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Computation of the gradient-induced electric field noise in 12-lead ECG traces during rapid MRI sequences

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Background

Successful physiological monitoring using a 12-lead ECG during MR imaging is essential for safe conduction of cardiovascular interventions within a MR scanner. However, ECG artifacts induced by magnetic field gradients severely affect the signal quality and fidelity. Previously, the gradient-induced artifacts were reduced by blocking ECG transmissions during all gradient ramps [1], which has been shown feasible while the method is not suitable for short-TR sequences. Theoretical and experimental

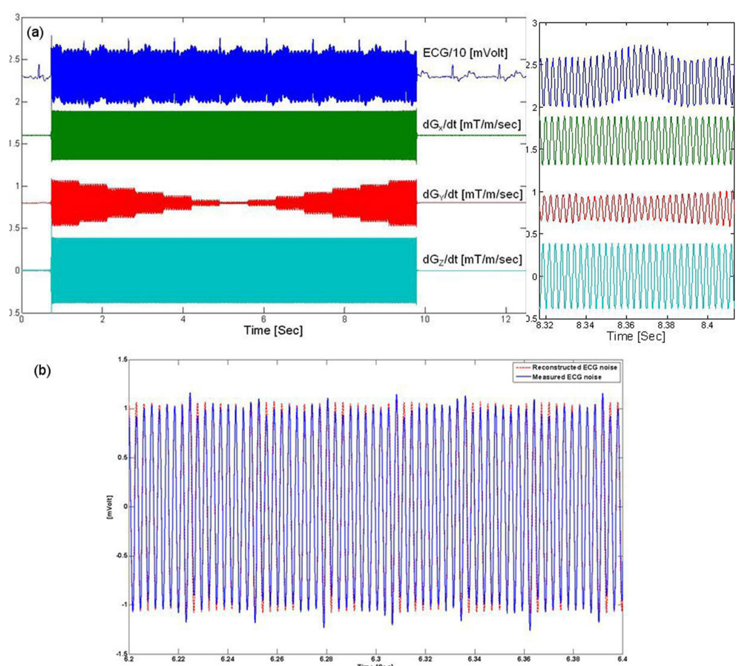

Figure 1 (a): One representative ECG trace (V5) and gradient waveforms derivatives from the X, Y and Z axes during a transverse (Z, or Superior-Inferior, slice encode) SSFP acquisition, with frequency encoding along × (Right-Left) and phase encoding along Y (Anterior-Posterior). A time domain magnification (right), shows the gradient derivative signals (only 0 to 500 Hz are shown). (b): Plots of measured ECG noise (solid blue) and computed ECG noise vector (dashed red) based on the fitted parameters. There is an 18% difference between the reconstructed and measured ECG noise.

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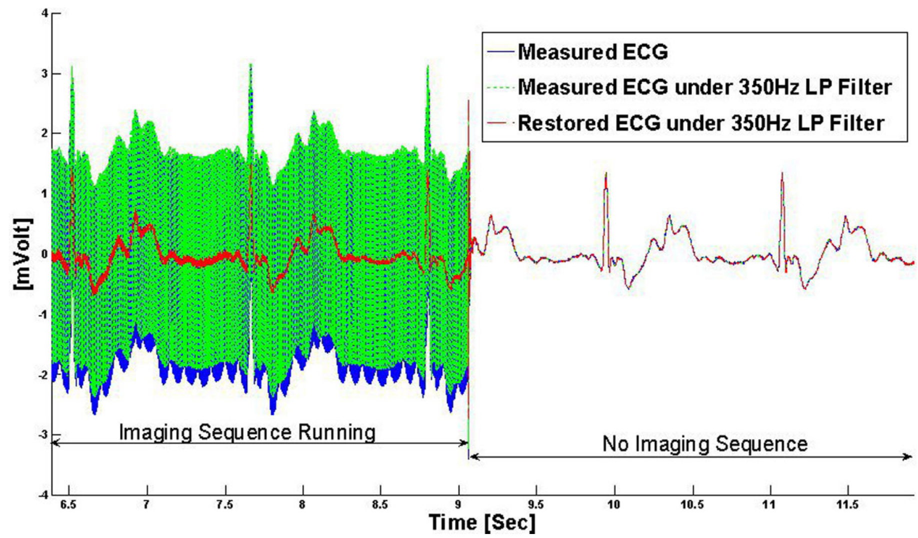
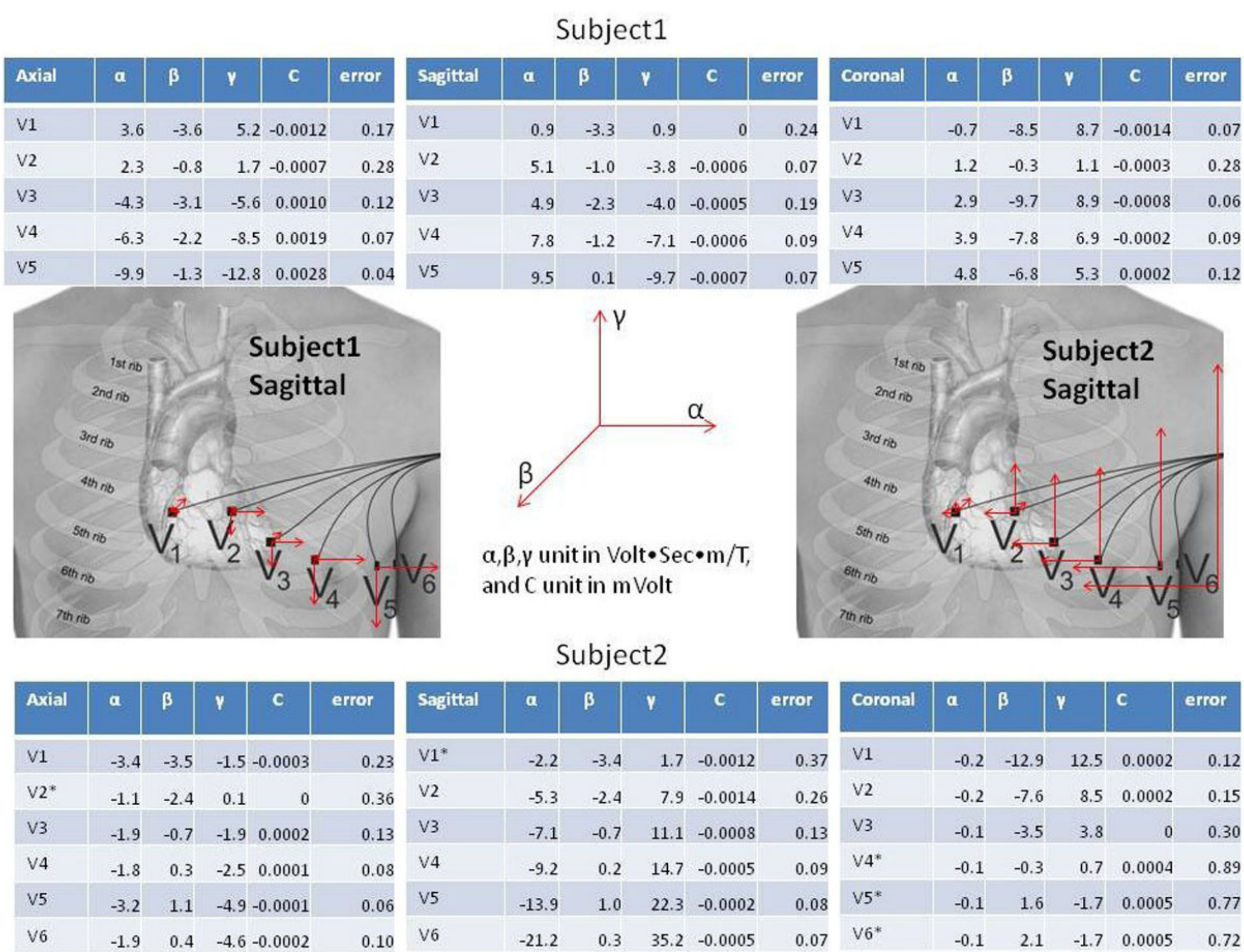


Figure 2 During Imaging, the restored ECG (red line) signal preserves the same signal shape as the ECG has in the absence of imaging (no gradient switching), while low-pass (LP) filtering (green dashed line) fails to clean the gradient-induced artifacts.



Subject1

Axial	α	β	γ	C	error
V1	3.6	-3.6	5.2	-0.0012	0.17
V2	2.3	-0.8	1.7	-0.0007	0.28
V3	-4.3	-3.1	-5.6	0.0010	0.12
V4	-6.3	-2.2	-8.5	0.0019	0.07
V5	-9.9	-1.3	-12.8	0.0028	0.04

Sagittal	α	β	γ	C	error
V1	0.9	-3.3	0.9	0	0.24
V2	5.1	-1.0	-3.8	-0.0006	0.07
V3	4.9	-2.3	-4.0	-0.0005	0.19
V4	7.8	-1.2	-7.1	-0.0006	0.09
V5	9.5	0.1	-9.7	-0.0007	0.07

Coronal	α	β	γ	C	error
V1	-0.7	-8.5	8.7	-0.0014	0.07
V2	1.2	-0.3	1.1	-0.0003	0.28
V3	2.9	-9.7	8.9	-0.0008	0.06
V4	3.9	-7.8	6.9	-0.0002	0.09
V5	4.8	-6.8	5.3	0.0002	0.12

Subject2

Axial	α	β	γ	C	error
V1	-3.4	-3.5	-1.5	-0.0003	0.23
V2*	-1.1	-2.4	0.1	0	0.36
V3	-1.9	-0.7	-1.9	0.0002	0.13
V4	-1.8	0.3	-2.5	0.0001	0.08
V5	-3.2	1.1	-4.9	-0.0001	0.06
V6	-1.9	0.4	-4.6	-0.0002	0.10

Sagittal	α	β	γ	C	error
V1*	-2.2	-3.4	1.7	-0.0012	0.37
V2	-5.3	-2.4	7.9	-0.0014	0.26
V3	-7.1	-0.7	11.1	-0.0008	0.13
V4	-9.2	0.2	14.7	-0.0005	0.09
V5	-13.9	1.0	22.3	-0.0002	0.08
V6	-21.2	0.3	35.2	-0.0005	0.07

Coronal	α	β	γ	C	error
V1	-0.2	-12.9	12.5	0.0002	0.12
V2	-0.2	-7.6	8.5	0.0002	0.15
V3	-0.1	-3.5	3.8	0	0.30
V4*	-0.1	-0.3	0.7	0.0004	0.89
V5*	-0.1	1.6	-1.7	0.0005	0.77
V6*	-0.1	2.1	-1.7	0.0005	0.72

Figure 3 The fitted parameters for three SSFP acquisition orientations are listed for Subject 1 (top) and subject 2 (bottom). The 3D vector plots in the center illustrate graphically sagittal acquisitions in both subjects utilizing phase-encoding along Y (Anterior-Posterior) for the precordial electrodes V1-V6. A gradually increasing influence of the magnetic gradient fields on the ECG noise was observed from V1 to V6.

studies have shown a linear relationship between electric fields and the temporal derivatives of the magnetic field gradients [2,3]. We propose an algorithm to restore the true ECG signal by subtracting system response functions, based on the MR gradient signals, from ECG signals distorted by gradient interference.

Methods

Data Acquisition: An MRI-conditional 12-lead ECG system [1] was used to acquire data on two healthy volunteers inside a 3T MRI. Outside the MRI room, high-fidelity ECG traces, along with the x, y and z gradient waveforms were digitally recorded simultaneously at 62kHz. Balanced SSFP sequences with various slice orientations (axial, coronal, sagittal and oblique) were acquired. Data Analysis: The gradient-induced ECG noise was computed as the difference between aligned ECG traces with and without MR sequence running. The noise voltage (Vni) at each electrode (i) was modeled as a linear combination of gradient derivatives and system factors, $Vni = \alpha i \bullet dGx/dt + \beta i \bullet dGy/dt + \gamma i \bullet dGz/dt + Ci$, where αi, βi, γi and Ci are position-dependent. These parameters were then used to reconstruct the noise, for comparison with the measured ECG noise, and to further derive the restored ECG.

Results

The recorded ECG traces and low-pass filtered gradient derivatives are displayed in Figure 1a. The computed noise vector (Vni) and the measured noise (Figure 1b) had differences of 21% ± 20% in normalized Euclidean distance. The restored ECG signal was comparable to the clean ECG segments (Figure 2), providing higher signal quality and fidelity relative to low-frequency filtering of the ECG signal. Vectorial display of the fitted parameters (Figure 3) demonstrated systematic changes across the precordial leads, and varied in magnitude between subjects.

Conclusions

The gradient-derivative model closely fit the measured ECG noise, possibly allowing for efficient gradient-noise removal utilizing rapid calibration scans, combined with hardware blocking of extremely high noise intervals.

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