Analysis and correction of gradient nonlinearity and B_0 inhomogeneity scaling errors in 2DPC flow measurements

J. M. Peeters[1], C. Bos[1], C. J. Bakker[1]
[1]Image Sciences Institute, University Medical Center, Utrecht, Netherlands

Introduction

In the past twenty years, numerous studies have been devoted to the various sources of error in phase contrast (PC) flow measurements, including velocity aliasing, intravoxel phase dispersion and partial voluming. However, none of these studies considered the effects of gradient nonlinearity (*G'*) and inhomogeneity of the main field (ΔB_0). Our interest in these effects was aroused by the somewhat disappointing results of a study in which flow measurements were performed in the loop graft in the forearm of hemodialysis patients (Figure 1a). On average, PC measurements in these grafts were 9% higher than ultrasound (US) dilution measurements [1], and, occasionally, significant differences were observed between the flow values of the arterial and venous limbs. We hypothesized that these observations could be attributed to the large off-center distance of the measurement location, where *G'* and ΔB_0 may be large. Therefore, we performed theoretical and experimental analyses of the influence of measurement position on PC flow measurements. Finally, we examined whether knowledge of the spatial dependence of *G'* and ΔB_0 would allow correction of biased flow values.

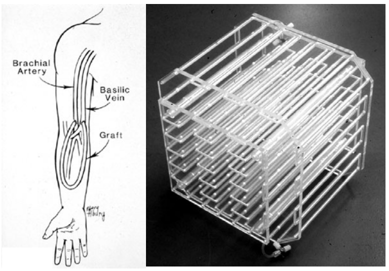

Figure 1: (a) Drawing of the forearm of a patient with a loop graft for hemodialysis, (b) Picture of the flow phantom.

Methods

Theory - PC flow values depend on both the reconstructed cross-sectional area of the vessel and the reconstructed blood velocity. The cross-sectional area is influenced by the linearity of the phase-encoding and read gradients and by the inhomogeneity of the main field [2], the velocity is scaled by the velocity-encoding gradient waveform [3]. A local background gradient as a result of ΔB_0 along the read direction scales the reconstructed cross-sectional area by a factor $\lambda_{\Delta B}^{-1}$ [4], with $\lambda_{\Delta B}(\mathbf{r}) = G_x(G_x+G_{\Delta B}'(\mathbf{r}))^{-1}$. Here, *G* denotes the applied gradient, $G_{\Delta B}'(\mathbf{r}) = \delta(\Delta B_0(\mathbf{r}))/\delta x$ the constant local gradient of the main field and *x* the read direction. A similar expression, $\lambda_G(\mathbf{r}) = G(G+G'(\mathbf{r}))^{-1}$, is found for a gradient nonlinearity *G'* in either the read (*x*), phase-encoding (*y*) or velocity-encoding (*ve*) direction, in which the value of *G'* is dependent on the gradient strength. Hence, to the first order, flow values will be scaled by the product of all factors: $Q_{measured}/Q_{real} = (\lambda_{\Delta B}\,\lambda_{Gx}\,\lambda_{Gy}\,\lambda_{Gve})^{-1}$, and observed flow values can be corrected once the λ's are known.

Determination of the spatial dependence of ΔB_o and G' - A single phantom consisting of 77 serially connected parallel 3-mm diameter tubes placed in a regular matrix and covering a volume of 336x336x336 mm^3 was used to determine the spatial dependence of ΔB_0 and *G'* and to perform flow measurements (Figure 1b). ΔB_0 and *G'* were mapped by determining the difference between real (x,y,z) and distorted (x_1,y_1,z_1) positions of tubes on 2D SE images [2]. Assuming G_y' to be proportional to G_y, which was verified using gradients with different strengths and inverse polarities, λ_{Gy} could be calculated as $\lambda_{Gy} = \delta y/\delta y_1$. $\lambda_{\Delta B}$ was determined by differentiation of the ΔB_0-map: $\lambda_{\Delta B} = G_x(\ \delta(\Delta B_0)/\delta x + G_x\ )^{-1}$. Note that this scaling factor is dependent on the read gradient strength of the 2DPC acquisition.

Flow – Steady flow, the same in every tube because of the serial connection, was generated with a constant pressure head of 1.9 m water. Timed collection of fluid served as control for PC flow measurements. The maximum velocity (± 25 cm/s) and T_E (20 ms) were chosen low enough to ensure that the measurements at a specific location were only affected by the local ΔB_0 and *G'* values ($v_{max}T_E \cong 5$mm).

MR sequences – All imaging was done on a 1.5 T scanner (Gyroscan, Philips, Best, The Netherlands). Transverse and sagital multislice 2D SE images were acquired such that the whole phantom volume was examined (9 slices of 20 mm, gap 22 mm, FOV 384 mm, MTX 256, G_x = 3.4 mT/m). 2DPC flow measurements were performed in the transverse plane (slice thickness 10 mm, FOV 384 mm, MTX 1024, T_R/T_E 31/20 ms, flip 40˚, Venc 30 cm/s, G_x = 4.5 mT/m, NEX 8). In some experiments without flow, T_E was varied to examine concomitant gradients effects; T_E indirectly influences concomitant gradient phase-offsets by means of limiting gradient timings and strengths [5].

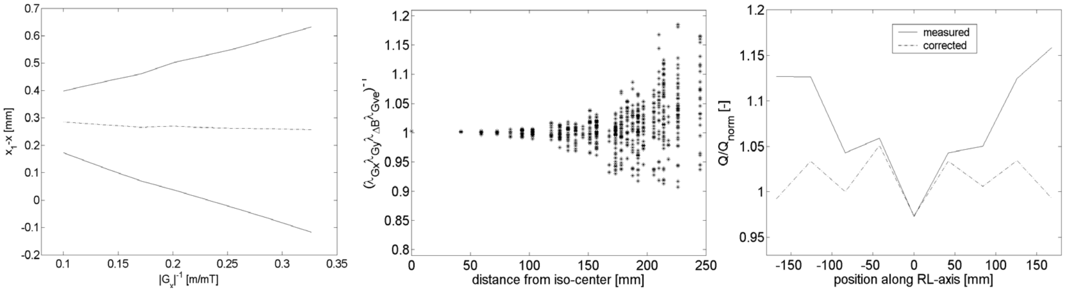

Figure 2: (a) Position error for different gradient strengths with negative and positive polarity (solid lines) and the mean of both (dotted line). (b) Flow measurement scaling factors as a function of off-center distance. (c) Flow values normalized with respect to flow measured by timed collection (Q_{norm} = 0.78 ± 0.02 ml/s) before (solid line) and after correction (dotted line).

Results

The positional error caused by ΔB_0 was found to be inversely proportional, and the error caused by *G'* constant with respect to the gradient strength (Figure 2a), indicating that *G'* is indeed proportional to |*G*|. An overview of the scaling parameters at all the grid points of the examined volume is depicted in Figure 2b. Flow, on average, tends to be more often overestimated than underestimated. Errors become increasingly important further off-center; they may be well over 10 % at off-centers larger than 18 cm and even 20 % at 23 cm. At shorter T_E's, we found significant effects of concomitant gradients, resulting in spatially dependent phase-offsets on top of the scaling due to gradient nonlinearity. Correction of the biased flow values was performed with the empirically determined scaling factors; accuracy improved significantly as is demonstrated for tubes in a row with a posterior offset of 126 mm in the transverse plane through the iso-center (Figure 2c).

Conclusions and discussion

We have shown that inhomogeneity of the main field and gradient nonlinearity have a significant effect on PC flow measurements; only a single phantom and two sequences are needed to analyze both phenomena and their mutual relation. Information of the spatial dependence of *G'* and ΔB_0 then allows correction of the observed PC flow values. Scaling errors are larger and increase faster further off-center. This can explain the difference in flow value that is sometimes observed between the venous and arterial limb in patients with a hemodialysis graft. Also, the observed overestimation of MRI data with respect to US dilution data may be related to *G'* and ΔB_0. However, depending on the location of the forearm in the bore, underestimations may occur as well. Therefore, a good comparison of MR and US dilution flow measurements requires an accurate description of the forearm position. The influence of ΔB_0 can be diminished by increasing the readout gradient strength, but at the expense of SNR and, hence, measurement precision. Gradient amplitude does not influence gradient scaling, but only plays a role in concomitant phase evolution. The influence of concomitant gradients can be reduced with longer echo times. In conclusion, the best position for flow quantification is close to the iso-center, but if this is impossible because of painful or exhausting poses for the patient, the appropriate scaling parameters can be used for correction.

References

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