Characterization of the Magnetohydrodynamic Effect as a Signal from the Surface Electrocardiogram during Cardiac Magnetic Resonance Imaging

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Abstract

The magnetohydrodynamic (MHD) effect was investigated in the electrocardiogram (ECG) during MRI. ECGs were obtained for 10 healthy subjects inside and outside a 3.0T MRI scanner. The MHD effect was characterized as the difference in voltage between the mean beat from the ECG taken inside the magnet and the one taken outside the magnet. A significant deflection was observed and was found to primarily correspond to the flow of blood in the aortic arch. When subjects were in supine position, the mean amplitude of this deflection was 0.19 ± 0.04 mV, and its mean duration was 0.34 ± 0.02 secs; when in prone position, the mean amplitude of this deflection was 0.21 ± 0.08 mV, and its mean duration was 0.33 ± 0.02 secs. An axial orientation 2D phase contrast MRI scan was performed to allow blood flow and velocity measurements. By characterizing the MHD signal, we can obtain flow information and can develop improved subtraction techniques for ECG during MRI.

1. Introduction

When a conductive fluid travels through a magnetic field, there is a voltage induced in the direction orthogonal to both the magnetic field lines and the direction of flow of the fluid [1,2]. This phenomenon is called the magnetohydrodynamic (MHD) effect. The magnitude of the observed voltages produced by the MHD effect is determined by the velocity of flow of the fluid (v), the diameter of the tube through which the fluid is flowing (d), and the strength of the magnetic field (B). This induced voltage (U) is determined by the equation:

$$U = |B| * |v| \, d \sin \theta$$

where θ is the angle between the magnetic field lines and the direction of flow, and d is orthogonal to B and v.

During magnetic resonance imaging (MRI), the flow of blood through the body produces such voltages, which may be observed on the electrocardiogram (ECG) [3]. For ECG-gated cardiac MRI, the ECG is used to synchronize segmented data acquisition to the cardiac cycle. Typically acquisition is synchronized to the R-wave because this waveform has the largest magnitude in the normal ECG and precedes ventricular contraction. However, the presence of the MHD effect causes the ECG to be distorted. The ejection of blood from the left ventricle into the aorta does not occur simultaneously with electrical activation of the ventricles; therefore, there is a delay in the ECG before which the MHD effect can be observed. The MHD effect is typically observed during the S-T segment or at the beginning of the T-wave [4], which can potentially cause the magnitude of the T-wave to become greater than that of the R-wave. This may lead to incorrect R-wave detection and thus presents a problem for ECG-gated cardiac MRI. As scanners evolve from 1.5T to 3.0T in clinical use, the MHD effect only worsens the quality of the ECG. In addition, due to the MHD effect, the diagnostic usefulness of the ECG in the magnet is adversely affected. For example, if performing stress imaging studies, the ECG cannot be used as a surrogate measure since the S-T segment potential is unreliable.

Due to the difficulty of removing the MHD effect from the ECG, techniques have been developed to altogether avoid the use of the ECG in triggering [5,6]. Though these techniques result in sufficiently accurate gating, since they avoid producing a clean ECG by removal of the MHD effect, information which may be observed from the ECG is not provided. By characterizing the MHD signal, techniques to completely remove it from the ECG during MRI may be developed. Furthermore, since the MHD effect is related to the velocity of blood flow, by considering the MHD as a signal instead of noise, information about the flow itself may be observed, which may be useful clinically.

2. Methods

The MR-compatible ECG lead system of the 3.0T Siemens TRIO scanner used in this study utilized a pair of electrodes oriented horizontally on the chest. This orientation is optimal since the magnetic field lines are oriented vertically across the body while in the magnet, so the electrode pair is orthogonal to the magnetic field lines. Furthermore, in this orientation the direction of blood flow through the aortic arch is roughly orthogonal to both the magnetic field lines and the electrode pair. The ECG electrodes were centered on the fourth intercostal space on the left sternal border, with an inter-electrode spacing of approximately 10 cm.

The raw ECG signals were obtained with a sampling rate of 100 Hz. All recordings were taken for thirty seconds during breath holds in order to reduce motion artifacts due to respiration. The ECG was recorded in two different body positions, supine and prone, such that the MHD effect could be observed when the body and thus the blood flow were in two alternate orientations. The ECG was recorded with the subject outside the magnet, inside the magnet without scanning, and inside the magnet during conventional gradient echo cine MRI scan.

MATLAB code was developed to accomplish the signal processing tasks. A fifth order highpass elliptical filter with cutoff frequency 0.5 Hz was used for the removal of baseline wander. QRS complexes were detected using a modified version of the Pan and Tompkins algorithm in order to align the beats to determine the mean beat [7]. Template matching was used to remove aberrant beats, and a representative mean beat was then generated. The mean beat of the ECG obtained outside the magnet was then subtracted from the mean beat of the ECG taken inside the magnet to extract the MHD signal.

In addition to the ECG recordings, an axial orientation 2D phase contrast (PC) MRI scan was performed to allow blood flow and velocity measurements in order to compare the timing with that of the MHD signal. Following each PC scan, Argus post-processing software (SMS) was used to draw region-of-interest within the lumen of the descending aorta and measure mean flow and mean velocity throughout the cardiac cycle. The PC MRI scans and the ECGs were recorded independently.

3. Results

3.1. Subjects

The healthy subjects with no known heart disease in this study consisted of four females and six males ranging in age from 20 to 31 years old, with a mean age of 24. The average resting heart rate for the subjects ranged from 46 to 76 beats per minute, with a mean of 60 beats per minute. The signals were usable in 9 out of the 10 subjects. In the unusable case, the signals were corrupted by noise and were subsequently unrecoverable.

3.2. ECG signal characteristics

The ECGs recorded outside the magnet varied substantially from those taken inside the magnet. In all nine subjects, the ECGs obtained outside the magnet had clearly defined P-waves, QRS complexes, and T-waves. However, the ECGs recorded inside the magnet both with and without scanning had deflections in the P-waves, S-T segments and T-waves. These deflections were consistent across all the ECG signals taken inside the magnet.

3.3. MHD signal characteristics

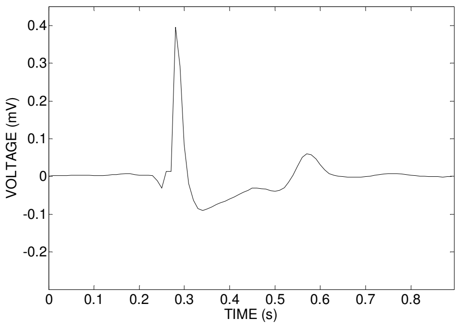
The mean beats for a sample subject outside and inside the magnet without scanning are shown in Figure 1 for supine position and in Figure 2 for prone position. In addition, the subtractions for this subject (inside magnet – outside magnet) are shown in these figures.

Although the subtractions for all the subjects had similar shapes, there were subtle differences in each. The most notable characteristic in the subtractions is a significant deflection following the QRS complex. In the supine position, the maximum amplitude of this deflection ranged from 0.13 to 0.27 mV (mean $\pm$ standard deviation, 0.19 $\pm$ 0.04 mV). Its duration ranged from 0.32 to 0.38 seconds (0.34 $\pm$ 0.02 seconds). In the prone position, the maximum amplitude of this deflection ranged from 0.14 to 0.30 mV (0.21 $\pm$ 0.08 mV). Its duration ranged from 0.30 to 0.36 seconds (0.33 $\pm$ 0.02 seconds). In both the supine and prone positions, the delay between the onset of the QRS complex and the onset of the MHD deflection ranged from 0.11 to 0.19 seconds (0.14 $\pm$ 0.02 seconds).

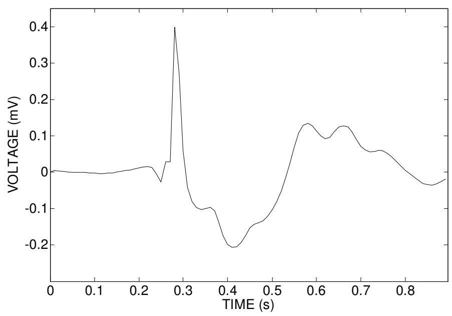
ECG signals were also recorded during conventional gradient echo cine MRI scans. A similar procedure was followed with these signals in generating a mean beat and then subtracting the mean beat generated while outside the magnet to extract the MHD signal. Minimal differences were observed between this subtraction and the one previously described which did not involve scanning. The correlation between the two subtractions ranged from 97.8% to 99.4% among the subjects.

3.4. Comparison with MRI data

In Figures 3 and 4, the MHD signal is shown with the MRI flow data superimposed for supine and prone positions, respectively. There is a delay between the deflection observed in the MHD signal and the deflection seen in the flow data obtained from the MRI scan. The difference in timing between the minima for both curves ranged from 0.02 to 0.09 seconds (0.06 $\pm$ 0.02 seconds). Velocity data were also obtained; the difference in timing between the minima for these curves and the minima of the MHD curves ranged from 0.01 to 0.05 seconds (0.03 $\pm$ 0.02 seconds).



(a)



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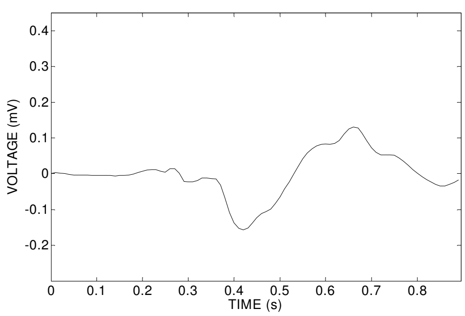
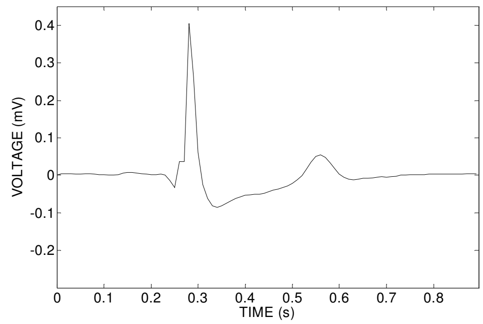
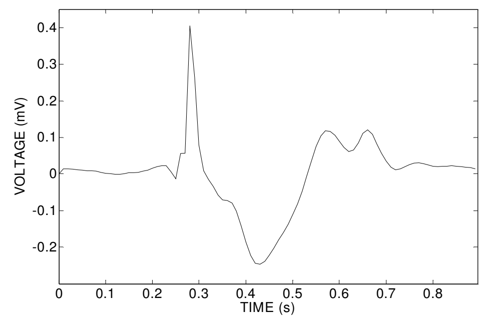


(c)

Figure 1: Supine (a) Mean beat for ECG outside magnet, (b) Mean beat for ECG inside magnet, (c) Subtraction of mean beats inside – outside magnet



(a)



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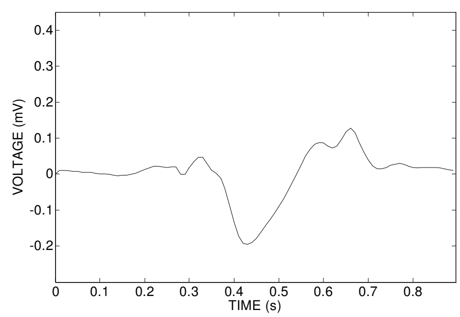

(c)

Figure 2: Prone (a) Mean beat for ECG outside magnet, (b) Mean beat for ECG inside magnet, (c) Subtraction of mean beats inside – outside magnet

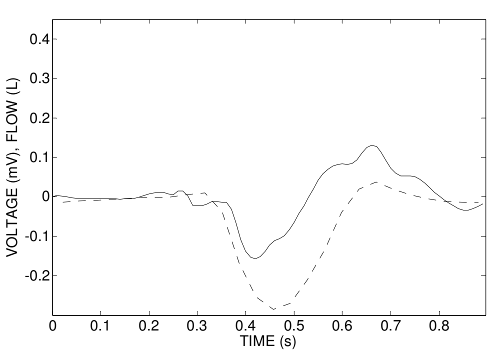

Figure 3: Supine: Subtraction of mean beats inside magnet – outside magnet (solid line) and MRI flow data (dotted line)

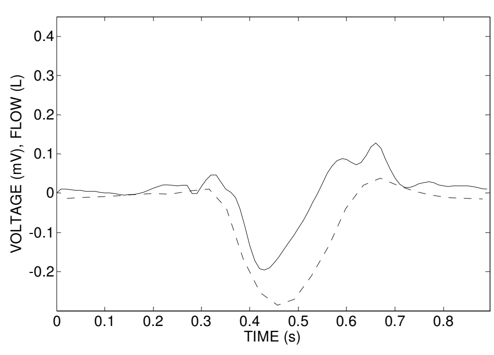

Figure 4: Prone: Subtraction of mean beats inside magnet – outside magnet (solid line) and MRI flow data (dotted line)

4. Discussion and conclusions

The MHD signal is the summed effect of the flow occurring at a particular instant. Therefore, from this signal alone, it is not possible to learn about flow through a particular vessel, but rather, net flow. However, there are several reasons why the primary contributor of the MHD effect observed in this study is the blood flow in the aortic arch. First, the velocity of blood flow is greatest in the aorta. Second, the aorta is the largest diameter blood vessel. Finally, the blood flow through the aortic arch occurs approximately perpendicular to both the electrode pair and the magnetic field lines, which subsequently produces the greatest MHD effect.

The MHD effect may be characterized as the difference in voltage between the mean beat of the ECG taken inside the magnet and the mean beat of the ECG taken outside the magnet. As expected, the amplitude and duration of the MHD signal were not significantly different between supine and prone positions since the aortic arch is roughly perpendicular to the magnetic field lines and electrode pair in both orientations, resulting in a similar MHD effect. The delay observed between the onset of the QRS complex and the onset of the MHD deflection was consistent with the expected delay due to the electromechanical delay, pre-ejection period, and time for the flow to propagate into the aortic arch [8].

The subtractions involving gradient echo cine MRI scans were very similar to the subtractions not involving scanning. This is reasonable since the voltages due to the MHD effect are directly related to the strength of the magnetic field. The 3.0T static magnetic field is several orders of magnitude larger than the gradient magnetic fields used for imaging. Thus, the induced voltages observed on the ECG are also several orders of magnitude smaller than those resulting from the uniform magnetic field and do not subsequently result in substantial changes in the morphology of the subtracted signal. Furthermore, other sources of interference were not noticeable since the radiofrequency interference was removed by filtering and the switching MR gradient artifacts were minimal due to an optimized lead system.

The flow data obtained through axial orientation 2D PC MRI scan helps to verify that the MHD signal observed from the ECG was in fact a flow-related effect. This was due to the correlation in timing between the two observations of flow. The velocity data also correlated in time with the MHD signal, further indicating that the MHD signal was flow-related. Though the MRI data has limited temporal resolution, it was presented here as a basis of comparison with the ECG data. In fact, the coarse time resolution of 50ms for the MRI flow and velocity data may explain the observed delays. The PC scan on each subject was performed separately from ECG recording because the scan itself would affect the acquired ECG signals.

The motivation behind this work was twofold. First, by characterizing the MHD signal, it may provide additional information about blood flow. Second, with this characterization, it may prompt additional methods for its removal during MRI, producing a diagnostically useful ECG signal. This goal is desirable clinically for simultaneous examination of electrical activity via ECG and anatomical information via MRI. Furthermore, having a clean ECG during MRI can result in better triggering, which in turn can result in better MR images of the heart. In addition, even though MRI is capable of generating flow information, which was actually used in the study to verify the results, an advantage of ECG over MRI is that the ECG has significantly higher temporal resolution than MRI.

Acknowledgements

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References

[1] Togawa T, Okai O, Oshima M. Observation of blood flow e.m.f. in externally applied strong magnetic field by surface electrodes. Med Biol Eng 1967 Mar;5(2):169-70.

[2] Dimick RN, Hedlund LW, Herfkens RJ, Fram EK, Utz J. Optimizing electrocardiograph electrode placement for cardiac-gated magnetic resonance imaging. Invest Radiol 1987 Jan;22(1):17-22.

[3] Tenforde TS. Magnetically induced electric fields and currents in the circulatory system. Prog Biophys Mol Biol 2005 Feb-Apr;87(2-3):279-88.

[4] Tenforde TS, Gaffey CT, Moyer BR, Budinger TF. Cardiovascular alterations in Macaca monkeys exposed to stationary magnetic fields: experimental observations and theoretical analysis. Bioelectromag 1983;4(1):1-9.

[5] Fischer SE, Wickline SA, Lorenz CH. Novel real-time R-wave detection algorithm based on the vectorcardiogram for accurate gated magnetic resonance acquisitions. Magn Reson Med 1999 Aug;42(2):361-370.

[6] Crowe ME, Larson AC, Zhang Q, Carr J, White RD, Li D, Simonetti OP. Automated rectilinear self-gated cardiac cine imaging. Magn Reson Med 2004 Oct;52(4): 782-8.

[7] Pan J, Tompkins WJ. A real-time QRS detection algorithm. IEEE Trans Biomed Eng 1985 Mar;32(3):230-6.

[8] Guyton AC, Hall JE. Textbook of medical physiology. 10th Ed. W.B. Saunders Co. Philadelphia, PA: 2000.

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