

### **POSTER PRESENTATION**



# 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



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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.

							5	Subje	ct1								
xial	۵	β	Y	С	error	Sagittal	a	β	Y	C	error	Coronal	a	β	Y	C	error
1	3.6	-3.6	5.2	2 -0.0012	0.17	V1	0.9	-3.3	0.9	0	0.24	V1	-0.7	-8.5	8.7	-0.0014	0.0
2	2.3	-0.8	1.7	7 -0.0007	0.28	V2	5.1	-1.0	-3.8	-0.0006	0.07	V2	1.2	-0.3	1.1	-0.0003	0.2
3	-4.3	-3.1	-5.(	5 0.0010	0.12	V3	4.9	-2.3	-4.0	-0.0005	0.19	V3	2.9	-9.7	8.9	-0.0008	0.0
4	-6.3	-2.2	-8.5	5 0.0019	0.07	V4	7.8	-1.2	-7.1	-0.0006	0.09	V4	3.9	-7.8	6.9	-0.0002	0.0
5	-9.9	-1.3	-12.8	0.0028	0.04	V5	9.5	0.1	-9.7	-0.0007	0.07	V5	4.8	-6.8	5.3	0.0002	0.1
	4th rib	-	R	/	A	-	β					4th rib	F	1	1	1	-
St Sth. 7th n	th nb h nb h b	V <sub>1</sub>	V <sub>2</sub>	V <sub>3</sub> V	4 ¥5	V <sub>6</sub> α	β ,β,γ un nd C ur	iitin Vo nitin m Subje	olt•Se iVolt ct2	c∙m/T,		4th nb 5th nb 6th nb 7th nb	V <sub>1</sub>	V <sub>2</sub>	V3 1	4 V5	V <sub>6</sub>
St Sth 7th n Axial	th nb h nb b	V <sub>1</sub>	V <sub>2</sub>	c	error	ν 6 al Sagittal	β ,β,γ un nd C ur S	itin Vα nitin m Subje	olt•Se iVolt ct2 v	c∙m/T, c	error	4th nb 5th nb 6th nb 7th nb	ν V1	β	V3 1	4 V5	error
St Sth, 7th n Axcial	ath nb h nb b -3.4	β -3.5	V2 -1.5 -	c	error 0.23	Ca Ca Sagittal V1*	β ,β,γ un nd C ur S α -2.2	itin Vα nitin m Subje β -3.4	olt•Se iVolt ct2 v 1.7	c∙m/T, c	error 0.37	4th nb 5th nb 6th nb 7th nb	ч V1 -0.2	β -12.9	V 12.5	c	error 0.1
St Sth. 7th n Axial V1 V2*	ath nb h nb b -3.4 -1.1	β -3.5 -2.4	V 2 -1.5 - 0.1	C 0.0003	error 0.23 0.36	Contraction Contr	β ,β,γ un nd C ur S -2.2 -5.3	itin Vα nitin m Subje β -3.4 -2.4	olt•Se iVolt ct2 v 1.7 7.9	c∙m/T, c -0.0012 -0.0014	error 0.37 0.26	4th nb 5th nb 6th nb 7th nb Coronal V1 V2	a -0.2 -0.2	β -12.9 -7.6	v 12.5 8.5	c 0.0002 0.0002	error 0.1
St Sth, 7th n Axial V1 V2* V3	a 	β -3.5 -2.4 -0.7	¥ -1.5 - 0.1 -1.9	C 0.0003 0 0.0002	error 0.23 0.36 0.13	Ca a Sagittal V1* V2 V3	β ,β,γ un nd C ur <u>S</u> -2.2 -5.3 -7.1	it in Vo nit in m Subje β -3.4 -2.4 -0.7	olt•Se iVolt ct2 v 1.7 7.9 11.1	c∙m/T, c -0.0012 -0.0014 -0.0003	error 0.37 0.26 0.13	4th nb 5th nb 6th nb 7th nb Coronal V1 V2 V3	a -0.2 -0.2 -0.1	β -12.9 -7.6 -3.5	v 12.5 3.8	4 V5 c 0.0002 0.0002 0	error 0.1 0.3
Sth., 7th n Axial V1 V2* V3 V4	a -3.4 -1.1 -1.9 -1.8	β -3.5 -2.4 -0.7 0.3	V -1.5 - 0.1 -1.9 -2.5	C 0.0003 0.0002 0.0001	error 0.23 0.36 0.13 0.08	V1 <sup>*</sup> V2 V3 V4	β ,β,γ un nd C ur -2.2 -5.3 -7.1 -9.2	it in Vc hit in m Subje β -3.4 -2.4 -0.7 0.2	olt•Se Volt ct2 v 1.7 7.9 11.1 14.7	c•m/T, c -0.0012 -0.0014 -0.0008 -0.0005	error 0.37 0.26 0.13 0.09	4th nb Sth nb Sth nb Th nb Coronal V1 V2 V3 V4*	a -0.2 -0.2 -0.1 -0.1	β -12.9 -7.6 -3.5 -0.3	v 12.5 8.5 3.8 0.7	C 0.0002 0.0002 0 0.0004	error 0.1 0.3 0.8
50 50 50 70 n 70 n 70 n 70 n 70 n 70 n 70 n 70	a -3.4 -1.1 -1.9 -3.2	β -3.5 -2.4 -0.7 0.3 1.1	V2 -1.5 - 0.1 -1.9 -2.5 -4.9 -	C 0.0003 0 0.0002 0.0001 0.0001	error 0.23 0.36 0.13 0.08 0.06	α   Sagittal   V1*   V2   V3   V4   V5	β ,β,γ un nd C ur S -2.2 -5.3 -7.1 -9.2 -13.9	itin Vα hitin m Subje -3.4 -2.4 -0.7 0.2 1.0	olt•Se Volt ct2 v 1.7 7.9 11.1 14.7 22.3	c+m/T, c -0.0012 -0.0014 -0.0008 -0.0005 -0.0002	error 0.37 0.26 0.13 0.09 0.08	4th nb 5th nb 5th nb 7th nb Coronal V1 V2 V3 V4* V5*	a -0.2 -0.2 -0.1 -0.1	β -12.9 -7.6 -3.5 -0.3 1.6	V 12.5 8.5 3.8 0.7 -1.7	c 0.0002 0.0002 0 0.0004 0.0005	error 0.1 0.3 0.8 0.7

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,  $Vni = \alpha i \cdot dGx/dt + \beta i \cdot dGy/dt$ + $\gamma$ i•dGz/dt+Ci, where  $\alpha$ i,  $\beta$ i,  $\gamma$ i and Ci are positiondependent. 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  $21\% \pm 20\%$  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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