Retrospective Correction for Induced Magnetic Field Inhomogeneity in Measurements of Large-Vessel Hemoglobin Oxygen Saturation by MR Susceptometry

M. Langham[1], T. Floyd[2], J. Magland[1], and F. W. Wehrli[1]

[1]Laboratory for Structural NMR Imaging, Department of Radiology, University of Pennsylvania School of Medicine, Philadelphia, PA, United States, [2]Department of Anesthesiology & Critical Care, Univesity of Pennsylvania School of Medicine, Philadelphia, PA, United States

Introduction

MR susceptometry-based blood oximetry [1, 2] relies on field mapping to measure the difference in magnetic susceptibility between intravascular blood and surrounding tissue. The main source of error in MR susceptometry is the static field inhomogeneity caused by an interface between air and tissue or between adjacent tissue types. High-pass filtering has previously been used in conjunction with shimming to reduce the effect of low spatial-frequency modulations of the phase caused by large-scale induced magnetic fields. We demonstrate that high-pass filtering is not optimum for MR susceptometry because the results are sensitive to filter size. We propose an alternative method which acquires data without scanner-implemented default shimming, and fit, after appropriate weighting and masking, the static field inhomogeneity to a second-order polynomial.

Methods

In MR susceptometry-based oximetry blood vessel is modeled as a long paramagnetic cylinder at some angle θ relative to the applied field B_o. The incremental field, ΔB, inside is given as $\Delta B = \frac{1}{2}\,\Delta\chi B_o(\cos^2\theta - 1/3)$, where $\Delta\chi = \Delta\chi_{do} Hct\cdot(1 - HbO_2)$ is the susceptibility difference (in SI units) between the intravascular blood and that of its surrounding muscle tissue, $\Delta\chi_{do} = 4\pi\cdot(0.27$ ppm) [3] is the susceptibility difference in SI units between fully deoxygenated and fully oxygenated erythrocytes, HbO_2 represents the fraction of the oxygenated hemoglobin (*Hb*) and hematocrit (*Hct*) is the fractional volume of the packed erythrocytes in the whole blood. The incremental field is obtained from the phase difference, $\Delta\phi_{map} = \gamma\Delta B\cdot\Delta TE$, where ΔTE is the time duration between two successive gradient echoes. The main source of error in MR Susceptometry is the static field inhomogeneity caused by the interface between air and tissue or between adjacent tissue types. We propose to remove the large-scale field inhomogeneity by fitting phase difference images to a second-order polynomial, i.e. we assume that the macroscopic field variation in phase difference images $\Delta\phi_{map}$ can be approximated as $\Delta\phi_{map} = c_1 1 + c_2 X + c_3 Y + c_4 XY + c_5 X^2 + c_6 Y^2$,where c_j's are the coefficients to be determined through the weighted least-squares fit. Before fitting the field inhomogeneity to a second-order polynomial, the vessels are masked out, and the background is suppressed by multiplying each pixel in the phase map with the corresponding value in the magnitude image. The corrected phase difference image is then given by $\Delta\phi_{corr} = \Delta\phi_{map} - \Delta\phi_{inhom}$. Written informed consent was obtained prior to all human studies following an institutional review board-approved protocol. Images were acquired with a multi-echo RF-spoiled GRE sequence programmed with SequenceTree Version 3.1 [4]. The pulse sequence includes fat suppression and first-moment nulling along the slice (i.e. blood flow) direction. The key parameters were: voxel size = 1 x 1 x 5 mm^3, FOV = 128x128 mm^2, resolution = 1 x 1 x 5 mm^3, BW = 488 Hz/pixel, TE = 4 ms, echo spacing = 2.5 ms, TR = 39 ms, flip angle = 13°, total scan time/image = 5 s. An eight-channel knee array coil (Invivo Inc., Pewaukee, WI) was used to acquire axial images containing the popliteal and femoral vessels from five volunteers with and without shimming for comparison. The raw k-space data was saved and processed offline. High-pass filtering and weighted least-squares fit were compared with blood oxygenation measurement in the femoral and popliteal vessels. Measurement of hemoglobin saturation in the femoral and popliteal artery or vein provides a simple consistency check since these are merely successive segments of the same vessel and no oxygen extraction takes place in between, therefore HbO_2 levels should be the same.

Results and Discussion

Typical magnitude and phase difference images of axial slices at the level of popliteal vessels are shown in **Figure 1.** In general, the static field inhomogeneity has no effect on the magnitude images (**Figure 1a**). In the phase difference images, the severity of the static field inhomogeneity is apparent even with shimming (**Figures 1b).** Measured oxygenation levels in femoral and popliteal vessels and their uncertainties from both methods are summarized in **Tables 1** and **2**. We note that the average discrepancies between blood oxygenation measurements in femoral and popliteal vessels are smaller by more than twice the standard error when shimming is disabled and retrospective weighted least-squares fit correction is implemented on the phase images. The effect of filter size in high-pass filtered images (derived from the data of **Figure 1b)** is illustrated in **Figure 2** and the resulting hemoglobin saturation levels are given in **Table 3**. The data show that the venous saturation measurement is critically dependent on filter size. The choice of filter size is determined by the desired level of uniformity in phase images. However, uniformity itself is subjective and beyond a critical filter size images become visually indistinguishable (**Figure 2** with filter size n > 10). Furthermore, the "critical" size is dataset-dependent and the extent to which the images are visually distinguishable depends on the window level.

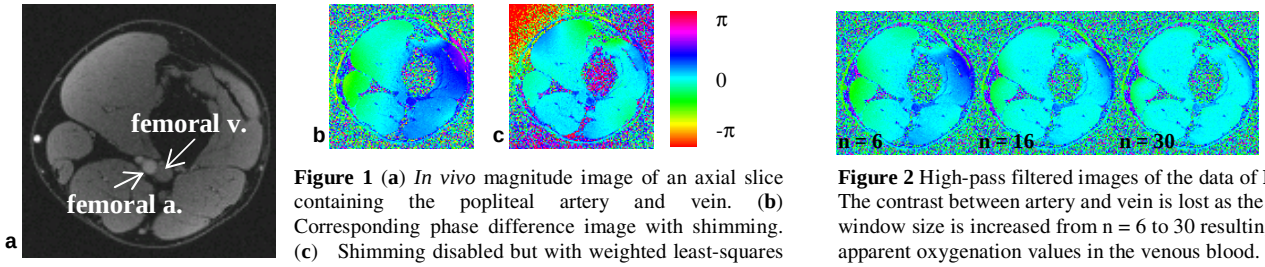

Figure 1 (**a**) *In vivo* magnitude image of an axial slice containing the popliteal artery and vein. (**b**) Corresponding phase difference image with shimming. (**c**) Shimming disabled but with weighted least-squares fit correction.

Figure 2 High-pass filtered images of the data of **Figure 1b**. The contrast between artery and vein is lost as the Hanning window size is increased from n = 6 to 30 resulting increased apparent oxygenation values in the venous blood.

Subject	Femoral A. $(S_AO_2 \pm \sigma_{SO_2})$ %HbO_2	Popliteal A. $(S_AO_2 \pm \sigma_{SO_2})$ %HbO_2	Femoral V. $(S_VO_2 \pm \sigma_{SO_2})$ %HbO_2	Popliteal V. $(S_VO_2 \pm \sigma_{SO_2})$ %HbO_2
1	99 ± 3	93 ± 3	59 ± 3	67 ± 3
2	98 ± 3	97 ± 3	56 ± 3	79 ± 3
3	92 ± 4	97 ± 4	54 ± 4	63 ± 4
4	91 ± 4	99 ± 4	54 ± 4	64 ± 4
5	97 ± 4	99 ± 3	68 ± 4	58 ± 3

Table 1 Hemoglobin oxygen saturation level from five healthy volunteers computed from axial images taken with shimming implemented by the scanner.

Subject	Femoral A. $(S_AO_2 \pm \sigma_{SO_2})$ %HbO_2	Popliteal A. $(S_AO_2 \pm \sigma_{SO_2})$ %HbO_2	Femoral V. $(S_VO_2 \pm \sigma_{SO_2})$ %HbO_2	Popliteal V. $(S_VO_2 \pm \sigma_{SO_2})$ %HbO_2
1	93 ± 3	93 ± 3	61 ± 3	60 ± 3
2	99 ± 3	99 ± 3	61 ± 3	63 ± 3
3	98 ± 4	95 ± 4	67 ± 4	62 ± 4
4	99 ± 4	97 ± 4	62 ± 4	64 ± 4
5	93 ± 4	96 ± 3	64 ± 4	61 ± 3

Table 2 Same as **Table 1** except that the oxygenation values were computed from the unshimmed images which were corrected retrospectively as described.

Filter Size	S_aO_2 (%HbO_2)	S_vO_2 (%HbO_2)
2	98	62
6	97	60
10	96	61
16	98	64
24	98	69
30	97	74

Table 3 Effect of filter size on the measured blood oxygenation.

Conclusion

High confidence in the retrospective correction technique is supported by the agreement between hemoglobin saturations measured in different segments of a vessel (femoral versus popliteal artery and vein) to within one standard error. Furthermore, the average derived blood oxygenation values are consistent with those reported in the literature and the coefficient of variation among the volunteers was 3% for both arterial and venous saturation levels. The weighted least-squares fit method removes the low spatial-frequency variations in phase images while the phase of interest remains unperturbed. On the other hand, high-pass filtering removes low-frequency variations indiscriminately without generally applicable criteria for choosing filter size, hence is not suited for MR susceptometry.

References: **[**1**]** Haacke et al. Human Brain Mapping 5:341-346 (1997); **[2]** Fernandez-Seara et al. MRM 55:967–973 (2006); **[3]** Spees et al MRM 45: 533 – 542 (2001); **[4]** Magland et al. Seattle, WA. Proc. ISMRM.2006, p 578.

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