial motion or blood flow, direct measurements of regional acceleration might provide valuable information regarding flow changes or tissue function and structure. However, there is only one report of the implementation of a direct acceleration encoding phase contrast (PC) technique based on an acceleration encoded scan and an acceleration compensated scan [1]. In this study we report, for the first time, a gradient optimized acceleration encoding strategy along all three spatial dimensions. A pulse sequence implementation with a minimized echo time is presented and validated in phantom studies. Moreover, initial results of three-directional left ventricular acceleration encoding in a volunteer are shown. As a reference, the obtained results were compared to acceleration data derived from standard phase contrast MRI [2].

**Methods:**

**Sequence:** An acceleration encoded phase contrast sequence scheme was developed (Fig. 1). In the read and the slice direction, the read dephaser and the slice rewinder were included in the tripolar gradient waveforms used for acceleration encoding. Two-sided acceleration encoding, i.e. equal distribution of the desired 2nd gradient moment $M_2$, was used to minimize echo time [3]. For each encoding direction (slice, read, phase), 2 tripolar waveforms with $+/M_2/2$ (up-case) and $-/M_2/2$ (down-case) were designed. For gradient optimization, the maximum gradient amplitude $h_{max}$ (23 mT/m), the ramp time $r$ (370 µs), and the known shapes of the slice selection and readout gradients were used. For a given acceleration encoding moment $M_2$, the up-case was calculated by solving for the flat-top times $t_1$, $t_2$ and $t_3$ of the three gradients. The down-case was then calculated from r, $t_1$, $t_2$, $t_3$ and the encoding moment $M_2$ by solving for $h_1$, $h_2$ and $h_3$. The read direction was a time-reversed version of the slice direction. Echo time TE was further minimized by systematically evaluating the gradient waveform durations for different distributions of $M_2$ between the up and down waveforms. The phase encoding direction was implemented by using $h_{max}$, r and $t_2 = 2*t_1$ and the maximum phase encoding moment.

**Measurements:** All measurements were performed on a 3 T MR-system (Trio, Siemens, Germany). Two sequences were implemented with acceleration sensitivities of ±24 m/s² (phantom scan) and ±4.5 m/s² (volunteer scan) resulting in TEs of 8.7 ms and 13.9 ms and TR of 11.3 ms and 16.5 ms, respectively. Velocity data were measured with a two-sided velocity encoded PC sequence. Imaging parameters were $venc_{xy}$ = 15 cm/s, $venc_z$ = 25 cm/s, TE = 5.4 ms, TR = 8.3 ms. Common parameters for all measurements were: imaging matrix 256x256, FOV = 360mm x 360mm x 8mm, BW = 450 Hz/Px, flip angle = 15° / 10° for the phantom / volunteer measurements. For the in-vivo measurements, velocity data were acquired in an interleaved manner and acceleration data sequentially with two k-space lines per cardiac phase leading to similar temporal resolutions of 33.2 ms and 33 ms, respectively. Acceleration was derived from the velocity data by approximating the temporal first order derivative using adjacent time-frames [2]. The accuracy of acceleration encoding was validated using a rotating phantom with know diameter and rotation frequency and thus known regional acceleration. Maxwell and eddy current-corrections were applied to all data.

**Results:** The results of the validation of the acceleration sequence using a rotating phantom are shown in Figure 2. Regional distribution of measured local acceleration (black vectors) was in excellent agreement with theoretical values (red arrows, max acceleration = ±18 m/s²). Comparison of measured and derived left ventricular acceleration fields for maximum systolic tangential acceleration (figure 3, left) and maximum systolic radial acceleration (figure 3, right) in a midventricular short axis plain demonstrated close agreement and showed the feasibility of direct acceleration measurements. A quantitative comparison of results is shown in the top row of Figure 4 depicting the time courses of the average left ventricular acceleration along the three spatial directions. Note that direct acceleration encoding (continuous blue line) revealed higher peak values compared to data calculated form standard velocity encoded PC-MRI (dashed red line) .The errorbars represent the standard deviations of the acceleration values over the segmented left ventricle for the individual time frames. Correlation analysis (figure 4, bottom) of regional myocardial acceleration (24 angular segments) for all time frames demonstrated good agreement between both methods.

**Discussion:** The acceleration encoding implementation presented in this study resulted in considerably reduced echo times compared to previous approaches. Furthermore, compared to Fourier acceleration encoding, the two-sided implementation of acceleration encoding provided shorter total measurement times. The phantom validation of the sequence showed good agreement between theoretically and experimentally determined acceleration values. The accuracy was somewhat degraded in the phase encoding direction due to ghosting artefacts related to the non-optimal precession of the rotating phantom. For in-vivo measurements, the average time courses of the myocardial acceleration agreed well between the directly measured and derived acceleration data. Note that the standard deviation was increased for the directly measured time courses which may be due to the longer echo time and thus reduced SNR and stronger artefacts. However, acceleration peaks were more pronounced in the directly measured data indicating that the interpolation of 3 time frames for the derived data might act as a low-pass filter. For peak acceleration, the direct measurement of acceleration might thus be more adequate. Moreover, direct measurement of acceleration might be especially useful if applied to the higher acceleration of blood flow which can be measured using a higher acceleration sensitivity and thus reduced TE and TR. A further advantage compared to derived acceleration may be given by the convective term of the acceleration which is expected to play a bigger role in a fluid than in tissue.

**References:** [1] Forster J, et al. Med. Phys. 28;1:28-35 [2] Staehle F et al, Proceedings of the ISMRM 2008 [3] Bernstein et al., JMRI 1992;2:583-588

**Acknowledgements:** Grant support by the Deutsche Forschungsgemeinschaft (DFG), Grant # MA 2383/4-1 and the Bundesministerium für Bildung und Forschung (BMBF), Grant # 01EV0706

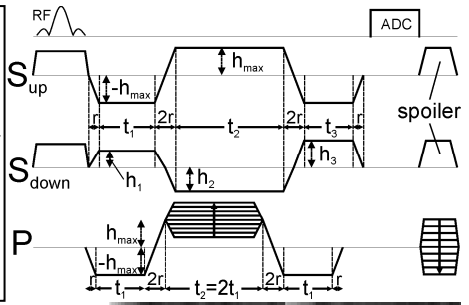


**Figure 1:** Pulse sequence diagram of the sequence for acceleration encoding. ($S_{up}$) and ($S_{down}$) denote the the gradient waveforms for the up-case and the down-case of the slice direction, respectively. The second gradient waveform in the phase encoding direction (P) corresponds to the negative waveform of the one shown.

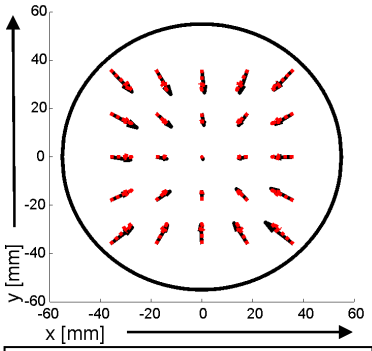


**Figure 2:** Rotating phantom used for validation with measured (black arrow) and theoretical (red arrow) acceleration values.

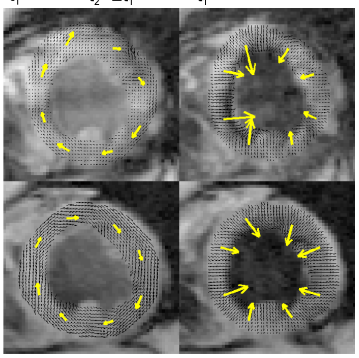

**Figure 3:** Measured (top) and calculated (bottom) vector plots of the in-plain myocardial acceleration.

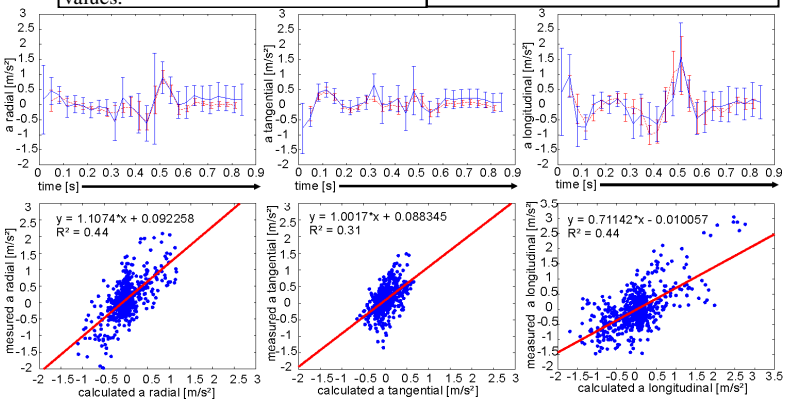


**Figure 4**: Average time courses (top) and correlation analysis (bottom) of measured (blue continuous line) and calculated (red dashed line) acceleration volunteer data. The red lines represent the linear regression fits.